Liege University

Faculty of Medecine

How to identify future fallers among older adults, based on gait patterns and using data mining

Sophie Gillain, MD
Geriatric Department
University Hospital of Liege

Thesis presented for the achievement of the academic degree of Doctor in medical sciences
2018-2019

How to identify future fallers among older adults, based on gait patterns and using data mining ?

Sophie Gillain

Medical doctor in the geriatric department at University Hospital of Liège, Belgium

Directed by

Pr. Jean PETERMANS and Pr. Gaëtan GARRAUX

Thesis Jury

Jacques Verly, Olivier Brüls, Jean-Louis Croisier, Eric Salmon, Jean-Yves Reginster, Olivier Bruyère, Patrick Manckoundia, Ivan Bautmans, Gaëtan Garraux and Jean Petermans

Thesis presented for the achievement of the academic degree of

Doctor in medical sciences

2018-2019

This research was supported by
the Fonds National de la Recherche Scientifique (FRS-FNRS)
and the University of Liège (ULiège)

Remerciements

En rédigeant ces quelques lignes, j'ai le plaisir de me remémorer l'ensemble des personnes ayant participé à la réalisation de ce travail. C'est une grande joie et une réelle fierté d'avoir pu vous compter sur vous tout au long de ces années. Je voudrais tous vous en remercier chaleureusement.

Tout d'abord, aux volontaires ayant participé à ce travail, MERCI. Votre enthousiasme et votre implication ont été un réel atout pour ce travail de recherche clinique.

Ensuite je souhaite remercier les collègues qui m'ont permis de mener ce travail de recherche malgré les difficultés d'organisation de l'activité clinique; les docteurs Allepaerts, Christelbach, Doupagne et Weber. Je remercie plus particulièrement les docteurs Christelbach et Ricour qui ont aidé au recrutement des volontaires, les docteurs Geurten et Gerontitis qui m'ont aidé à l'acquisition des données et le docteur Xavier Schmitz et mademoiselle Christina Léonard qui ont réalisé les évaluations neuropsychologiques. Bien entendu, je remercie chaleureusement, Madame Vinciane Wojtasik sans qui le recrutement et le suivi des volontaires n'aurait pu être possible. Ta disponibilité, ton expérience et ton enthousiasme ont été précieux. Enfin, je souhaite remercier Mesdames Véronique Jonart et Isabelle Narinx pour leur amitié et soutien sans faille.

Au sein de l'équipe de recherche du Cyclotron, je tiens à remercier particulièrement le Dr Frédérique Depierreux et Madame Evelyne Balteau pour leurs conseils lors de l'apprentissage de la manipulation de l'IRM et messieurs Bahri, Degueldre et Philips pour leur soutien logistique et méthodologique lors de l'analyse des IRM. Merci à Dorothée, Sophie, Julia et Christina pour leur écoute et leur soutien.

Je remercie les Professeurs Francis Perée et Stéphane Adam pour leurs conseils méthodologiques. Je remercie le Professeur Moustapha Dramé du CHU de Reims, rencontré sur les bancs de l'EAMA pour son accueil à Reims, son soutien, ses conseils méthodologiques et sa gentillesse. Merci à tous les trois pour ces échanges enrichissants.

En outre, je remercie les docteurs Marie Demonceau, Cédric Schwartz et Mohamed Boutaayamou pour votre aide lors de la réalisation de ce travail. Votre gentillesse et votre bonne

humeur n’ont d’égal que votre disponibilité et votre compétence. Quel plaisir d’avoir travaillé avec vous.

Merci aussi à Nadia Dardenne pour les analyses statistiques réalisées et pour ta patience à mes côtés dans la recherche de l’utile. Merci à Fiona pour ses conseils tout au long de la rédaction de ce travail.

Je remercie les professeurs Brüls, Bruyère, Croisier, Reginster, Salmon et Verly pour leur soutien et leurs conseils. Votre disponibilité et vos compétences m’ont permis de réaliser le travail de recherche multidisciplinaire que je souhaitais. Je remercie plus particulièrement le Professeur Gaëtan Garraux pour son soutien méthodologique et sa disponibilité.

Bien entendu, je ne remercierai jamais assez le Professeur Jean Petermans qui, depuis le début, a encouragé et soutenu ce projet. Cet aboutissement est également le fruit de votre travail et de votre implication dans la recherche. Ce qui suit n’existerait pas sans vous et j’espère que vous en serez satisfait.

Et comme sans racine, pas de croissance, je voudrais remercier mes parents, Tatie et Tonton pour m’avoir appris l’invisible avec les yeux. Je remercie aussi, Andrée et Louis de me soutenir au quotidien.

Enfin je remercie l’homme qui partage ma vie depuis tant d’années. Tu m’as encouragé dès le début et « supporté » dans tous les sens du terme. A chaque étape, tu as répété les mêmes conseils bienveillants et, le cas échéant, tu as su me consoler avec un bon repas. Tu m’as offert le luxe d’une vie de couple et de famille malgré le travail à accomplir.

Summary

Background: Falls among older adults are a major public health problem. One of the main challenges is identifying older adults at risk of future fall(s) before the first event. While the assessment of gait patterns could be useful in this context, no consensual literature currently exists concerning the gait parameters and walking conditions that are useful for identifying, among older adults, those who are at risk of future fall(s).

Objective: The main goal of this exploratory study was to examine the usefulness of gait patterns assessed in comfortable and in challenging walking conditions, in order to identify, among older adults, those at risk of future fall(s).

Materials and methods: A two-year, longitudinal, observational study among community-dwelling adults older than 65 years, living independently at home and without a recent fall history, was conducted between July 2014 and February 2017. At inclusion, all participants underwent comprehensive geriatric assessment and gait analysis recording gait speed, stride length, frequency, symmetry and regularity and minimal toe clearance using accelerometer-based and opto-electronic methods in comfortable, fast and dual task walking conditions. Fall(s) during follow-up were self-recorded in a personal notebook and later recorded by the research team via phone contact at 3 months. Comparisons and regression analyses were performed. Additionally, data mining software including a classification tool (J48) was used to better understand the relationship between gait patterns and the future fall(s) risk.

Results: One hundred and five participants were included; two-year follow-up was available for 96 (91.4%); 35 participants fell at least once during follow-up. Comparative analysis showed that future fallers had longer stride length in the fast walking condition (after adjustment for leg length) and higher stride symmetry in the dual task walking condition than non-fallers. Regression analysis showed higher stride symmetry cost was significantly and

independently associated with higher future fall(s) risk. Moreover, the use of the J48 yielded a classification tree able to identify 80% of future fallers based on the stride symmetry dual task walking cost, the fast walking stride length, the stiffness and MTC mean and variability measures.

Conclusion: This exploratory study shows the usefulness of data mining to understand the nonlinear relationships between gait patterns and future fall(s) risk, and the utility of considering clinical characteristics in association with gait patterns assessed in challenging walking conditions, in order to identify, among older adults, those who are at risk of future fall(s). While these results are encouraging, further research is warranted to confirm our results and to improve the identification of older adults at risk of future fall(s).

Résumé

Introduction : La chute est un problème majeur de santé publique et l'identification précoce des chuteurs reste un défi. Bien que l'analyse instrumentale de la marche semble être utile, aucun consensus ne précise les paramètres ou les conditions de marche à considérer lors de l'identification des sujets âgés à risque de chute.

Objectif : L'objectif de ce travail est de préciser l'intérêt des paramètres de marche mesurés lors de différentes conditions de marche lors de l'identification précoce des sujets âgés à risque de chute.

Matériel et méthode : Entre juillet 2014 et septembre 2015, 105 volontaires âgés de 65 ans et plus, autonomes au domicile et sans antécédents de chute, ont été inclus dans une étude longitudinale observationnelle comprenant un relevé prospectif des chutes durant deux ans. A l'inclusion, et après une évaluation gériatrique complète, la vitesse de marche, la longueur, la cadence la régularité et la symétrie des pas et la distance minimale entre le gros orteil et le sol lors de la phase de swing ont été enregistrés en marche de confort, en marche rapide et en tâche double. Les volontaires ont reçu un cahier de suivi des chutes et un contact téléphonique trimestriel a permis de collecter les évènements. Après deux ans de suivi, les sujets ont été classés en chuteurs et non-chuteurs sur base des évènements relevés. Des analyses de comparaison et de régression ont été réalisées. Enfin, un outil de classification (J48) a été appliqué aux données relevées à l'inclusion.

Résultats : Parmi les 105 volontaires inclus, 96 sujets ont été suivis à deux ans dont 35 ont présenté au moins une chute au cours du suivi. Les analyses de comparaison ont montré que dès l'inclusion, les sujets chuteurs présentaient une longueur des pas en marche rapide plus courte (y compris après ajustement à la taille de la jambe) et une diminution de la symétrie des pas en condition de tâche double plus importante que les sujets non chuteurs lors du suivi. Enfin, l'application du J 48 a permis d'obtenir un outil de classification permettant d'identifier 80 %

des futurs chuteurs sur base du coût de la symétrie des pas en double tâche, de la longueur des pas en marche rapide, de la rigidité, et des valeurs moyennes et de la variabilité de la distance minimale entre le gros orteil et le sol lors de la phase de swing.

Conclusion : Cette étude exploratoire, dont les résultats sont à vérifier au sein d'un échantillon plus large, suggère que l'utilisation d'un outil de classification permet de considérer le profil individuel de marche qui, étudié dans différentes conditions de marche et associé à des données cliniques, permet l'identification précoce des sujets âgés à risque de chute.

Contents

1. Introduction

According to the World Health Organization (WHO), falls are the second leading cause of deaths from accidental or unintentional injury worldwide. Indeed, WHO estimates that each year, approximately 646 000 individuals die from falls, with adults aged over 65 years suffering the greatest number of fatal falls. Moreover, 37.3 million falls severe enough to require medical attention occur every year(WHO 2018).

In European countries, the population of older adults is growing. In Belgium, the declining birth rate, increasing life-expectancy and the consequences of the baby boom generation, now aged over 40 years, mean that by 2030, one third of Belgian adult will be older than 66 years (M. Vandresse 2017). In this context, the prevalence of pathological conditions linked to ageing is growing and, among these, falls represent major issue. Indeed, even in community-dwelling adults, the prevalence of falls is around 30% per year (Tinetti , Speechley et al. 1988, Watson, Clapperton et al. 2011, Craig, Murray et al. 2013, Morrison, Fan et al. 2013, Sun, Huang et al. 2016), reaching up to 45% per year in the study by Delbaere et al. (Delbaere, Close et al. 2010).

Furthermore, falls among community-dwelling older people are known to be a cause of injury, disability, functional decline, decreased quality of life and ultimately, death (Rubenstein 2006, Thiem, Klaaßen-Mielke et al. 2014). Additionally and even in the absence of injury, falls lead to a fear of falling, which in turn is associated with an increased fall risk (Scheffer, Schuurmans et al. 2008), and reduced social or physical activities (Delbaere, Crombez et al. 2004, Choi and Ko 2015), leading to functional decline.

In this context, the health care costs linked to falls have to be considered. Several studies investigating the direct healthcare costs linked to fatal and/or non-fatal falls have been

published, and although absolute amounts are difficult to compare between studies (due to differences in data sources, national health care organisations, national healthcare resource allocations, individual financial resources to allocate to healthcare, etc.), most of the studies agree that the direct healthcare costs related to falls are especially incurred in higher age groups, in women, when bone fractures or head injury occur, and in hospitals and long-term care (Heinrich, Rapp et al. 2010, Watson, Clapperton et al. 2011, Craig, Murray et al. 2013). Indeed, according the study published by Heinrich et al., the costs linked to falls among community dwelling older adults were between 0.85% and 1.5% of total health care expenditure in Europe, the USA, United Kingdom and Australia, meaning falls are a major public health concern (Heinrich, Rapp et al. 2010). According to the same authors the cost per fall victim ranges from 2,044 to 25,955 US dollars (purchasing power parity) depending on the fall severity. However, these amounts do not take into account of the indirect costs related to the loss of income (for the patient and for the caregiver) and the costs related to the potential functional decline or disability. Unfortunately, to the best of our knowledge, there are no studies to date that have published robust data concerning the amount of these indirect costs.

In summary, considering the social and individual burden linked to falls among older adults, and as underlined by the WHO, “Prevention strategies should emphasize education, training, creating safer environments, prioritizing fall-related research and establishing effective policies to reduce risk”.

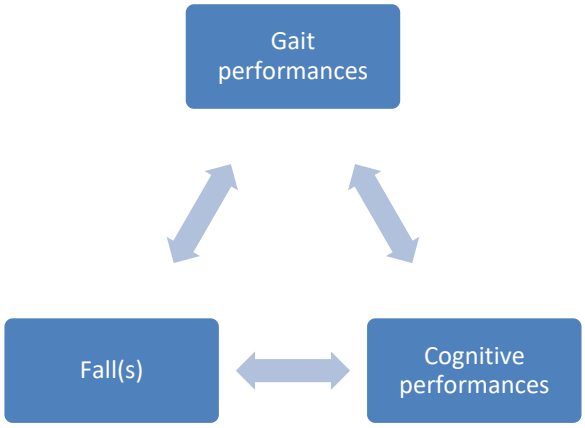
Among older adults, several processes are linked to an increased fall risk. First, specific diseases increase the fall risk, such as Parkinson's disease, stroke or atrial fibrillation for example. In these cases, the fall risk is most often recognized and the physician may implement strategies to decrease fall risk, by focusing, for example, on the daily-life environment, use of walking aids, shoe or drug review, detection and treatment of osteoporosis, and

physiotherapeutic care. Otherwise, non-disease-specific health factors may also increase the fall risk, such as pain or sleep disorders. Furthermore, several physiological ageing processes (e.g. reduced visual acuity, vascular burden or sarcopenia) and geriatric syndromes (e.g. mood or cognitive disorders, incontinence and fall(s) themselves) can also increase the fall risk. Finally, socio-demographic characteristics seem to be linked to risk of falls, such as female gender, exposure to toxins, social isolation and the daily-life environment (Tinetti , Speechley et al. 1988, Rubenstein 2006). In summary, the risk of falls increases with older age, affects individuals in different ways, occurs in different conditions and leads to different consequences. Even among community dwelling older adults, fallers represent a heterogeneous population.

Regarding the potential consequences of falls, one of the main issues is to detect, among healthy old people, those who are at risk of a fall before the first fall occurs. In this context, more comprehensive knowledge of the mechanisms underlying gait could help clinicians to detect older adults at risk of falls. Studies focusing on fall risk among older adults have highlighted a relationship between gait performance and fall history (Hausdorff, Edelberg et al. 1997, Auvinet, Berrut et al. 2003), and have also suggested that some gait parameters could be used as markers of fall risk (Hausdorff, Rios et al. 2001, Verghese, Holtzer et al. 2009, Bautmans, Jansen et al. 2011).

Recent literature suggests that studies of gait performances and underlying gait mechanisms need to assess gait parameters using instrumental methods and better understand the neural structures involved in central locomotor commands (Gillain and Petermans 2013). In fact some correlations between gait parameters and data related to brain imagery have been already shown (Annweiler and Montero-Odasso 2012, Annweiler, Beauchet et al. 2013, Annweiler, Beauchet et al. 2013, Annweiler, Montero-Odasso et al. 2014) and recent literature suggests that gait performance, falls and cognitive performance could form a triangular relationship in which gait and cognitive performances could be linked (Hausdorff, Yogev et al. 2005) and used as a marker

of fall risk (Maki 1997, Beauchet, Annweiler et al. 2009, Ambrose, Paul et al. 2013, Gillain, Boutaayamou et al. 2018) and to investigate cognitive decline (Scherder, Eggermont et al. 2007, Gillain, Warzee et al. 2009, Gillain, Dramé et al. 2015, Allali, Launay et al. 2017).



In fact, our department has previously used instrumental method to discern people with cognitive decline based on their gait patterns (Gillain, Warzee et al. 2009, Gillain, Dramé et al. 2015). At the same time, other teams have confirmed that investigating gait parameters could be useful for the assessment of the cognitive profile in older adults (Scherder, Eggermont et al. 2007, Hausdorff, Schweiger et al. 2008, Scherder, Eggermont et al. 2011, Allali, Ayers et al. 2016).

Based on previous literature concerning the relationships between brain structure, gait parameters and falls, and in light of our encouraging preliminary results concerning the relationship between gait parameters and cognitive performance, it therefore appears logical to study gait parameters using an instrumental method, and to study neural brain structures with a view to earlier identification of persons at risk of fall and of cognitive decline. We conducted a two-year longitudinal study including persons aged over 65 years, living independently in their own home, and free from pathological processes known to disturb gait or cognitive performance. At inclusion, volunteers underwent comprehensive geriatric assessment, gait

analysis, cognitive assessment and structural brain imagery. Unfortunately, at the time of writing, the results related to the cognitive follow-up are not fully available. Thus, the results presented here will focus on fall incidence and its relationship with data related to clinical characteristics, functional performance or gait patterns. In view of the focus of this work, the background and methods sections will not further discuss aspects related to neuropsychological assessment and brain imagery. Nonetheless, in order to provide a comprehensive overview of the data acquired in this research work, brain imagery and neuropsychological assessment will be described briefly in the experimental section.

2. Theoretical background

This section will present the main definitions, concepts, tools, methods and parameters related to the relationships between fall(s) and gait parameters. After a brief explanation of the main terms, risk factors for falls and the methods used to assess fall risk will be presented. Moreover, the instrumental methods used to study gait patterns will be briefly overviewed and the choice of the method used in this research work will be justified. Finally, the gait parameters usually available using the two instrumental methods used in this work, and their relationship with fall history and/or fall risk, will be discussed.

2.1.1. Walk

The normal human walking can be defined as “a method for locomotion involving the use of two legs, alternately, to provide both support and propulsion”. In order to exclude running, the definition should include “at least one foot being in contact with the ground at all times” (Levine, Richards et al. 2012). Walking was previously described as an automatic, rhythmic and regular motor activity characterized by alternated, coordinated movements of crossed flexion –extension of the lower limbs while steady-state walking (J.G. Nutt 1993). The French language definition published in 2006 seems to be more comprehensive: “Walking is an intentional motor act, aimed towards a goal, leading to movement of the body in the horizontal plane, via postural and balance constraints”(Beauchet and Berrut 2006).

2.1.2. Gait

Gait can be defined as the manner or style of walking (Levine, Richards et al. 2012). Actually, the word “gait” is often used in a technical context, underscoring the biomechanical aspects linked to the action. More specifically, the human gait could be considered as a complex and cyclical process requiring the synergy of muscles, bones, and the nervous system mainly aimed at supporting the upright position and maintaining balance in static and dynamic conditions (Taborri, Palermo et al. 2016).

However, these systems all suffer age-related physiological effects (for example, sarcopenia). They may also be the target of specific disease processes (for example, diabetes). Some of these systems, organs or functions can also suffer from occupation-related effects (for example, peripheral neuropathy), iatrogenic side-effects (for example, the use of steroids in COPD leads to increased loss of muscle mass), or toxic life habits (for example, peripheral

neuropathy and neural toxicity in alcoholics). Thus, gait is the result of the simultaneous functioning of several complex organic systems, where any process affecting one of these systems may lead to alterations in the harmony and efficiency of gait pattern.

Moreover, two caveats have to be considered when discussing gait performance and gait profiles, namely “normal gait” and “compensatory gait”. Firstly, to consider a gait profile as “normal” or not, it is crucial to take into consideration the age, sex, walking condition and environment. Secondly, many gait abnormalities are a compensation for some problem experienced by the patient and, although abnormal, are nonetheless necessary to maintain mobility. In these cases, the gait pattern is different to a “normal” pattern, but is often “as efficient as possible”, thereby preventing negative clinical outcomes. A good example of this is the older adult, who contracted poliovirus during childhood, and who maintains unilateral lower trunk stiffness and or palsy; these older adults have a non-normal gait pattern. However some of them will never fall.

2.1.3. Balance

The "static balance" involved in standing or sitting is based on the need to keep the body center of mass (COM) into the base of support (BOS) as the BOS remains stationary and the COM moves (Woollacott and Tang 1997). The dynamic balance mechanism is involved in maintaining balance when the individual is moving (e.g. when walking). Indeed, bipedal human walking has an inherent unsteadiness arising from biomechanical disadvantages; 2/3-weighted upper body, a small base of support, and long single-support periods. When walking, both the BOS and COM are moving, and the COM is never kept within the BOS during the single-limb support periods (Woollacott and Tang 1997). Physiological age-related factors are linked to substantial changes in dynamic balance mechanisms. First, among older adults, one of the main goals of dynamic balance is to minimize the head and pelvis accelerations, in order to keep visual and vestibular inferences as stable as possible and ensure a more stable gait pattern (Winter, Patla et al. 1990, Menz, Lord et al. 2003, Menz, Lord et al. 2003, Kavanagh, Barrett et al. 2004, Kavanagh, Barrett et al. 2005). Moreover, dynamic postural control is based on visual, vestibular and somatosensory information which becomes impaired with age, affecting appropriate motor commands to perform balance corrections (Diener and Dichgans 1988). Furthermore, age-related dynamic postural control changes include less anticipatory control mechanisms, more reaction control mechanisms and longer reaction times (Remaud, Thuong-Cong et al. 2015). Finally, attentional capacities seem to be linked to dynamic balance performance. Indeed, according the systematic review of Woollacott et al., "studies using dual task paradigms to examine the effect of age-related changes in attentional requirements of balance control and age-related reductions in stability when performing a secondary task, suggest that these are important contributions to instability in both healthy and balance impaired older adults. For both healthy and balance-impaired older adults, attentional demands

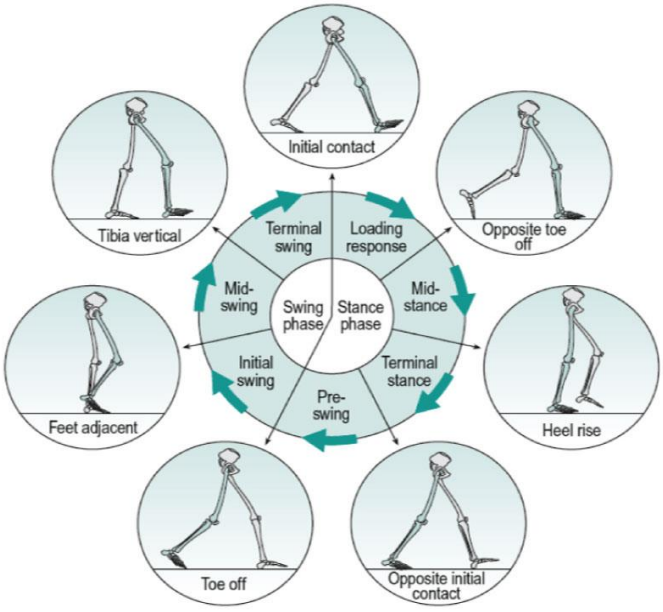
associated with postural control vary, depending on the complexity of the task and the type of second task being performed" (Woollacott and Tang 1997).

In this context, older-age-related dynamic balance performance changes, with some older people adopting a "cautious gait" pattern, reducing gait speed, step length and swing phase (increasing double support), more flat foot landing and more stepping reaction in order to maintain dynamic balance during walking (Winter, Patla et al. 1990, Woollacott and Tang 1997, Jensen, Brown et al. 2001, Menz, Lord et al. 2003, Brodie, Menz et al. 2014).

2.1.4. Normal gait pattern

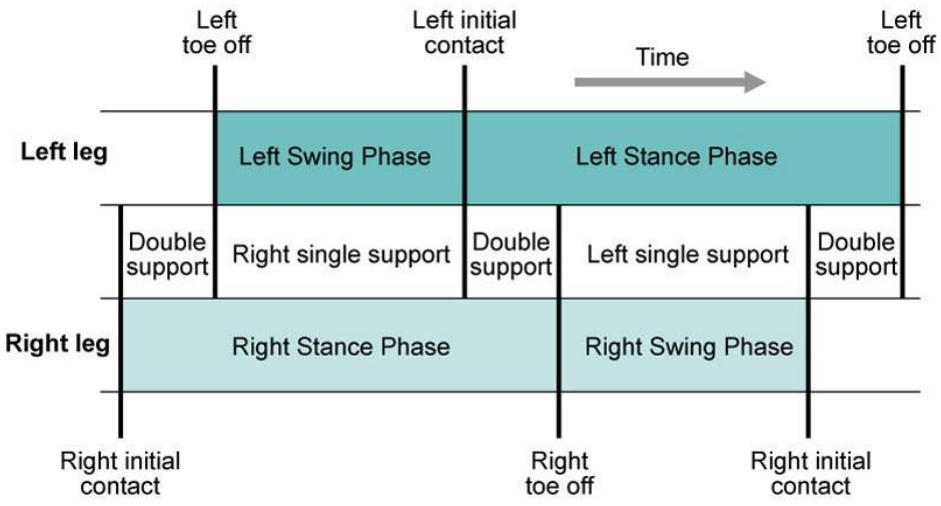
Gait results from complex, coordinated nerve signals sent to the muscles, which in turn move the joints and limbs, producing a repeated sequential contraction and relaxation of muscle groups, resulting in walking. This automatic stepping mechanism is organized by the "central pattern generators" (CPG), which consist of networks of neurons located in different places in the brain and spinal cord, and which receive feedback from sensors in the muscles, joints and skin of the leg (Gabell and Nayak 1984, Levine, Richards et al. 2012). The gait results from stereotyped limb motions repeated in a cycle (gait cycle), where the occurrence of a step is the consequence of the horizontal movement of the COM beyond the limits of the BOS (Jensen, Brown et al. 2001). In addition to this rhythmic pattern, the postural control system, through many complex substrates, regulates the body segment (pelvis, head, arms) position to keep the COM within the BOS (Maki and McIlroy 1996).

The gait cycle is defined as the period of time between the initial contact of one foot, and the next occurrence of the same event with the same foot. The gait cycle can be represented as follows (Levine, Richards et al. 2012):



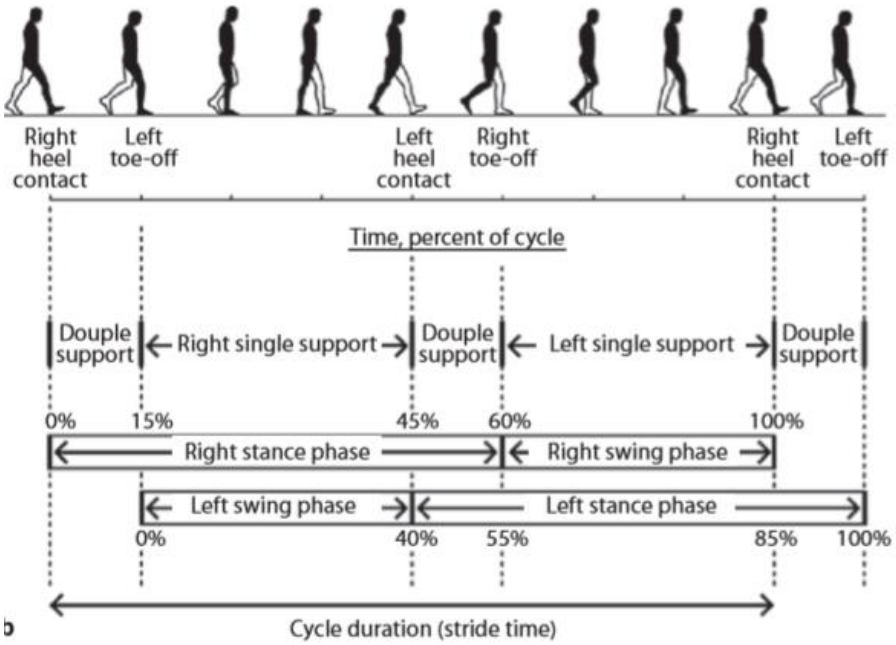
In terms of leg position, the gait cycle includes several events. The first event is the right heel contact; this step starts the right stance phase. The second event is the left (opposite) toe off; this second event starts the swing phase of the left leg. The third event is the right heel rise. The fourth event is the left heel strike and this event stops the left swing phase. The fifth event is the right toe off and this event starts the right swing phase and stops the left stance phase. The sixth event is the left heel rise corresponding to the swing of the right leg. Finally, the last event is the last right swing phase through until right heel contact.

In terms of timing, the figure below shows the different cyclic periods following each other and constituting the gait cycle (Levine, Richards et al. 2012).

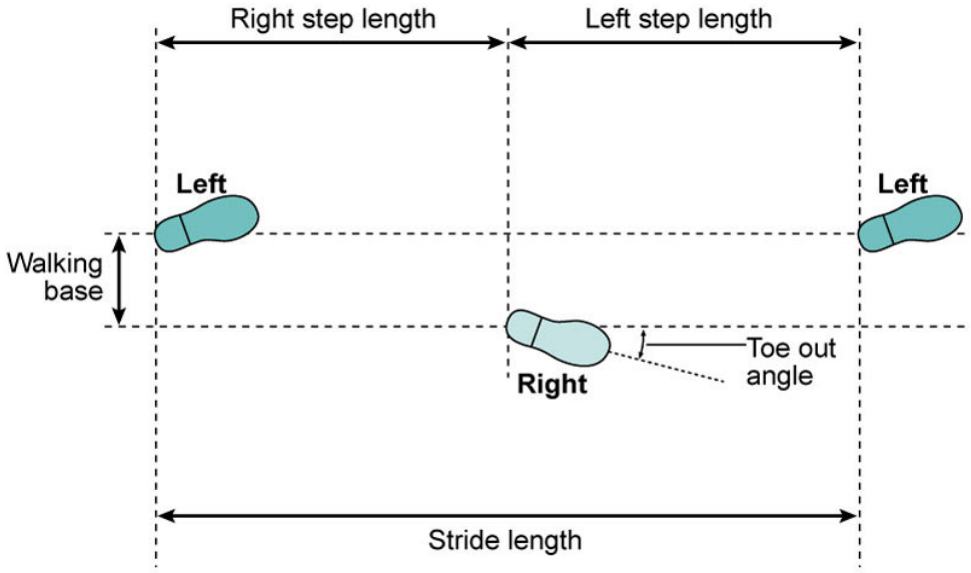


Considering the first event of the cycle, the right initial contact occurs when the left foot is still on the ground; this defines the “double support time”, from the initial contact of the right foot to the toe off of the left foot. The next time is the “swing phase”; during the swing phase of the left foot, only the right foot is on the ground as “right single support”, which ends with the initial contact by the left foot. After this, there is another double support time until toe off on the right side. The left single support corresponds to the right swing phase and the cycle ends

with the next initial contact on the right. Each gait cycle includes two periods of double support and two periods of single support. For each side, the stance phase takes up around 60% of the cycle, while the swing takes up 40%, and each double support period takes up almost 10% of the gait cycle. Actually, as shown in the figure below, the glossary of the temporal gait parameters arises from this gait cycle partitioning.



In terms of feet placement on the ground, the gait cycle includes: the stride length; the right and left step lengths, the toe out angle and the walking base. These terms are also called spatial gait parameters and are represented in the figure below (Levine, Richards et al. 2012).



By definition, the stride length is the distance between two successive placements of the same foot. It consists of two step lengths, left and right, each of which is the distance by which the named foot moves forward in front of the other (Levine, Richards et al. 2012). The right step length is the distance between the left heel and the right heel measured during the double support phase when the right foot is before the left foot. The left step length is the distance between the right heel and the left heel measured during the double support phase when the left foot is before the right foot. As described above, the stride length is the distance between two successive placements of the same foot. In the figure above, the stride length is the distance between the left heel and the next left heel strike. The walking base, also known as the "stride width" or "base support", is the stride to stride distance between the two feet, usually measured at the midpoint of the back of the heel, but sometimes below the centre of the ankle joint, often measured in millimetres (Levine, Richards et al. 2012). The toe out angle is the angle in degrees

between the direction of progression and a reference line on the sole of the foot. The reference line varies from one study to another; it may be defined anatomically but is commonly the midline of the foot, as judged by eye (Levine, Richards et al. 2012).

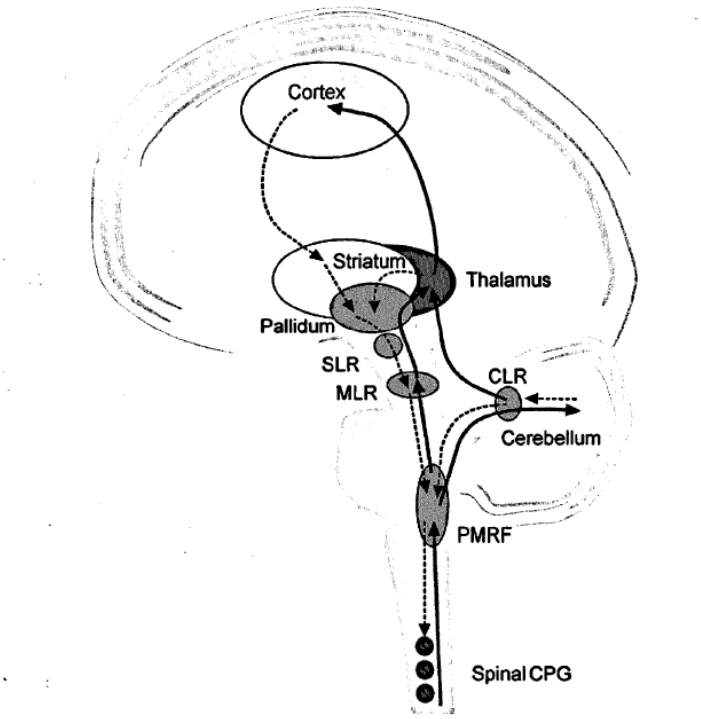
Furthermore, gait parameters consider the gait cycle relative to chronological time. Thus, the gait speed is the distance covered by the subject during a given time. It should be measured in meters per second. The cadence is the number of steps (step frequency) or the number of strides (stride frequency) taken in a given time, the usual units being steps per minute(Levine, Richards et al. 2012). The cadence, in steps per minute, corresponds to half-strides per 60 seconds or full strides per 120 seconds. Lastly, the walk ratio is defined as the step length (expressed in meters (m)) divided by the cadence (step/min) (Sekiya 1998).

2.1.5. Locomotor control of walk

According to Nutt et al.(Nutt 1993) and, from the biomechanical point of view, two abilities are essential to walk with efficient gait: (1) *equilibrium*, the capacity to assume the upright posture and to maintain balance; and (2) *locomotion*, the ability to initiate and maintain rhythmic stepping. The main structures involved in these abilities are the peripheral structures, including peripheral sensory (proprioceptive, vestibular and visual systems) and motor systems, and the central structures, including the spinal cord, brainstem, basal ganglia and cerebellum. The experiment of Grillner, showing that a cat who remained capable of walking on a treadmill and following the different speeds of the treadmill after mid-thoracic spinal transection, supports the existence of spinal interneurons organized as "locomotor generators" (Grillner 1985). In humans, the same experiment was not successful; humans with spinal cord transections may generate complex stereotyped movements but cannot generate rhythmic stepping. However, the theory of the existence of *spinal rhythmic generators* was born.

More than fifty years later, supported by progress in functional brain imagery, Jahn identified the anatomical structures involved in locomotor control. Indeed, using previous works and their own research using the mental imagery of locomotion on functional magnetic resonance imaging (fMRI), Jahn and co-workers found that "locomotion modulates sensory systems and is itself modulated by sensory signals".

As represented in the figure below, they further delineated separate and distinct areas in the brainstem and cerebellum that work together for the purpose of initiating walking initiation and regulating speed.



This schematic organization represents the brainstem and cerebellar locomotor regions (CLR), located close to the fastigial nuclei in the dorsal midbrain; mesencephalic locomotor region (MLR), which would correspond to the cuneiform and pedunculopontine nuclei in the dorsal midbrain; ponto-medullary reticular formation (PMFR); subthalamic locomotor region (SLR) located in the lateral hypothalamic area, the spinal central pattern generators (CPG), basal ganglia, thalamus, cerebellum and cortex. Descending pathways are drawn as dotted lines, ascending pathways in solid lines. Cortical signals project to the brainstem locomotor regions via the striatum and pallidum. The locomotor command conveys a message from the pallidum via the SLR to the MLR, which is further transmitted to the PMRF, where it converges with cerebellar signals from the CLR. The CLR also projects to the MLR via the thalamus and basal

ganglia. It receives input from the vermal and paravermal cerebellar cortex. The PMRF is a major site of interaction between ascending and descending pathways. Cortical signals are modulated via the thalamo-cortical-basal ganglia circuit.

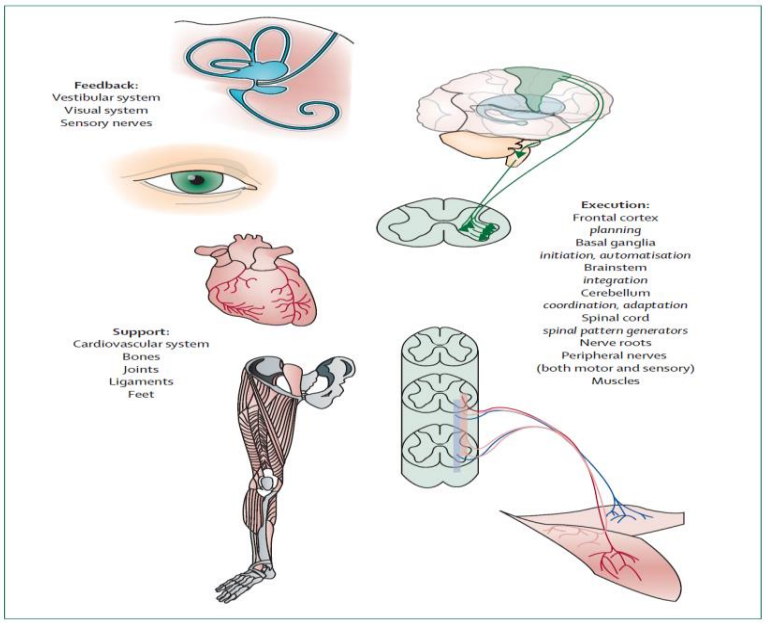
The use of fMRI and [^{18}F]-FDG-PET after real or imagined locomotor tasks has shown that different neuronal structures are involved according to the motor task; namely specific areas of the hippocampus processing spatial navigation (Jahn, Wagner et al. 2009), the role of basal ganglia during walk initiation (Wagner, Stephan et al. 2008), precise walking along a curve (Wagner, Stephan et al. 2008), during turning and during termination of gait (Wang, Wai et al. 2009), the role of bilateral supplementary motor areas during gait initiation (Verghese, Mahoney et al. 2010), the role of bilateral supplementary motor area, the bilateral premotor area (ventral and dorsal), the precuneus, the cuneus, the middle occipital gyrus, the cingulate and bilateral insula when stepping over an obstacle (Verghese, Mahoney et al. 2010). Moreover, in order to acquire deeper knowledge of the involvement of brain structures in locomotion, Jahn used two different acquisition methods, namely brain fMRI in subjects during imagined walking (imagined locomotion), and [^{18}F]-FDG-PET after a real walk (in the same walking conditions as the imagined locomotion), thus obtaining brain imagery of real locomotion. Based on this work, these authors highlighted a common network (common to real and imagined locomotion) including activations in the frontal cortex, cerebellum, pontomesencephalic tegmentum, parahippocampal, fusiform and occipital gyri, and deactivations in the multisensory vestibular cortices (esp. superior temporal gyrus, inferior parietal lobule). As a difference, the primary motor and somatosensory cortices were activated during real locomotion, as opposed to the supplementary motor cortex and basal ganglia during imagined locomotion. This specific activation of the supplementary motor cortex and basal ganglia during imagined locomotion was confirmed by another team (Malouin, L. Richards et al. 2003).

Activations of the brainstem locomotor centres were more prominent in imagined locomotion (la Fougère, Zwergal et al. 2010).

Moreover, the relationship between gait performance and neuronal networks was also highlighted in studies assessing the relationship between gait and cognitive performance. Indeed, Scherder, Verghese, Montero-Odasso and Beauchet have all highlighted the relationship between gait performance and cognition, not only in pathological processes such as dementia, but also in normal ageing. Indeed, in 2007, Scherder et al. published a review of previous data concerning the relationship between gait and cognitive performances (Scherder, Eggermont et al. 2007). In 2011, the same author published a well-documented review highlighting five neuronal circuits: (1) the superior longitudinal fasciculus, connecting the frontal lobe with the parietal, temporal and occipital cortex ; (2) the uncinate fasciculus, connecting the frontal lobe with the lateral and medial temporal lobe ; (3) the fronto-cerebellar connections between the frontal lobe and the medial and lateral cerebellum ; (4) the fronto-striatal connections between the frontal lobe, the caudate/putamen and the temporal lobe ; (5) the cingulum connecting the frontal, parietal, temporal and occipital brain regions; all contributing to the high level of locomotor control, and their implication in gait disturbances occurring in normal ageing and mild dementia (Scherder, Eggermont et al. 2011). Similarly, in 2012, Montero-Odasso et al. published a review underlining the relationship between gait and cognition in ageing and showed evidence that gait assessments can provide insights into cognitive function and fall risk in older people. According to these authors, the dual-task paradigm and the quantification of gait variability could help the clinician to discern people at risk of falls (Montero-Odasso, Muir et al. 2012).

To summarize the outstanding progress in our knowledge of locomotor control, electrical and chemical stimulation in animal and human models has highlighted the role of some brain areas, such as the locomotor region working as a dynamic pacemaker that gives rhythm to the human walk, taking into account somato-sensory inputs, while functional brain imaging studies provide evidence of widespread, complex and dynamic neural network organization and modulation of locomotor control based on visual, somato-sensory and vestibular inputs. Studies focusing on cognitive performance have highlighted a common neural network that is involved in both gait performance and cognitive performance.

Obviously, the central nervous system is associated with the peripheral nervous system conducting the locomotor order and somato-sensorial information. Both systems working together are needed to ensure efficient walking, as represented in the following figure taken from Snijders et al. (Snijders, van de Warrenburg et al. 2007).



2.1.6. Gait, balance and postural reaction ageing

Physiological ageing is associated with declines in several systems including musculoskeletal, cardiovascular, visual, vestibular and proprioception, coordination, slowed postural responses and cognitive function (Ambrose, Paul et al. 2013). This leads the gait pattern to become less coordinated, with poorer postural control (Ambrose, Paul et al. 2013). Moreover, as dynamic balance is involved during walking, and considering that among older adults, one of the main goals of dynamic balance is to minimize head and pelvis accelerations in order to keep visual and vestibular inferences as stable as possible and ensure a more stable gait pattern (Winter, Patla et al. 1990, Menz, Lord et al. 2003, Menz, Lord et al. 2003, Kavanagh, Barrett et al. 2004, Kavanagh, Barrett et al. 2005), age-related changes in walking are numerous, and most will be further detailed.

Before listing the characteristics of gait ageing, a comment has to be notice. Namely, the knowledge concerning gait ageing is improving in the same time the knowledge on the ageing. Actually, the growing research on the ageing and especially on the gait ageing allows reviewing previous knowledge taking into account the heterogeneity of the ageing. Thus the literature focusing on gait aging is not consensual, depending on the participants the studies involved.

Until the end of the previous century, the physiological gait ageing was described as a reduced gait speed, with reduced arms swing, wilder base of support, longer double support time, shorter single support time, shorter steps and reduced ankle and knee flexion/extension movements (Murray MP 1969, Abram, Beer et al. 1995). According to Murray, the shorter swing phase, longer double support, wider walking support base and reduced stride length, reduced vertical movement of the head while the lateral movements are increasing, are all supposed to improve the security of walking (Murray MP 1969).

However, several years later, a publication by Nakamura associated these gait pattern modifications with pathological neurodegenerative processes (such as Alzheimer's disease) rather than to age alone. Moreover, according to the same author, people with vascular dementia show the same gait pattern changes, but to a more pronounced degree (Tanaka, Okuzumi et al. 1995).

More recently, the Baltimore Longitudinal Study of Aging comparing gait patterns according to age ranges showed that older age is associated with slower self-selected walking speed, shorter stride length, a greater tendency to land flat-footed (instead of heel strike landing) and lower hip generative mechanical work expenditure and lower knee absorptive mechanical work expenditure (Ko, Ling et al. 2009, Ko, Stenholm et al. 2011, Ko, Tolea et al. 2011). Indeed, several of these changes could be secondary to age-related changes in the angular excursion of the joints, including a reduction in the total range of hip flexion and extension, a reduction in swing phase knee flexion and a reduced ankle plantar flexion during the push off (Levine, Richards et al. 2012). However, people included in this longitudinal study have not been screened for mild or moderate cognitive disorders then some gait modifications observed could more be linked to neurodegenerative process than to age alone.

In another hand, some authors have shown that the decreased stride length observed among older adults could be linked to weakness in hip extensor and ankle plantar flexors, reduced push-off phase, increased swing phase and a reduced ability to move the body forward during gait (Winter, Patla et al. 1990, Bassey, Fiatarone et al. 1992, Judge J 1996). Furthermore, walking more slowly with a higher stride frequency and shorter stride length also may help to stabilize the gait pattern and allow greater adjustment and flexibility to changes in walking condition (Barak, Wagenaar et al. 2006). However, lastly, Zijlstra et al. showed that there is no evidence for a change in the step length – frequency relationship with age among physically active community-dwelling older women (Zijlstra, de Bruin et al. 2008).

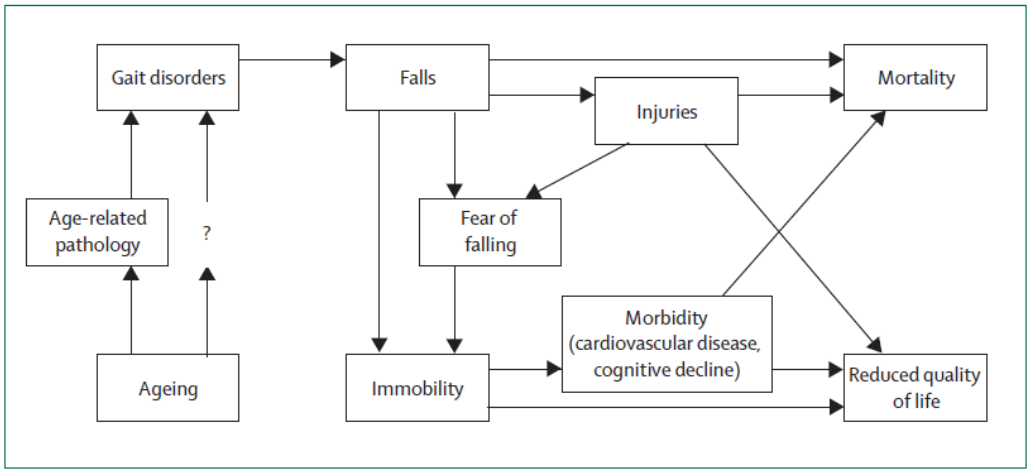
Furthermore, the term “cautious gait”, has been coined to characterise the gait of older adults who display gait pattern alterations due to unknown disease, associated with higher mortality, and which is not to be considered as a normal ageing pattern (Bloem, Gussekloo et al. 2000). Cautious gait is characterised by decreased speed and stride length (Giladi, Herman et al. 2005). As shown by Giladi et al. (Giladi, Herman et al. 2005), this gait pattern is associated with more anxiety, more depressive symptoms, greater fear of falling and lower cognitive performance, which are also to be linked to gait pattern modifications and fall risk. This gait pattern seems to be linked to the vascular burden on the central nervous system (Giladi, Herman et al. 2005).

To summarize, the amplitude and the meaning of this gait pattern (decreased speed and stride length) seem to be different in terms of the robustness and health profile of the people exhibiting it. Indeed, the walk ratio, although remaining unchanged in the very fit, could change as an adaptive and protective behaviour in less fit older adults, while pronounced gait pattern modifications could be linked to the “cautious gait” described by Bloem et al. (Bloem, Gussekloo et al. 2000), signifying reduced locomotor capacities. Thus, while these studies support the idea of a clinical gait continuum between age-related walking patterns, walking pattern changes related to neurodegenerative processes and the walking pattern changes related to cerebral vascular burden, the same studies also support the idea that individual profiles must be kept in mind when discussing gait patterns and their relationship with negative outcomes.

Concerning postural reactions, older adults seem to be more likely to use a stepping reaction to recover balance and step at lower perturbation magnitudes than younger adults (Jensen, Brown et al. 2001). However, studies focusing on stepping reactions when balance is overcome are not unanimous. One study has shown that a stepping reaction is often used in older adults not only when the COM overcomes the BOS, but also when the velocity of the COM change is high (Pai, Rogers et al. 1998). Conversely, other studies have shown that older adults (Wolfson, Whipple et al. 1986, McIlroy and Maki 1996) are less prone to weight shifting or taking a rapid step when balance abilities were exceeded than younger adults. In our opinion, previous falls among older adults included in these studies could explain the differences in the results obtained.

Additionally, older adults appear to be more reliant on arm reactions than younger adults but are less able to execute reach-to-grasp reactions rapidly (Maki and McIlroy 2006).

Finally and regardless of the individual gait ageing process and the reason(s) why older adults fall, gait performance, fall risk and fall(s) are intricately inter-related as shown in the following figure (Snijders, van de Warrenburg et al. 2007) showing the indirect relations between gait ageing and geriatric gait disorders.



2.1.7. Gait Variability

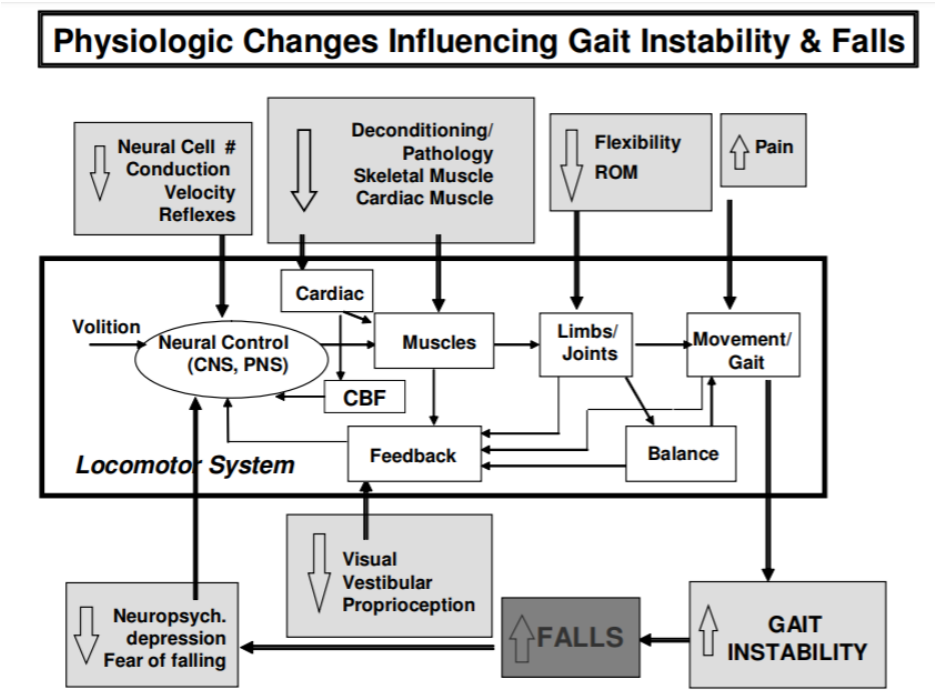
Gait variability is the fluctuation in gait characteristics from one step to the next (Brach, Perera et al. 2010) and reflects gait patterning process abilities and balance control mechanisms.

According to Gabell and Nayak, the gait parameters related to balance control mechanisms seem to be the stride width and the double support time, while the step length and stride time reflect the functioning of the gait patterning mechanisms (Gabell and Nayak 1984). According to same authors, an increase in the variability of the stride width or double support time could reflect a lack of compensation for stability occurring in case of circumstances that stress the balance mechanisms, while an increase in step length or stride time reflects a problem in the stepping mechanisms. According to these assumptions, these authors assessed gait variability in a large sample of 1187 people aged at least 64 years old, and showed that younger and older adults have similar gait variability. The authors concluded that increased gait variability occurring in old people is not normal but is due to some pathological cause (Gabell and Nayak 1984).

Some years later, several teams published papers confirming these findings, and showing that an increase in gait variability has to be considered as an important indicator of impaired mobility in older adults associated with falls and/or disability (Maki 1997, Hausdorff, Rios et al. 2001, Brach, Studenski et al. 2007, Brach, Studenski et al. 2008, Storti, Pettee et al. 2008). In fact, greater variability in stride or swing time seems to be predictive of future falls (Maki 1997, Hausdorff, Rios et al. 2001), and a greater stance time variability is an independent predictor of future mobility disability (Brach, Studenski et al. 2007). Gait variability has been related to less confidence in walking, and lower levels of daily physical activity (Brach, Studenski et al. 2008). The relationship between gait impairment and disability is thus clearly established (Hausdorff 2005).

Furthermore, gait variability has also been considered to reflect cognitive functioning. In healthy older adults, stride to stride variability is regulated by automated processes and requires minimal cognitive resources (Hausdorff 2005) even during challenging cognitive tasks in healthy older adults without fall history, while the gait variability increases in dual task walking in older fallers, especially in case of executive function decline (Springer, Giladi et al. 2006). In another cohort, Brach et al. (Brach, Perera et al. 2010) confirmed that gait variability is related more to executive functioning than to age. Finally, greater gait variability was observed in impaired executive function syndrome, as in demented adults with frontal lobe dysfunction (Allali, Kressig et al. 2007), Parkinson's disease (Hausdorff, Cudkowicz et al. 1998) or Alzheimer's disease (Tanaka, Okuzumi et al. 1995).

The figure below shows a sample of the alterations that occur in ageing and disease and which affect gait stability, at least as reflected in terms of stride time variability, and fall risk (where Flexibility ROM means flexibility range of motion, CNS means central nervous system, PNS means peripheral nervous system, CBS means cerebral blood flow) (Hausdorff, Nelson et al. 2001, Hausdorff 2005).



The variability of a gait parameter can be measured as the within-subject standard deviation of the gait parameter derived from all of the steps (Brach, Perera et al. 2010) or as the coefficient of variation of the gait parameter= [standard deviation of gait parameter/ mean of gait parameter) X 100](Hausdorff, Rios et al. 2001). Depending on the topic and the instrumental method used, gait parameters assessed for their variability include the step length, step width, stride time, stance time, swing time and double support time. At this time, the number of recorded strides needed to measure gait variability, the walking condition and/or the gait speed required or the prioritisation task to be performed remains non-consensual.

2.2. Assessing gait

Considering the meaningfulness of gait disorders in terms of risk, and considering gait as a reflection of the functioning of several organic systems, gait assessment should be systematically targeted by anamnesis and clinical evaluation during comprehensive geriatric assessment.

The following section will provide an overview of the various ways to assess gait in the clinical setting following recently published guidelines by groups of experts (Beauchet, Allali et al. 2017) and according the usual daily practice implemented in our geriatric care department.

In fact, depending on the setting, the subjects addressed for assessment and the purpose of gait pattern assessment, the gait assessment may be rapid, or more in-depth.

However, irrespective of the context, gait assessment must, at the very least, include anamnesis, a comprehensive clinical exam, and a visual evaluation of standing, transferring, walking and turning.

Moreover, depending on the context, a range of questionnaires, functional tools and gait analysis methods can be added, and evaluation in different walking conditions may be useful.

2.2.1. Anamnesis

After having considered socio-demographic data, medico-surgical events (including sensorial deficits and geriatric syndromes), usual medications and toxic abuse, the anamnesis must also consider walking complaints such as pain, weakness or unsteadiness, the need for walking aids, the usual shoes, the usual daily living environment, social and physical activities, and the potential history of falls, fear of falling, aspects related to well-being, mood and cognitive complaints, self-perceived quality of life and social support.

2.2.2. Clinical exam

The clinical exam should include investigation of osteo-articular and muscular system abnormalities, including nervous system assessment, i.e. sensorial abilities, static and dynamic balance. Moreover, a cardio-respiratory examination is recommended.

2.2.3. Visual evaluation of walk

The visual examination should consider the position on chair, standing up and potential backward disequilibrium, steady walk in different walking conditions, initiation and ending of the walk, turning, ability to deal with obstacles and psycho-motor reactions when faced with different level ground surfaces. After training, the clinician can qualitatively assess static and dynamic balance performance visually, as well as the quality of the gait (more or less fast, with short or long step length, with more or less gait variability, with more or less confidence).

However, visual evaluation has some limitations. The clinician must be well trained to systematically analyse aspects all of the gait pattern and discern among individual gait characteristics those that have to be considered as abnormalities and those that are potential compensation mechanisms. Moreover, whatever the level of expertise of the clinician, visual

evaluation remains subjective. Furthermore, this qualitative assessment does not yield any numerical score that can be recorded and compared between individuals.

2.2.4. Walking conditions

To the best of our knowledge, the term “walking condition” is not clearly defined in the scientific literature. A comprehensive definition could be “the circumstances in which the subject is walking”. These circumstances could be linked to conditions external to the subject (for example, the ground, or walking on a treadmill, walking on smooth or irregular surfaces) or to conditions internal to the subject, where the volunteers are asked to walk following specific instructions, such as a comfortable walking condition, meaning the subject is asked to walk at their spontaneous gait speed; fast walking conditions; fastest walking conditions; slow walking condition; dual task walking condition; slow walking condition. Since comfortable, fast and dual task walking conditions will be used in this research work, these walking conditions will be further developed here without consideration for other potential walking conditions.

“Spontaneous”, “preferred” or “comfortable” walking speed is always used in gait tests because this is the obvious way to assess the gait pattern, and secondly, because the spontaneous gait pattern reflects ability in terms of central locomotor function. Indeed, at the usual, self-selected pace, gait variability is lowest, leading to a more efficient gait pattern (McNeill Alexander 2002). In fact, humans seem to adjust their walking to minimise the metabolic energy cost of locomotion. The walking speed that we tend to prefer is the one that minimises energy cost per unit distance (McNeill Alexander 2002). Moreover, the comfortable self-selected gait speed is associated with the “more stable” gait pattern where variability in the accelerations in the tree axis was the lowest (Menz, Lord et al. 2003) and the stride to stride temporal variations are minimized (Hausdorff, Edelberg et al. 1997).

Among older adults, the ability to walk faster seems to be related to muscle strength, and especially to knee extension (Bohannon, Andrews et al. 1996, Bohannon 1997). As faster walking seems to be more challenging than a comfortable walking speed, this walking condition is also used during gait pattern assessments. Indeed, fast walking makes it possible to show

alterations in gait patterns in older adults with type 2 diabetes, even without peripheral neuropathy (Ko, Stenholm et al. 2011) and can discern fallers and non-fallers among older adults (Barak, Wagenaar et al. 2006), discern people at risk of cognitive decline (Fitzpatrick, Buchanan et al. 2007, Deshpande, Metter et al. 2009) or those at risk of disability (Artaud, Singh-Manoux et al. 2015). Moreover, the fast walking condition has been suggested to be useful for identifying, among older adults, those at risk of future fall(s) (Callisaya, Blizzard et al. 2012).

As previously explained, walking was initially considered as an automatic behaviour. However, in daily-life walking conditions, the ability to avoid an obstacle or deal with directional changes are evidence that intentional mechanisms are also involved, requiring explicit cognitive function. In order to confirm this intuitive feeling, some authors have used the dual task paradigm to demonstrate the involvement of cognitive functions during walking.

The dual-paradigm is based on the hypothesis that two tasks interfere when they have to be performed simultaneously and rely on the same functional and/or cerebral system (Woollacott and Shumway-Cook 2002). In this type of gait assessment, one of the tasks is walking, while the other task could be an "attention-demanding" task, such as an auditory Stroop task, a fluency task, or a task involving working memory. The gait modification observed during the dual-task walking condition is interpreted as the involvement of attention while walking, and demonstrates the involvement of cortical control (Woollacott and Tang 1997).

Furthermore, older adults seem to require more attention for motor control while walking than younger persons (Woollacott and Shumway-Cook 2002, Laessoe, Hoeck et al. 2008). Indeed, gait performances in the dual-task walking condition were more affected in healthy older adults than in young people (Al-Yahya, Dawes et al. 2011). In this regard, Bloem et al. showed that even among healthy subjects, older adults are less inclined to give priority to

ensuring balance ("posture first strategy") in the dual-task condition than younger people (Bloem, Valkenburg et al. 2001), meaning that some of them give priority to the cognitive task rather than to the walking.

Actually, the dual-task walking condition has also been used in clinical studies to study the involvement of attention mechanisms (Siu, Chou et al. 2009, de Bruin and Schmidt 2010) and executive function in gait performance; in healthy elderly subjects (Hausdorff, Yogev et al. 2005) and in people with dementia (Sheridan, Solomont et al. 2003, Allali, Kressig et al. 2007, Festa, Heindel et al. 2010). Thus, the dual-task paradigm can be used as a cognitive stress resistance model, where the threshold for the occurrence of dual-task-related gait interference is dependent on the cognitive load (Bridenbaugh and Kressig 2011). Given that the use of the dual-task paradigm leads to changes in gait performance that may go undetected in the single-task walking condition; the dual-task walking condition has also been used to detect older adults at risk of falls in their daily life activities. Indeed, a landmark study published by Lundin-Olsson et al. (L. Lundin-Olsson 1997) showed that people who stop walking when talking have an increased fall risk compared to those who keep walking. Some years later, several teams around the world confirmed that modifications in gait profile during dual-task walking are associated with an increased risk of falls (Verghese, Holtzer et al. 2009, Bridenbaugh and Kressig 2011, Gschwind, Wolf et al. 2011, Montero-Odasso, Verghese et al. 2012).

In the same field of interest, a review of the literature explored dual-task walking modifications in healthy young and older adults, compared to dual-task modifications in frail subjects or elderly fallers. While healthy elderly subjects essentially showed decreased gait speed and reaction time, frail subjects and fallers showed reduced gait speed, shorter steps, longer double support time, and increased gait variability (Hausdorff, Schweiger et al. 2008).

However, the scientific literature is not consensual regarding the utility of using the dual-task paradigm to better discern people at risk of fall(s). Indeed, two studies involving non-

demented people failed to demonstrate the utility of gait performance obtained in the dual-task condition in identifying people at risk of future falls (Bootsma-van der Wiel, Gussekloo et al. 2003, Beauchet, Allali et al. 2008). The prospective study by Herman et al., including healthy non-demented older adults, showed that among gait parameters obtained in the dual-task walking condition, only gait variability could be useful to discern future fallers (Herman, Mirelman et al. 2010). Finally, three literature reviews focusing on the usefulness of gait parameters obtained during the dual-task walking condition to detect people at risk of fall(s) among healthy older adults have concluded that further prospective studies including larger samples and using standardized data acquisition, data processing and statistical analysis are needed to show whether the dual-task walking condition could be useful to discern people at risk of a future first fall (Zijlstra, Ufkes et al. 2008, Beauchet, Annweiler et al. 2009, Muir-Hunter and Wittwer 2016). In one of these literature reviews, O. Beauchet et al. confirmed that "no association between dual-task-related gait changes and fall incidence was found in the three studies that focused on community-dwelling older adults. In contrast, the highest predictive values for falls based on dual-task-related gait changes were found in institutionalized participants (i.e. frail older adults) and geriatric inpatients" (Beauchet, Annweiler et al. 2009).

2.2.5. Functional tools

Several tools have been developed to assess static and dynamic balance and gait performance. The main advantage of these tools is that they yield a score to record, thus allowing comparison between individuals, but also across further gait assessments in the same person. Most of the functional tools require specific training, but are easy to implement in clinical practice.

The available functional tools are numerous, and to draw up an exhaustive list of them is not the goal of this paragraph. Suffice to say that the most important aspects to consider when choosing a functional tool are its validation among the population targeted by the gait assessment (e.g. among community-dwelling older adults, among institutionalized subjects, among fit or frail people, etc.), its validation for the detection of the negative outcome under investigation, and finally, the requirements in terms of training and equipment.

However, even functional tools that give global or sub-scores assessing balance and gait performances have failed to assess gait cycle time and gait parameters, which require instrumental methods to be adequately addressed.

2.2.6. Instrumental methods

Considering the limits inherent to visual assessment and functional tools, and in order to improve accuracy of assessment, instrumental methods have been applied to walking to enable the measurement of specific gait parameters. Compared to visual assessment and functional tools, instrumental methods offer several advantages, i.e.: (1) accurate and reliable measurements, (2) knowledge of the individual and detailed gait performance, i.e. “gait profile”, (3) the opportunity in the field of research to link “gait profiles” with physiological and/or pathological conditions, thereby improving the current state of knowledge regarding the mechanisms underlying the gait process, and finally, (4) the opportunity to link “gait profiles” to future negative clinical outcomes, such as fall risk, for example. In this context, methods to analyse gait have been widely developed over the last twenty years and numerous methods are now available. We will detail available approaches in the next chapter.

Methods to assess gait can be classified according to the variables they measure: temporal and spatial gait parameters, muscle activities or energy expenditure. Regarding the topic of this research, this chapter will only overview the instrumental methods that assess temporal and spatial gait parameters.

Instrumental methods to assess gait cycles and gait parameters can be considered as “kinetic” or “kinematic” methods, depending on their variable of interest. Kinetic methods measure the variables that create the motion (i.e. the force(s)), while kinematic methods measure the motion itself (i.e. linear or angular position(s)). Kinematics is the measurement of movement in terms of displacement, velocity and acceleration. Kinematic systems are used in gait analysis to record the position and orientation of the body segments, the angle of the joints and the corresponding linear and angular velocities and acceleration (W.Whittle 2007).

Another classification considers the properties of the device, rather than the variables measured, and divide the instrumental methods into solution-based wearable sensors (including

footswitches, foot pressure insoles, accelerometers, gyroscopes and inertial measurement units combining accelerometers and gyroscopes or a combination of footswitches or foot pressure insoles and inertial measurement units) and non-wearable sensors (including opto-electronic systems, and force platforms).

Regardless of the computational methodologies adopted, different sensor systems can be used to capture gait phases/parameters. Nowadays, wearable sensors are largely used to decompose the phases of gait; foot pressure insoles or footswitches are the gold standard to extract these phases, since each transition between gait phases can be associated with a specific value of the sensor output. Alternatively, accelerometers and gyroscopes are widely used to inform algorithms for gait phase extraction. In a controlled laboratory environment, non-wearable sensors, such opto-electronic systems or force platforms, are widely used (Taborri, Palermo et al. 2016). Once again, in view of the focus of this research, this chapter will present a general overview of available methods assessing gait parameters, focusing particularly on the methods involved in the experimental section, namely acceleration-based methods and opto-electronic methods.

A. Force platforms

Force platforms (also known as force plates), are used to measure the ground reaction forces of a subject when they hit the force platform during walking. Within the platform, a number of transducers are used to measure tiny deformations of the upper surface, in all three axes, when forces are applied to it. Ideally, a force platform should be mounted below floor level, with the upper surface being flush with the floor. Acquisition of data using force platforms can be subject to several problems. One problem is linked to the "aiming" effect. Indeed, to obtain good data, the whole of the subject's foot must land on the platform. It is tempting to tell the subjects where the platform is and to ask them to make sure that their footstep lands squarely on it. However, this is likely to lead to an artificial gait pattern.

Furthermore, when a recording from both feet is required, the relative positioning of two force platforms can be a considerable impediment, considering the range of step length in the general participants across ages, gender, and pathophysiological processes. There is no single related position that is satisfactory for all subjects. Force platforms are essentially used to assess balance, rather than walking. To address this problem, some researchers have developed devices consisting of a small number of pressure sensors, which are fixed to the sole of a shoe. First prototypes seem to be easy to use and efficient, and research in this area is growing with a view to enabling gait analyses based on walking periods during daily living (Levine, Richards et al. 2012).

In fact, the human movement laboratory performing this study regularly uses force platforms. However, considering the previous cited limitations and in the absence of any available and recently developed prototype, force platforms will not be involved in the experimental section of this research.

B. Footswitches

Footswitches included sensors able to directly detect the contact between the foot and the ground during gait cycles. They are considered the gold standard in the field of gait phase detection (Taborri, Palermo et al. 2016) and are often used with shoes to record the timing of the cycle such as the timing of initial contact, foot flat, heel rise and toe-off, and the duration of the stance phase, the swing phase, and single and double support, according to the measurements using one, two or four footswitches. However, they have several disadvantages. In fact, focusing on their use among older people, some practical limitations include: (1) the placement of the sensors on patients with pathological gait affects the accuracy and reliability of the measure; (2) reliability is susceptible to mechanical failures (Willemsen, Bloemhof et al. 1990, Taborri, Palermo et al. 2016). Considering the usual gait pattern heterogeneity present among older adults and in order to avoid encountering technical constraints or limitations, footswitches will not be involved in this study.

C. Walkways

Instrumental walkways are used to measure the timing of the foot contact, the position of the foot on the ground, or both. In their most advanced versions, walkways contain a large number of switch contacts that detect the foot position as well as the timing of the heel contact and toe off. Timing information from the foot contacts is used to calculate the cycle time and the combination of cycle time and speed may be used to calculate the stride length (GaitRite/CIR systems). This system is commonly used in research and in the clinical setting; an example of one such system is the "GAITRite" (Levine, Richards et al. 2012). The advantage of this system is that walkways are easy to use, and validated for use among older adults (Bridenbaugh and Kressig 2011). Disadvantages include the fact that their use in the clinical context requires the availability of a dedicated corridor to install the walkway, and the length of the walkway (generally less than 10 meters) limits the walk distance available for assessment. Finally, this method does not assess the sub-phases of the swing. Nevertheless, considering that it is easy to use and yields accurate and reliable gait parameters, this instrument could be used for gait assessment among older adults and previous literature already exists (Bridenbaugh and Kressig 2011). Unfortunately, at the time of the design of this research project, we did not have any available walkway, precluding its use here.

D. Camera-based motion analysis

Pioneering researchers in this field used photography with a succession of pictures at a high frame rate to decompose the walk into several steps. Nowadays, advanced kinematic systems measure the position of markers, fixed on specific landmarks, in three dimensions. In passive marker systems, the markers reflect (infra-red) light emitted by the cameras. Confusion between the markers is possible in certain conditions. In active marker systems, the position marker emits its own signal (usually light-emitting diodes (LEDs)). Active markers have their own emission timing, which enables them to be automatically distinguished. These systems allow accurate 3D-position analysis during movement or walk. Actually, these systems could be used to perform gait analysis. Kinematic data recorded by an opto-electronic system represent the gold standard for quantified gait analysis in the laboratory environment, but cannot be used outside the laboratory in real-life situations (Taborri, Palermo et al. 2016). Different algorithms can be chosen to extract the gait events from the position of the markers, such as the timing of the heel strike (HS) and toe off (TO), the events that mark the transitions between stance and the swing phase of gait (O'Connor, Thorpe et al. 2007). The literature reports three main sources of error with this type of system, namely: soft tissue artefacts (STA) (Cappello, Stagni et al. 2005), errors of marker placement (Della Croce, Leardini et al. 2005, Sint Jan and Croce 2005) and instrumental errors (Chiari, Croce et al. 2005). Soft tissue artefacts are linked to the relative motion of the soft tissues on which the markers are fixed, and the bone structures below the soft tissue. Errors of marker placement are usually reduced through conscientious compliance with the instructions and protocols. Instrumental errors are usually no longer a major issue, since recent systems offer sub-millimetre accuracy after appropriate calibration (Windolf, Götzen et al. 2008). The accuracy depends on the device, on the distance between the markers and the cameras, and of course, on the accuracy of the algorithm detecting gait events based on the position signal. This method, as the gold standard in motion analysis, is

accessible in the Laboratory of Human Motion Analyses installed in the University of Liège, and therefore, we chose to use this method to assess gait parameters among older adults.

E. Accelerometer-based methods

The use of accelerometric methods to assess human movement was first suggested in the 1970s, but has been refined and perfected thanks to the research of two Norwegian physicians, R. Moe-Nilssen and J.L. Helbostad (Moe-Nilssen 1998, Moe-Nilssen 1998, Moe-Nilssen 1998, Moe-Nilssen and Helbostad 2004). Briefly, the main objective underpinning their research was to devise an instrument to measure linear acceleration of the trunk during gait under real-life environmental conditions. To achieve this goal, they used a triaxial piezo resistant accelerometer connected by a co-axial cable to a battery-operated portable data logger. Analog signals were low pass filtered at 200Hz before being sampled at 512 Hz to avoid aliasing. The piezo accelerometer registered gravity as a static vertical component in addition to the dynamic acceleration caused by change of velocity during locomotion. Using the accelerometer as an inclinometer, R. Moe-Nilssen showed an algorithm able to compute estimates of the gravitational components in the antero-posterior and the medio-lateral directions. Knowing the gravitational components of the antero-posterior and the medio-lateral acceleration vectors and the average tilt of the accelerometer, this system was able to transform acceleration data to a horizontal-vertical orthogonal coordinate system. The accelerometer was tested for precision and accuracy in dynamic conditions showing results close to the precision limits of the gold standard instrument used (a servo hydraulic jig).

Attaching a triaxial accelerometer over the L3 spinous process is suggested to represent a close reflection of actual lower trunk accelerations during walking (Moe-Nilssen 1998). This is due to the fact that this region of the body has low transverse plane rotation relative to axial rotation in the sagittal and frontal planes. A tilt compensation procedure often includes post-data-collection signal manipulation (specific to the device used). In this context, acceleration-based instruments were developed and applied to human gait, firstly to demonstrate the

reliability of the measurements at different self-selected speeds, confirming that this method could be used in walking tests planned in real-life conditions.

Based on this development process, R. Moe-Nilssen and J.L. Helbostad applied this method to human gait and validated the processing method to obtain an estimation of cadence, step length and step and stride regularity and symmetry (Moe-Nilssen and Helbostad 2004). The use of acceleration-based methods to assess gait among older adults was also validated by several others authors (Zijlstra and Hof 2003, Zijlstra 2004, Hartmann, Luzi et al. 2009, Hartmann, Murer et al. 2009, Zijlstra and Zijlstra 2013).

Compared to other gait analysis instruments, accelerometers are low-cost, small and non-invasive (Kavanagh and Menz 2008). The availability of direct measurement of 3D accelerations eliminates errors associated with differentiating displacement and velocity. A major advantage of accelerometers is that measurements are not restricted to a controlled laboratory environment, and can handle gait analysis in an entirely natural setting, with the possibility to obtain gait parameters over longer walking distances. Thus, accelerometers tend to be the most widely used solution for ambulatory gait analysis (Taborri, Palermo et al. 2016).

Depending on the focus of investigation, accelerometers can be placed on different parts of the body such as the head, the trunk, the lumbar position or the feet. In addition, this method makes it possible to measure the sub-phases of the swing phase (Taborri, Palermo et al. 2016). In fact, in view of these numerous advantages, the use of accelerometric methods is growing in research. Moreover, recently, the use of automated gait event detection software (i.e. toe off and heel strike) enables more detailed analysis of the granularity of gait cycles.

Considering previous expertise and these numerous advantages using this method, we chose to implement this instrumental method in this research work.

F. Gyroscopes

Gyroscopes measure angular velocity, which is not influenced by the gravity and by the vibration occurring during the heel strike (Taborri, Palermo et al. 2016). The data obtained using this method are complementary to those obtained using acceleration-based methods (Bonnet, Ramdani et al. 2014). As with accelerometers, and depending on the focus of the investigation, the data obtained with a gyroscope depend on its placement. At present, the use of this method is growing, especially in research focusing on gait and balance disorders (Brodie, Psarakis et al. 2016). However, their use remains restricted and further studies are needed to demonstrate opportunities for its use among older people. Considering that this method was not available for use it at the beginning of this work, we did not use gyroscopes in our assessments.

In summary, the potential gait parameters available after gait analysis depend on the instrumental method and material used. Most gait parameters can be obtained using different instrumental methods. Moreover, gait parameters can be measured or calculated. The table below summarizes the instrumental methods available, and shows a non-exhaustive list of the gait parameters that can be obtained using the instrumental methods discussed, and whether the gait parameters were recorded or calculated. The methods used and the parameters considered in the experimental section of this work are highlighted in bold font.

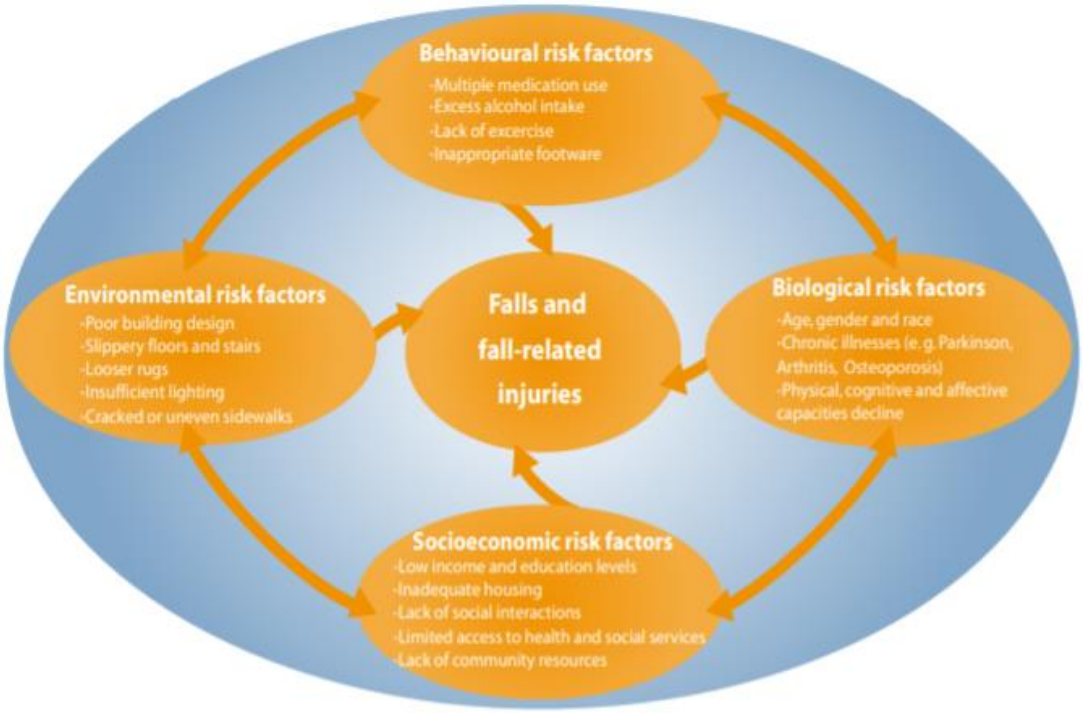
Methods	Gait cycle/ parameters	Measured	Calculated
Force platforms	- Ground reaction forces in vertical axis during stance.	X	
Footswitches	- Timing of initial contact, foot flat, heel rise, toe-off.	X	
	- In case of at least one stride available: duration of the stance phase.		X
	- In case of two or more available: duration of stride and swing phases.		X
	- In case footswitches are on both feet: single and double support time.		X
	- Stride to stride variability measures.		X
Walkways	- Timing of initial contact, foot flat, heel rise, toe-off.	X	
	- Durations of the stance phase, the swing phase and the single and double support phase.		X
	- Gait speed, step and stride length, cadence.		X
	- Stride to stride gait variability measures.		X
Camera-based motion analysis or opto-electronic method	- Three dimensional positions of active markers in time.	X	
	- In case of markers on feet: step or stride length.		X
	- Cadence, duration of the stance phase, the swing phase and the single and double support phase and stride to stride variability measures.		X
	- **Minimal toe clearance.**		X
Accelerometer-based method	- Accelerations measured on three axes.	X	
	- Gait event timings: heel strike, toe strike, heel-off, toe-off, maximum heel clearance, and maximum toe clearance.	X	
	- Depending on the place on the body where the accelerometer is placed: step or **stride length**, step or **stride cadence.**		X
	- Step or **stride regularity and symmetry**, harmonic odds ratio, root mean square, step time variability, step time asymmetry, step stability index, stride to stride variability.		X
Gyroscopes	- Angular velocities in three axes.	X	
	- Mean, root means square and coefficient of variation of angular velocities in three axes.		X

2.3.1. Fall definitions

Several definitions of a fall are available in the literature. One of the most comprehensive is that used by Bautmans et al., namely: “The fall is an unintentional coming down to the floor or a lower level, not induced by a major intrinsic or extrinsic event” (Bautmans, Jansen et al. 2011). The definition chosen by the Word Health Organization and edited by the U.S. Preventive Services Task Force is quite similar, namely “a sudden, unintentional change in position causing an individual to land at a lower level, on an object, the floor, or the ground, other than as a consequence of sudden onset of paralysis, epileptic seizure, or overwhelming external force” (Feder, Cryer et al. 2000).

2.3.2. Fall(s) risk factors

According the published report of the technical meeting organised by the World Health Organisation on falls prevention in older age which took place in Victoria, Canada in February 2007, several societal and individual aspects have to be considered to reach an integrative point of view of the fall risk factors that may be present among older adults. These factors are summarized in the following figure, taken from the report:



Accordingly, it is not surprising that the literature regarding the risk factors for falls is plentiful. Systematic reviews are available and three literature reviews and meta-analyses (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013) considering the fall risk factors recognized among community dwelling older will be further detailed in this chapter.

According to Bloch et al.(Bloch, Thibaud et al. 2013), fall risk factors can be divided into extrinsic risk factors, iatrogenic risk factors, and intrinsic risk factors.

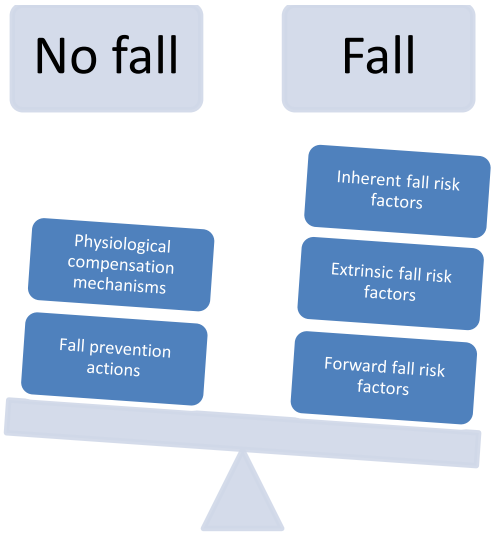
Extrinsic risk factors include walking aids, non-adapted shoes, or obstacles on the ground, or in the environment (Rubenstein 2006, Bloch, Thibaud et al. 2013).

Iatrogenic risk factors include laxatives, sedatives, antidepressants, anti-epileptics, antiparkinsonians, digoxin, analgesics, anti-inflammatory drugs, angiotensin-converting enzyme (ACE) inhibitors, vasodilators, and antihypertensive drugs (Bloch, Thibaud et al. 2013). Polypharmacy, defined as > 4 drugs per day (Tinetti and Kumar 2010), was also shown to be an independent risk factor for falls. Furthermore, the intrinsic risk factors for fall(s) include: fall history (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), fear of falling (Bloch, Thibaud et al. 2013), sensory disorders including proprioceptive, hearing and visual impairments (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), balance impairment (Rubenstein 2006, Tinetti and Kumar 2010), decreased muscle strength (upper and lower limbs) (Tinetti and Kumar 2010), gait impairment or walking difficulty (Rubenstein 2006, Tinetti and Kumar 2010), dizziness or postural hypotension (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), vertigo (Rubenstein 2006), drop attack (Rubenstein 2006), delirium (Rubenstein 2006), disorientation (Bloch, Thibaud et al. 2013), cognitive impairment (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), basic and instrumental activity of daily living disabilities (Rubenstein 2006, Tinetti and Kumar 2010), age > 80 years (Tinetti and Kumar 2010), female gender (Tinetti and Kumar 2010), low body mass index (Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), urinary incontinence (Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), pain (Tinetti and Kumar 2010), Parkinson's disease (Bloch, Thibaud et al. 2013), neurological disease (Bloch, Thibaud et al. 2013), digestive disease (Bloch, Thibaud et al. 2013), stroke (Bloch, Thibaud et al. 2013), high blood pressure (Bloch, Thibaud et al. 2013), depression (Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), anaemia (Bloch, Thibaud et al. 2013), fracture history (Bloch, Thibaud et al. 2013), cardiac and vascular problems (Bloch, Thibaud et al. 2013), cancer (Bloch, Thibaud

et al. 2013), arthritis (Rubenstein 2006, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), diabetes (Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013), alcohol (Rubenstein 2006), epilepsy (Rubenstein 2006), falling from bed (Rubenstein 2006), acute illness (Rubenstein 2006).

Moreover, Rubenstein et al. (Rubenstein 2006) have shown that among older adults without a history of falls in the previous year, gait and balance impairment were the most consistent risk factors, followed by medications, orthostatic hypotension, visual impairment, limitation in the activities of daily living (ADLs) and cognitive impairment. Indeed, in most cases, older adults presented more than one risk factor for falls. Even in those having the same risk factors, the consistency of each risk factor will differ depending on the individual compensation mechanisms of each person. These individual abilities to compensate (or not) for the presence of one or more fall risk factor(s) and different daily-life environments probably explain the wide range of odds ratios or relative risks obtained in observational or interventional studies (Bloch, Thibaud et al. 2010, Tinetti and Kumar 2010, Bloch, Thibaud et al. 2013).

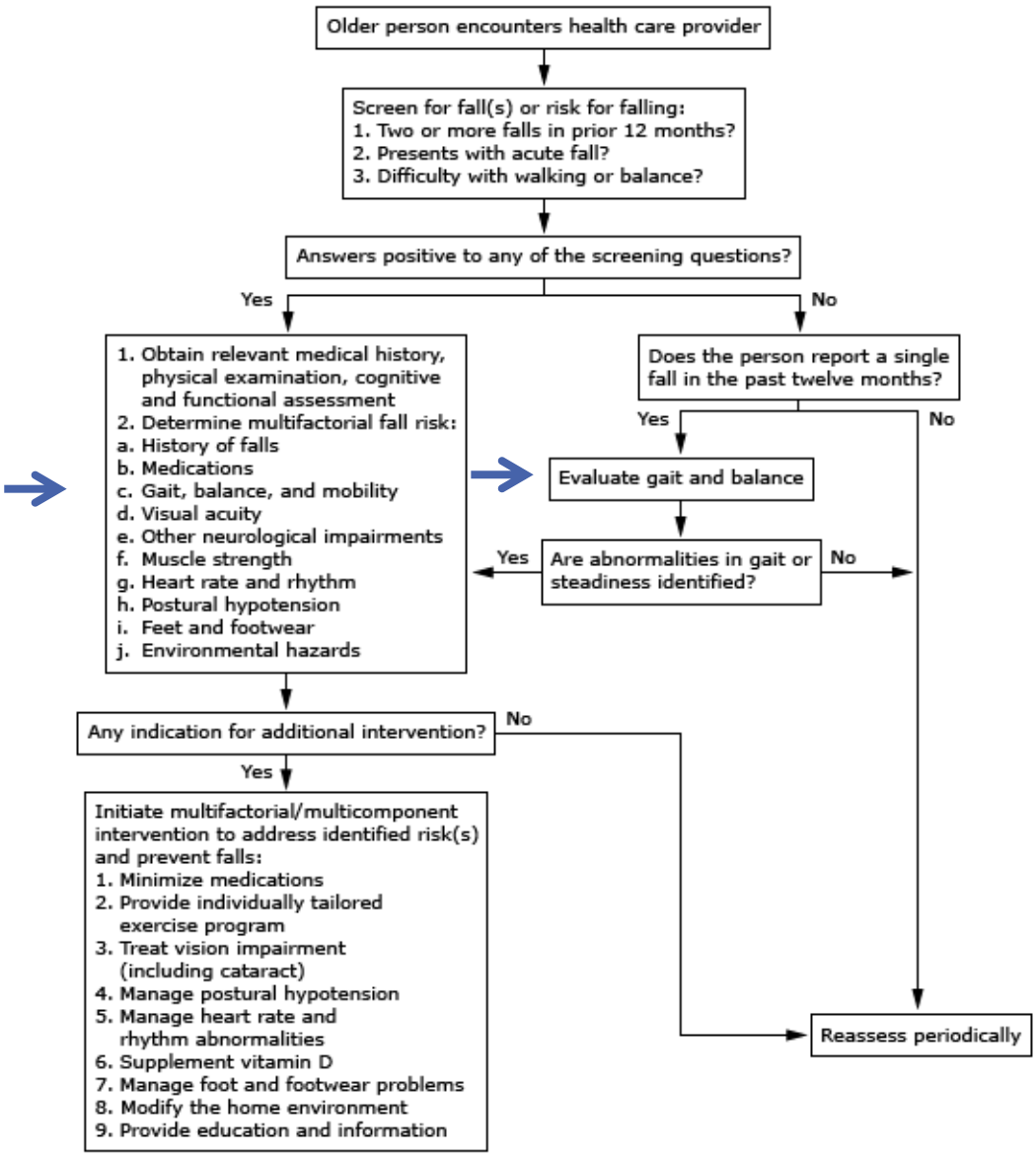
Finally, these relationships are constantly changing across the life-time as health-related conditions, activities or environments vary. Thus, the fall event could be considered as the result of an impaired balance between the fall risk factors and the compensatory fall risk mechanism, bearing in mind that this balance is individual, and varies across the life-time.



The inherent risk factors for fall(s) have been previously been summarized. According to Rubenstein et al. (Rubenstein 2006), extrinsic fall risk factors that merit consideration include environmental hazards, poor footwear and restraints. The same author considers trips and slips, acute medical illness, and dizziness as forward-fall risk factors.

2.3.3. Guidelines for fall detection

Obviously, assessing fall risk includes the detection of extrinsic, iatrogenic and intrinsic risk factors. This time-consuming step needs to be comprehensive in order to be effective. Accordingly, the American Geriatrics Society has issued a consensual fall prevention algorithm (see below)(Kenny, Rubenstein et al. 2011).



According to this algorithm, the first step is to assess fall history, and any potential difficulties with walking or balance. In fact, the algorithm has included a comprehensive

anamnesis similar to that already detailed in the previous chapter. The most important point to note is that, irrespective of whether the person has walking or balance complaints or a fall history, gait and balance must be assessed.

2.3.4. The functional tools used to address the fall risk

Several tools are available and validated for obtaining either a qualitative description or a quantitative evaluation of gait, transferring and balance. Unfortunately, these functional tools not always validated for use among elderly people, and threshold values related to community dwelling-older are not always published.

According Bloch et al. (Bloch, Thibaud et al. 2013), the most important tests to assess fall risk are as follows: abnormal walking test, slow walking speed, to be unable to get up from a chair, abnormal Tinetti test score, abnormal balance test, reduction of step length, lower grip strength. Unfortunately, these recommendations did not provide threshold values and have an obvious ceiling effect among robust or pre-frail people living independently at home. A systematic literature review addressing the use of functional tools in fall risk detection has shown that the most popular functional tools used include the performance-oriented mobility assessment, four square step test, short physical performance battery, Berg balance scale, mini-balance evaluation system test, dynamic gait index, Timed Up and Go test and dual task walking (Ambrose, Paul et al. 2013).

Finally, since the fear of falling can lead to gait modifications (Maki 1997) and increase the fall risk (Scheffer, Schuurmans et al. 2008), one of the available tools to assess walking confidence (FES-I, ABC scale, or SAFE) should also be implemented (Ambrose, Paul et al. 2013). Unfortunately, to the best of our knowledge, at the current time, none of these functional tests is validated as a predictive functional tool to detect, among older people, those at risk of a first fall.

2.3.5. The gait parameters to consider in assessing fall risk

The gait parameters that may be potentially useful when assessing fall risk are numerous, and the available literature is abundant in this regard. To keep this background section as targeted as possible, this chapter will focus solely on the gait parameters obtained using the accelerometric and opto-electronic methods detailed in the experimental section of this research work.

Considering the equipment used (further detailed in the experimental section), gait speed, stride length and frequency, stride regularity and symmetry, and minimal toe clearance (MTC) will be assessed. All studies cited in this chapter involved community-dwelling older adults. In cases where the gait parameters were not recorded in single-task and comfortable walking conditions, the walking conditions have been detailed.

A. Gait speed

Using a cross sectional study design, Cho et al.(Cho and Kamen 1998), Auvinet et al.(Auvinet, Berrut et al. 2003) and Cui et al.(Cui, Peng et al. 2014) reported that fallers walked significantly slower than non-fallers. Similarly, after defining groups according to the fall risk based on functional tools, Menz et al.(Menz, Lord et al. 2003), Senden et al. (Senden, Savelberg et al. 2012) and Bautmans et al. (Bautmans, Jansen et al. 2011) showed a significant effect of the fall risk category on gait speed. Using a longitudinal study design, Verghese et al. showed that slower gait speed was associated with a higher risk of falls in a model adjusted for age, gender, education, falls, chronic illnesses, medications, cognition, disability and for functional gait and balance test scores (Verghese, Holtzer et al. 2009). Consequently, in this cohort, each 10 cm/s decrease in gait speed was associated with a 7% increase in the risk of fall. Participants with a gait speed slower than ≤ 70 cm/s had a 1.5-fold increase in the risk of fall compared with those with normal speed, even after accounting for several fall risk factors (even fall within the

previous year). In a cohort of seniors living independently but in senior care facilities (according to the participants' characteristics published, they were less mobile and sometimes used walking aids) followed for fall(s) for one year, Maki et al showed that higher stride-to stride gait speed variability was associated with future fall(s) (OR. 2.30, CI. 1.17-4.51)(Maki 1997).

However, the relationship between gait speed and future fall(s) is not consensual. For example, in the longitudinal study by Hausdorff et al (Hausdorff, Edelberg et al. 1997) involving thirty-five community-dwelling elderly subjects aged over 70 years, walking independently for 6 minutes and categorized as fallers and non-fallers based on history, gait speed did not differ between fallers and non-fallers, while the stride-to-stride variability of temporal gait parameters was significantly different between the two groups (see below). This observation suggests that gait speed and stride-to stride temporal gait variability parameters could represent two different and independent fall risk factors.

B. Stride length

In a cross sectional study, Auvinet et al. (Auvinet, Berrut et al. 2003) reported that fallers had shorter stride length than non-fallers. However, in the longitudinal study by Verghese et al.(Verghese, Holtzer et al. 2009), the prospective relationship between stride length and future fall(s) was not found to be significant in the full adjusted model.

C. Stride frequency or cadence

In a cross sectional study, Auvinet et al. (Auvinet, Berrut et al. 2003) showed that fallers had significantly slower stride frequency than non-fallers ($p < 0.001$). Menz et al.(Menz, Lord et al. 2003) found no difference between groups in terms of step frequency assessed in steps/min. Senden et al.(Senden, Savelberg et al. 2012) showed that people with a Tinetti score ≤ 24/28 had slower step frequency than those with a Tinetti score > 24/28 ($p < 0.01$), while subjects with a Tinetti score between 19 and 24 showed no difference in stride frequency compared to those with a Tinetti score < 19/28. There was no difference between groups in terms of fall history. In the longitudinal study by Verghese et al. (Verghese, Holtzer et al. 2009), the prospective relationship between cadence and future fall(s) was not found to be significant in the final adjusted model (relative risk: 1.066 with CI: 0.984-1.155).

Furthermore, similar non-consensual relationships between cadence and fall(s) risk or history have also been found using different instrumental methods to assess cadence (Mortaza N 2014). In order to explain why the relationship between fall risk and stride frequency remains unclear, different hypotheses can be put forward. Either the stride frequency has a non-linear relationship with fall(s), or their relationship is mediated by a third component that is unrecognized at this time (e.g. gait speed, or stride regularity), or perhaps stride frequency has no relationship with fall(s) risk or/and fall(s) history.

Moreover, previous studies did not take into account the influence of gender and/or leg length or total body height in their comparison groups, introducing bias in the interpretation of their results.

D. Stride regularity

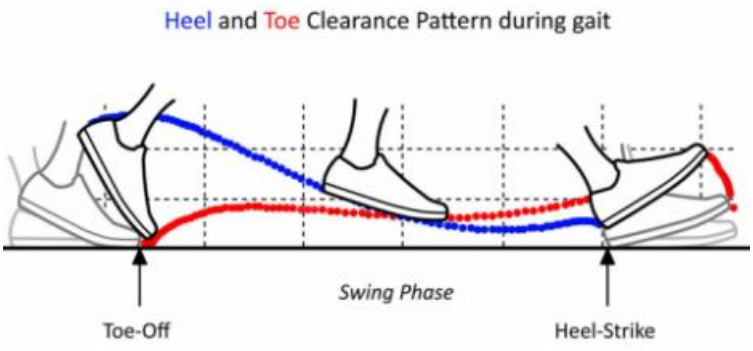
In Auvinet et al.(Auvinet, Berrut et al. 2003), stride regularity was automatically derived from two coefficients (C1 and C2) computed from an autocorrelation function on the cranial-caudal signal and quantifying the peak values of the first and second dominant periods of the autocorrelation function, respectively corresponding to the step regularity and the stride regularity (Moe-Nilssen and Helbostad 2004). These coefficients are automatically transformed according to equations detailed in the paper by Auvinet et al (Auvinet, Berrut et al. 2002). Stride regularity is calculated as (C1+C2) x100 and describes the similarity of vertical movements over successive strides (dimensionless). In their study (Auvinet, Berrut et al. 2003), Auvinet et al. showed that fallers had significantly lower stride regularity (191.3 ± 56.0) than non-fallers (291.9 ± 51.9) ($p < 0.001$).

E. Stride symmetry

According to Auvinet et al.(Auvinet, Berrut et al. 2002), stride symmetry is calculated as (C1/C2) x100 and describes the similarity of left and right cranial-caudal movements, and is independent of fluctuations in the successive movement of each limb. In their study (Auvinet, Berrut et al. 2003), Auvinet et al. reported that fallers had significantly lower stride symmetry than non-fallers ($p < 0.01$).

F. Minimum toe clearance (MTC)

The minimum toe clearance is defined as the local minimum in separation between the ground and the toe region of the forward swinging foot (Alan and John 2014).



Tripping while walking seems to be one of the causes of falls in older adults living at home (Blake, Morgan et al. 1988). The risk is highest at the point of MTC (Mills and Barrett 2001). Actually, the risk of tripping is the result of the proximity of the swing foot to the ground, the high velocity of the swinging foot and the forward-travelling centre of mass being in front of the base of support (Winter 1992, Mills and Barrett 2001, Barrett, Mills et al. 2010). Indeed, a one-year prospective study including 8 young people (men age 47 ± 12 years) with transtibial amputation showed a positive relationship between lower MTC measured on the prosthetic-side and future trips (Rosenblatt, Bauer et al. 2016).

Killeen et al. measured MTC in 121 healthy individuals aged 20-80 during normal treadmill walking and in dual task walking. During normal treadmill walking, there was no difference between the age groups in terms of mean MTC values or standard deviation. However, age differences became apparent during the cognitive dual-task, whereby older adults showed significantly smaller mean MTC values (Killeen, Easthope et al. 2017). An additional study showed similar decreased foot clearance during stair negotiation in the dual task condition (Hashish, Toney-Bolger et al. 2017). Furthermore, a study by Schulz et al., including 10 volunteers ranging in age from 22–58 years (44 ± 13 years), confirmed the decrease in mean MTC values during dual task walking. The same study also highlighted that the MTC increases

or decreases depended on the nature of the specific concurrent task performed, and were not linearly correlated with toe speed, and were independent of the gait variability of step length, step width, step time also assessed in several dual tasking conditions(Schulz, Lloyd et al. 2010).

To summarize the relationships between mean MTC and fall(s), MTC could be a potential fall risk marker given its critical biomechanical aspects (Schulz, Lloyd et al. 2010). In fact, mean MTC is lower in people who have a fall(s) history and during dual task walking. Nevertheless, to date, and to the best of our knowledge, no study has assessed the relationship between mean values of MTC (obtained either in comfortable or in dual task walking conditions) and the risk of future fall(s) among healthy older adults.

Considering the relationships between gait variability and fall risk (Hausdorff, Edelberg et al. 1997, Maki 1997, Hausdorff, Rios et al. 2001), the variability in MTC has also been considered as a potential marker of fall risk. Indeed, a first study assessing minimum foot clearance (measured as MTC) showed a significant relationship between the minimum foot clearance variability measured during treadmill walking and fall(s) history (Khandoker, Palaniswami et al. 2008). Moreover, studies assessing the relationship between the ageing process and variability of the MTC showed that, among healthy older adults without fall history and relative to normal walking conditions, the variability of MTC in dual-task walking did not significantly change (Hamacher, Hamacher et al. 2014, Santhiranayagam, Lai et al. 2015), whereas stride length and stride time variability increased (Hamacher, Hamacher et al. 2016). According to these authors, older adults with an intact central control mechanism of MTC are not more prone to fall in a dual task condition, while older adults with impaired central control mechanisms would be at risk of falling because of tripping in dual task walking.

To summarize, available literature shows that MTC variability increased after fall history, MTC variability remains stable among healthy older people, and MTC variability seems to be independent of the variability of stride length and time. To date, and to the best of

our knowledge, no study has assessed the relationship between MTC variability (obtained in either normal walking or dual task walking conditions) and the risk of future fall(s) among healthy older adults.

2.2.4. Summary of the rational supporting this research

In summary, relationships between gait parameters and fall(s) history and/or fall risk have been partially identified in cross-sectional and longitudinal studies. Since cross-sectional studies are more numerous than longitudinal studies, the relationship between gait parameters and fall risk over time are not fully understood.

In fact, slower gait speed, shorter stride length, higher swing time variability, higher stance time variability, lower minimal toe clearance and higher minimal toe clearance variability in single-task walking seem to have a relationship with fall history, while lower minimal toe clearance seems to have a prospective relationship with trips but the relationships with future falls among people without fall history is not yet clearly established.

Furthermore, a recent systematic literature review concluded that, while available results support the usefulness of gait parameters obtained using accelerometric methods, further prospective studies remain necessary to further elucidate the prospective relationships between gait parameters and future fall(s), as well as the usefulness of gait parameters to identify, among community-dwelling older adults, those at risk of future fall(s) (Gillain, Boutaayamou et al. 2018).

Regarding fall consequences, both for the individual and at a societal level, the major challenge is to discern, among older adults, those who are at risk of falling, before the first fall and its negative consequences occur. According to previous findings, it appears logical to assess the usefulness of gait parameters, obtained in comfortable and in more challenging walking conditions, using instrumental method(s) in order to identify at an earlier stage, among older people, those at risk of future fall(s).

In this context, the main goal of this work is to better understand the underlying relationships between gait parameters and falls among older adults. To this end, I have chosen to assess gait patterns using two instrumental methods (accelerometric and opto-electronic) and in three different walking conditions (comfortable, fast and dual tasking).

In light of the previously outlined literature on the topic, I hypothesized that gait parameters obtained in comfortable, fast or dual task walking conditions and their changes would be useful to identify, among healthy older adults, those at risk of future falls. I also hypothesized that, individual gait patterns, being the results of individual compensatory mechanisms, could be the most efficient marker to identify, among community-dwelling subjects free of fall history, those at risk of future fall(s). Finally, I hypothesized that data mining could be useful to confirm the second hypothesis.

The next section will outline the experimental methods used in this work.

3. Objectives

The aims of this exploratory study were: (1) to investigate the relationship between gait parameters and future fall(s) among older people living independently at home, free from disease-related processes recognized as a fall risk factors, including a history of recent fall(s); (2) at the level of the cohort, to assess the predictive association between these gait parameters and future fall(s); (3) at the individual level and using data mining, to explore the usefulness of gait parameters in classifying the volunteers as being at risk of a fall or not.

4. Methods

4.1. Study design

In order to achieve our objective, we performed a two-year longitudinal, observational study.

4.2. Participants

Volunteers were invited through a publicity campaign in national and local news media, or recruited during a meeting organized for the general public focusing on healthy ageing issues. Inclusion criteria were: Individuals aged at least 65 years, living independently at home, who could understand French, and who were able to travel to the motion analysis laboratory and who provided written informed consent. Exclusion criteria were: a history of fall(s) in the previous year, need for walking aids, gait disorders and/or an increased fall risk related to neurological or osteoarticular disease (e.g. stroke, Parkinson's disease, lumbar spinal stenosis or polyneuropathy), dementia, recent hip or knee prosthesis (≤ 1 year), musculoskeletal pain when walking, acute respiratory or cardiac illness (< 6 months), recent hospitalization (< 3 months), untreated or uncontrolled comorbidities (e.g. hypertension, diabetes), presence of a cardiac pacing device (an exclusion criterion to apply bioelectrical impedance analysis).

4.3. Anamnesis, clinical and functional assessment

At inclusion (T0), all participants underwent anamnestic, clinical and functional assessment and gait analysis. The aims were to: (1) verify the inclusion and exclusion criteria, (2) investigate the relationship between gait performance and social, clinical and functional status. Subjects were assessed for age, gender, education, medications, alcohol and tobacco consumption and self-perceived quality of life assessed using the question “Currently, are you globally satisfied with your quality of life” where the volunteers had to answer “yes” or “no”.

The burden of medical and surgical histories was scored by the Cumulative Illness Rating Scale geriatric version (CIRS-g) (Linn, Linn et al. 1968, Miller, Paradis et al. 1992). The total score ranges between 0 and 56 and combines the number of diseases and their functional impact.

Physical activity, exercise and sports habits were assessed by the Physical Activity Status Scale (PASS) developed by Jackson et al. (Jackson AS 1990, Jackson and Ross 1996). The total score ranges between 0 and 7, with a score of zero indicating no exercise or physical activity at all and a score of seven indicating at least three hours of intense physical exercise per week.

Acute or chronic pain perceived before testing was measured using a visual analogue pain scale (Hawker, Mian et al. 2011) scored from 0 (no pain) to 10 (maximal pain the subject can imagine). Standardized specific material including one graduated side used to measure the intensity of the pain (the experimenter side) and another, ungraduated side used by the subject to evaluate their own pain without taking numerical value into account. The subject had to move the red cursor from left to right on a blue line ungraduated to locate potential pain. Once finished, the experimenter looked at the corresponding value on the graduated side and noted the intensity of the pain.

Functional assessment included the activities of daily living (ADL) (Katz, Ford et al. 1963) and the instrumental activities of daily living (IADL) scales (Lawton MP 1969). The ADL is a score ranging from 6 to 24, where a lower score corresponds to higher autonomy in the activities of daily living. The IADL is a score ranging from 0 to 8, where a lower score corresponds to less autonomy in the instrumental activities of daily living. Considering that some housework is usually and preferentially done by the same member of the family (e.g. gardening or preparing meals), the IADL score was calculated as the sum of the scores obtained on the items applicable to each subject, divided by the sum of the maximum possible score on the applicable items (in order to not consider the volunteer disabled for a housework item he or she never performs) (Gillain, Warzee et al. 2009).

Mood was assessed using the geriatric depression scale short version (GDS-4) (D'Ath 1994, Clément 1997), which is a score ranging from 0 to 4, where 0 means no depressive symptom is present. A score ≥1/4 means depressive disorder may be suspected (Clément 1997). Anxiety was assessed using the Covi anxiety rating scale (Lipman 1982), which yields a score ranging from 3 to 15, where a higher score means higher anxiety complaints.

Cognitive performance was assessed using the Montreal cognitive assessment (MoCA)(Nasreddine, Phillips et al. 2005), which yields a score ranging from 0 to 30, where a higher score means higher overall cognitive performance. A score ≥ 26/30 indicates that the subject does not have cognitive disorders (Nasreddine, Phillips et al. 2005).

Nutritional status was assessed using the Mini Nutritional Assessment short version (MNA-14) (Rubenstein, Harker et al. 2001), which yields a score ranging from 0 to 14, where a score from 12 to 14 means the subject has no risk of malnutrition, a score from 8-11 indicates a risk of malnutrition, and a score less than 8 indicates that the subject is malnourished.

Frailty was detected using two different tools, namely the Gérontopôle frailty screening tool (GFST) (Vellas, Balardy et al. 2013), and the Edmonton Frail Scale(Rolfson DB 2006). The

GFST is validated for its specificity to detect frail people among community dwelling adults in the general practice setting. In this screening tool, the subject has to answer 6 questions to detect whether the patient suffers from social isolation, weight loss, weakness, mobility or memory problems, and slower gait speed within the previous three months. If at least one question is answered positively, then the practitioner has to subjectively answer the question "you're your patient seem to be frail?" The result obtained is a binary answer (yes/no). The Edmonton Frail Scale is a more sophisticated screening tool assessing physical, cognitive, and social components of frailty and scored out of a total of 17. A score from 0 to 3 means no frailty, a score from 4 to 5 means mild frailty, a score from 6 to 8 means moderate frailty and a score ≥ 9 indicates that the subject is very frail.

The fear of falling was assessed using the French version of the falls efficacy scale (FES-I) (Delbaere, Close et al. 2010), in which 16 activities of daily living are investigated for fall concern and scored from 1 to 4, where 0 means the subject is confident during the activity.

Clinical evaluation included a visual examination of spontaneous gait in order to exclude pain, limp or lateral motor deficit during gait. To assess and quantify any extrapyramidal stiffness, the examiner applied the unified Parkinson's disease rating scale criteria (UPDRS)(Goetz, Tilley et al. 2008). Stiffness ranged from 0 to 3 (0 = no stiffness; 3= stiffness decreasing the amplitude of the movement). As recommend by the Task Force, the measure of stiffness was applied for the neck and for the two upper and lower limbs. The total UPDRS score ranges from 0 to 12. Distance vision was tested using the French Monoyer's scale for 3 meters (Casellato 1994). Visual acuity less than 5/10 was reported as a visual impairment. Anthropometric data assessment included the measure of body height, weight, waist circumference and hip circumference and the length of the right leg. The body mass index was calculated as the weight (in kilograms) divided by the height (in m) squared. The use of a bioelectrical impedance analysis device (BodyStat® 1500) enabled us to obtain resistance in

ohms. This amount was then included in the following validated equation from Janssen estimating skeletal muscle (SM) mass, as follows: SM mass in kg = [(Ht^2/R x 0.401)+(gender X 3.825)+(age X - 0.071)]+5.102, where Ht is height in centimetres; R is BIA resistance in ohms; for gender, men=1 and women=0; and age is in years (Janssen, Heymsfield et al. 2000). In order to adjust for stature and the mass of non-skeletal muscle tissues (fat, organ, bone) and according to Janssen et al. (Janssen, Heymsfield et al. 2002), the skeletal muscle mass was expressed as a skeletal muscle mass index (SMI) where SMI = skeletal muscle mass/body mass x 100 and expressed in %.

Grip strength, muscle fatigue resistance (time in seconds while the contraction is over 50% of the maximal contraction force) and the grip work (fatigue resistance x 75% of the maximal grip strength) of the dominant hand were assessed with a Martin's Vigorimeter used as per Bautmans et al. (Bautmans I. 2005).

Gait and balance were assessed by the Timed Up and Go (TUG) test (Podsiadlo and Richardson 1991), the Functional Gait Assessment (FGA) (Wrisley, Marchetti et al. 2004) and the Short Physical Performance Battery (SPPB) (Guralnik, Simonsick et al. 1994). During the TUG test the subject was asked, at the signal given by the experimenter, to stand up from a chair without armrests, to walk at a self-selected comfortable speed over a straight forward distance of 3 meters, turn around, and come back and sit down in the chair. The time required to perform this test was measured using a stopwatch manually started at the signal, and stopped when the subject was sitting in the chair at the end of the test. Functional gait assessment investigating static and dynamic balance and walking is scored out of 30 where a higher score means higher balance performance. The SPPB investigating static balance, walking speed and standing up performance is scored out of 12, where a higher score means better physical performance.

4.4.1. Material

A. The Locometrix®

The Locometrix is a validated gait analysis system including a 3-D-acceleration sensor, a data logger and a computer program for processing the acceleration signals and calculating the gait parameters (see figure below). The sensor weighs 20 g. and is composed of three accelerometers placed perpendicularly to each other housed in a plastic box (40x18x18mm). The sensor is incorporated into an elastic belt, which was placed around the waist of the subject in the lumbar position close to the L3-L4 inter-vertebral space. The first accelerometer was aligned with the cranio-caudal axis of the body, the second with the anterior axis and the third with the medio-lateral axis.

Locometrix placed in the elastic belt worn in the lumbar position

Signals were recorded by a data logger at a sampling frequency of 100 Hz and an anti-aliasing filter of 50 Hz was applied (Mignardot, Deschamps et al. 2014).

Except for gait speed, which is calculated by the software based on the time measured with an electronic stopwatch synchronized with the gait data logger, the Locometrix® software automatically calculates gait variables considering a steady state walk sample of 20.48s.

According to the validation data relating to the Locometrix® (Auvinet, Berrut et al. 2002), a period of steady state walking of 20.48 s was selected from the recording of each subject. This period contained about 1024 acceleration measurements and provided an optimal calculation time corresponding to 19–21 gait cycles, about 28 m for healthy adult subjects. In addition, the validation paper showed a figure summarizing gait cycle events identified based on the vertical acceleration curve. Actually and according Auvinet et al., kinetic data recorded by the Locometrix® were compared with the kinematics data from a synchronized video image analysis, making it possible to detect heel contact, foot flat, opposite toe-off, mid-stance and initial pushoff and to calculate right and left stance phases and initial and terminal double supports based on right and left cranial–caudal accelerations (unpublished data). According Auvinet et al. the software use a fast Fourier transformation to convert the cranial–caudal acceleration signal to the fundamental frequency of periodic movement, the step frequency. By definition, a complete stride includes two steps; thus stride frequency (SF) was calculated as one half the fundamental frequency and is expressed in Hz or strides per s. The stride length (SL) was calculated from the average speed (m/s) divided by the SF (Hz) and expressed in m/s. Stride symmetry (SYM) and regularity (REG) were derived from two coefficients of correlation, C1 and C2, obtained by calculating the autocorrelation function of the vertical acceleration signal. Stride symmetry describes the similarity of left and right cranial– caudal movements and is independent of fluctuations in the successive cranial–caudal movements of each limb. Stride regularity describes the similarity of vertical movements over successive

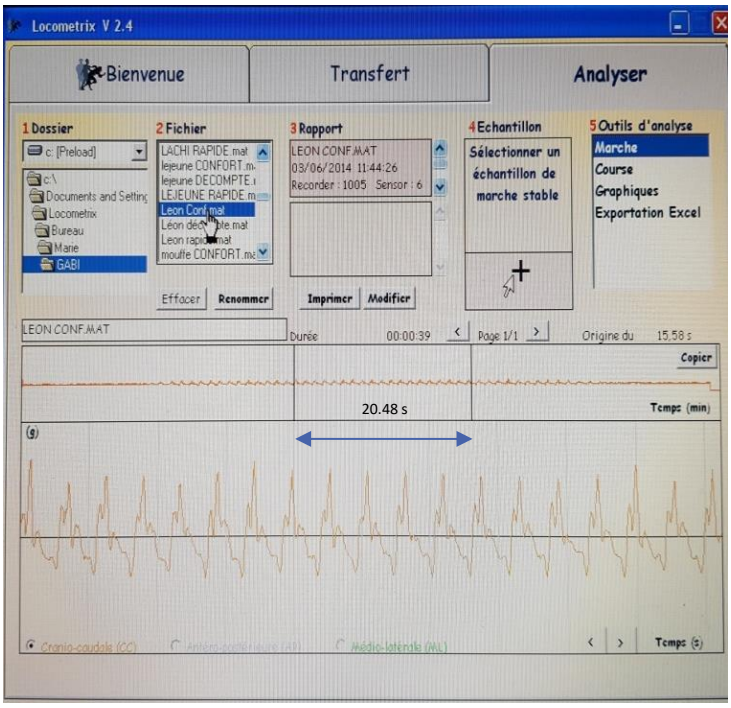
strides. Symmetry and regularity are dimensionless. The validation of the measurement of these parameters, their reliability and their use in the clinical setting has been reported by Auvinet et al. (Auvinet, Berrut et al. 2002).

In the University of Liège, the Locometrix was bought by the Motricity sciences department and has been used in a number of research protocols (Gillain, Warzee et al. 2009, Beaudart, Maquet et al. 2013, Buckinx, Beaudart et al. 2014, Buckinx F 2015, Gillain, Dramé et al. 2015) with expertise using the instrument in this field of research.

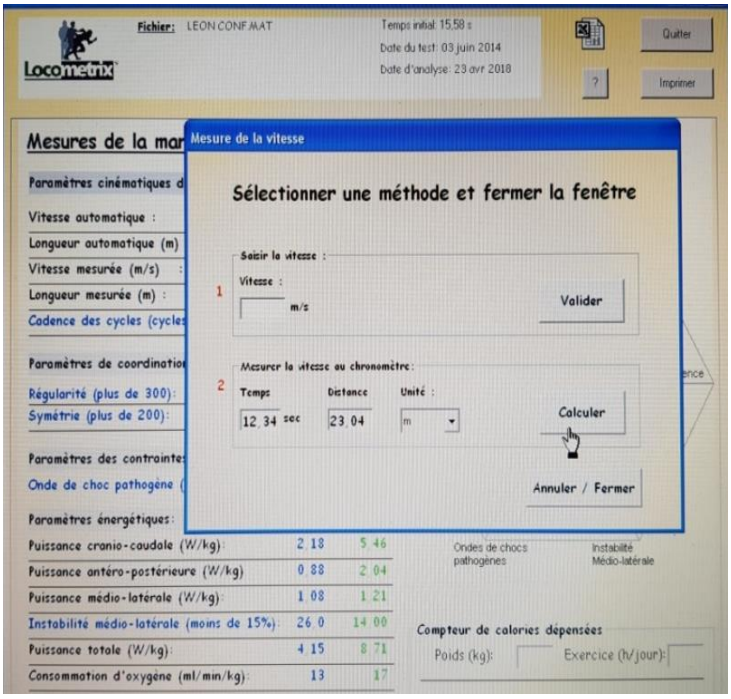
The following pictures show the extraction data steps.

First, the accelerometer is connected to the computer and the user interface of the software is opened.

Second, the data are transferred from the accelerometric device to the computer.

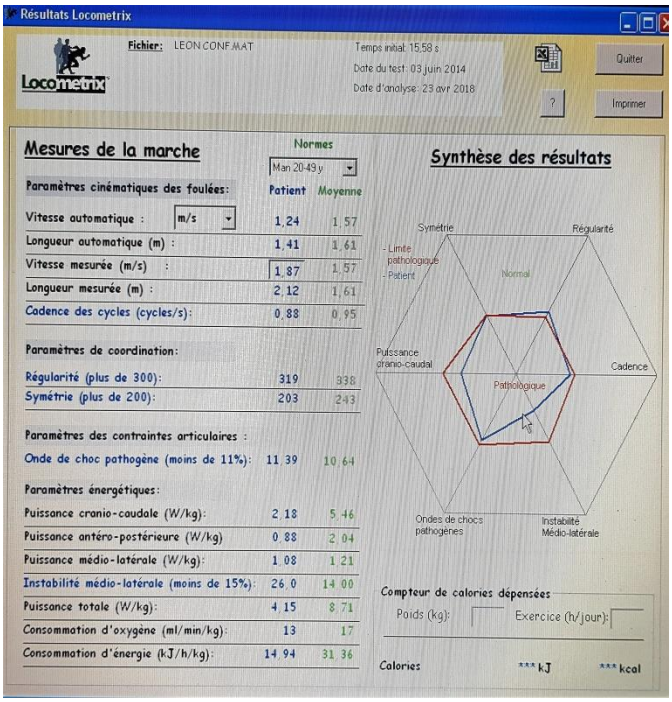


Third, and as notified in red on the picture, the process includes the following steps: - 1 the selection the filed containing the walk sample -2 and 3 the selection of the walk sample to process, - 4 the selection of the walk period to analyse by moving the cursors (considering the distance between the two cursors is always the same and corresponds to a 20.48 second period), -5. the selection the movement to analyse is a walk (rather than a run) and check that cranio-caudal curves are selected for analysis (usually preset).



Then, the user interface asks the user to manually insert the whole distance walked and the time required to walk this distance.

Of note, gait analyses processed by the software are based on the walking period of 20.48 seconds previoulsly selected by moving cursors, while the gait speed will be calculated based on the walking distance and time taken that were manually inserted.

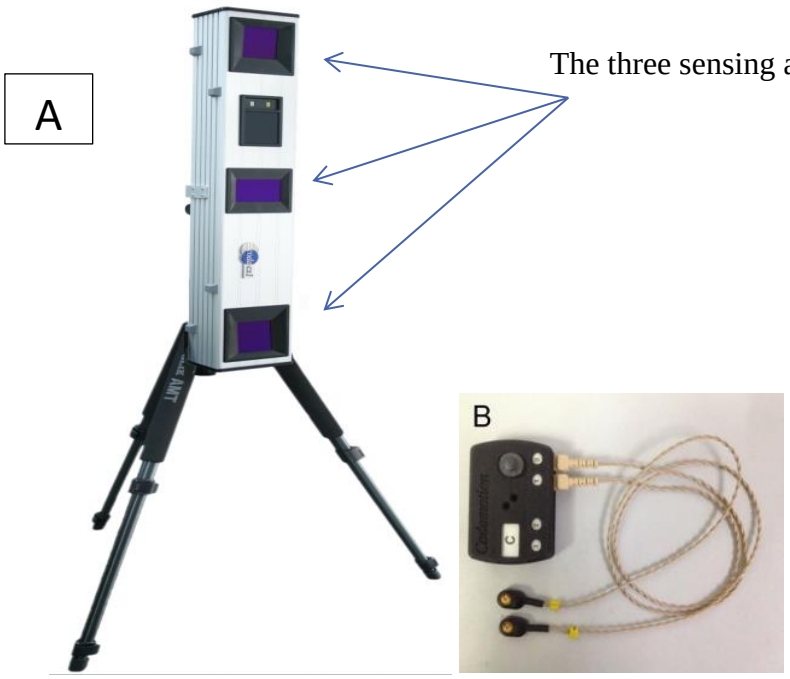


The last window shows the gait performance and a graphical view.

An export option to Excel is available.

B. The CodaMotion®

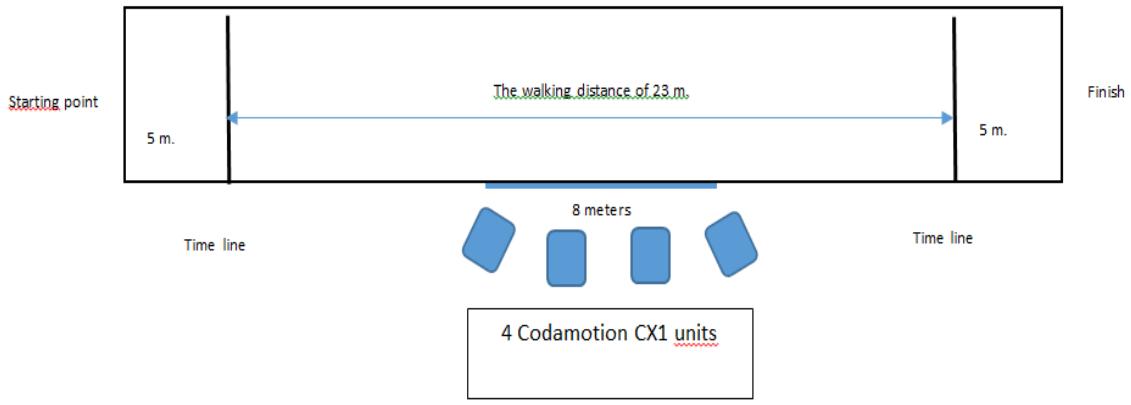
The kinematics system used in this project is based on an active optical system able to accurately measure the 3D position of active markers placed on the body on the points of interest (e.g. ankle, knee, foot). The Codamotion® system (Charnwood Dynamics, Rothley, UK) is a 3-dimensional kinematic tool validated for use in laboratories (Gorton Iii, Hebert et al. 2009, Schwartz, Denoël et al. 2015). The use of position markers attached to the feet of the volunteers enables the application of the kinematic system to gait analysis while the 3-dimensional position and orientation of the feet were tracked using Codamotion CX1 units (cameras) (see picture A below). The Codamotion system used in this research protocol includes active position markers connected with the battery pack (see picture B below), a battery pack able to support 4 position markers, 4 Codamotion CX units (cameras), each including 3 sensing arrays (picture A), and software to capture kinetic data from each position marker.



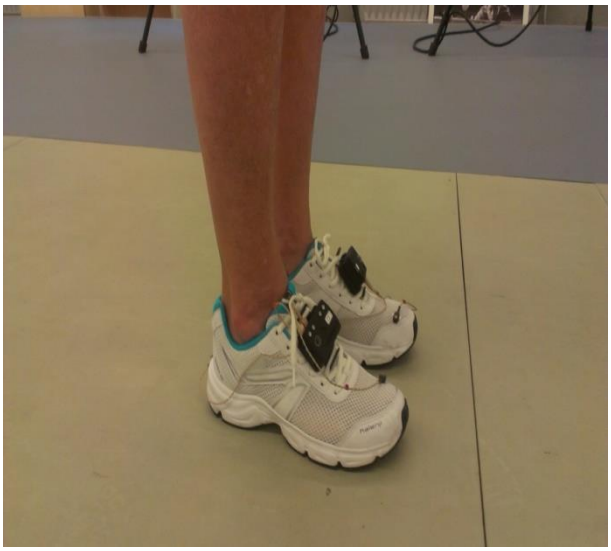
4.4.2. Data acquisition

A. Calibration, installation of equipment

Prior to first data acquisition, the experimenter was trained to use the opto-electronic method. In order to avoid measurement error and according to the gait parameters to be measured, a specific set up was dedicated to this study. The camera position was standardized as shown in figure below.



In order to ensure standardized gait parameter acquisition conditions, a specific procedure was followed when placing the markers on the volunteers.



Two active position markers were attached on the left and right shoes of the volunteer on the extremity of the toe. Two active position markers were attached to the left and right shoes of the volunteer in the heel position in the middle of the shoe for the medio-lateral axis, and above the shoe heel for the cranio-caudal axis.

Furthermore, all volunteers wore an elastic belt in the lumbar position containing the accelerometer as recommended by Locometrix.

Furthermore, before inclusion of older participants and in order to ensure that data acquisition using acceleration-based methods was as accurate as possible, and considering opto-electronic methods as the gold standard, a “validation” step was performed.

In this validation step, 19 young volunteers, equipped with the two instrumental methods, were assessed for gait speed, stride length and stride frequency during five walking periods.

The within-subject reliability of these 3 gait parameters using two different instrumental methods was measured using the intra-class correlation coefficient.

The ICC was excellent (> 0.95) for all three parameters.

These data remain unpublished but were presented in poster format during the annual congress of the Belgian Society of Geriatrics and Gerontology in October 2014 I Liège, Belgium (See Appendix 1).

B. Gait tests

According to previous data acquisition rules, gait tests were performed by the same operator in a wide, clear, straight hallway. Subjects wore their own usual shoes (shoes had to have laces in order to attach the battery box). Walking was recorded under 3 different experimental conditions: self-selected comfortable walking speed (CW), self-selected fast walking speed (FW) and during a dual-task walking condition (DTW). The instruction for FW was to “walk quickly without running in order not to miss your train” and for DTW “walk at a comfortable walking speed while simultaneously doing mental arithmetic task (a seven counting down from the number 100 task). Because explicit instructions concerning priority could affect dual tasking (Yogev-Seligmann, Rotem-Galili et al. 2010), the single instruction given to volunteers was to “do both tasks simultaneously as well as you can”.

Since the order of the different walking conditions (CW, FW or DTW) could influence gait performance, CW is always the first walking condition requested, and the DTW is randomized in order to avoid systematic measurement error. Indeed, one volunteer out of two was asked to perform CW, FW and DTW while the other was asked to perform CW, DTW and FW.

At the starting point, the volunteer was asked to walk 5 m in order to increase their walking speed and reach a steady walking state. The examiner starts the stopwatch when the first foot of the volunteer crosses the time line. The time required to walk 23 m is assessed. The last 5 m are used to decrease the walking gait and stop. In the middle of the 23 m, four position cameras compute the position of heels and toes over 8 meters.

4.4.3. Data extraction

Using Locometrix®, the software automatically extracts the data relative to the walk and calculates the gait parameters. After the data acquisition, the experimenter connects the accelerometer to the computer. Next, the software opens a window showing the entire walk period recorded, and among those, the experimenter has to select the 10.24 second period of stable walking that is to be considered for the calculation of gait parameters. After that, the distance walked (23 meters in this study) and the time required to walk it were manually entered by the experimenter, before pressing on an optional button to calculate the gait parameters (stride length, stride frequency, stride regularity and symmetry) based on the walk period selected by the experimenter. Gait speed was calculated based on the distance and the time required walking the distance, as manually entered by the experimenter.

Using Coda Motion®, the position signal obtained using the opto-electronic method was extracted using validated software automatically detecting gait events (heel strike and toe off) and calculating the gait parameters further explained below. The software allows visual control of gait event detection.

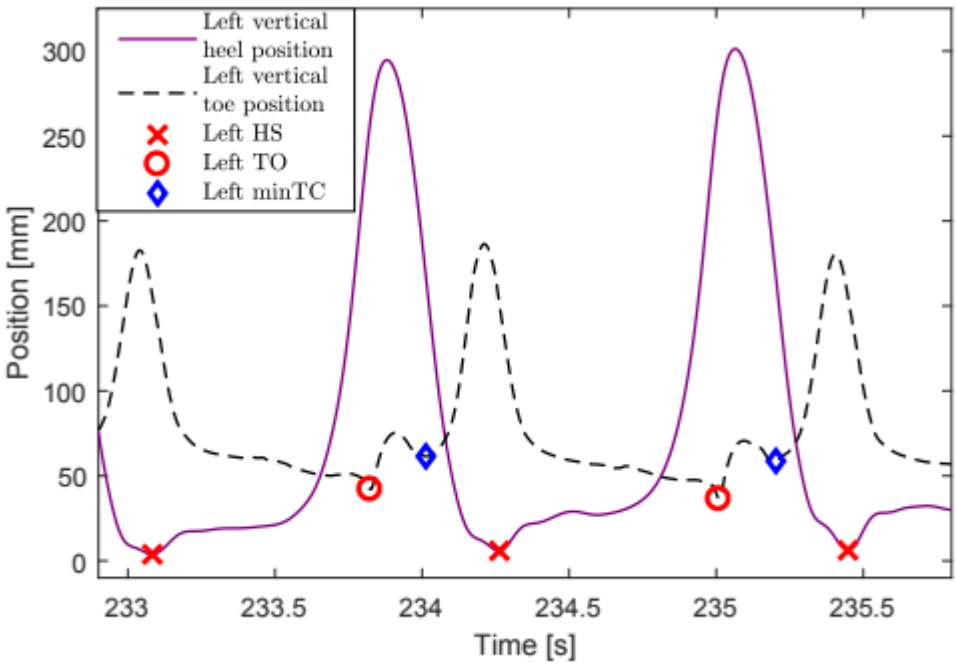
4.4.4. Data processing

Concerning Locometrix®, mean gait speed (GS) was computed from the time required to cover 23.04 meters according to the following formula: 23.04 (m) / time elapsed (s). Time was recorded by means of a stopwatch. The following gait variables were inferred from walking speed and 20.48 s of stationary cranial-caudal acceleration signal:

- the stride frequency (SF) or number of cycles per second (Hz) was calculated from the cranio-caudal acceleration following the application of a Fourier transform.
- the stride length (SL) was deduced from the equation [speed (m/s)= frequency (Hz) x stride length (m)] and expressed in meters.
- the regularity (REG) and symmetry (SYM) indices were automatically derived from two coefficients (C1 and C2) computed from an autocorrelation function on the cranial-caudal signal. The autocorrelation coefficients C1 and C2 quantify the peak values of the first and second dominant periods of the autocorrelation function, respectively corresponding to step regularity and stride regularity (Moe-Nilssen and Helbostad 2004). These coefficients are automatically transformed according to equations detailed in the paper by Auvinet et al. (Auvinet, Berrut et al. 2002). Regularity and symmetry indices are dimensionless. The regularity index describes the similarity of vertical movements over successive strides. The symmetry index describes the similarity of left and right cranial-caudal movements and it is independent of fluctuations in the successive movement of each limb (Auvinet, Berrut et al. 2002).

Concerning the position signal obtained from opto-electronic method, a signal-processing algorithm was then applied to these recorded coordinates to extract the heel strike (HS) and toe-off (TO) timings for the right and left feet. This algorithm is based on a piecewise linear fitting method (Boutaayamou, Schwartz et al. 2015) that identifies accurately, on a stride-by-stride basis, HSs and TOs as times associated with local minima in Zh and Zt, respectively (see figure below).

In addition, we extracted the minimum toe clearance time (minTC) as the first sampling period at which Zt reaches a minimum during the swing phase prior to HS (Figure XXXX). We visually checked the extraction of HS, TO, and minTC timings for each measured left/right gait cycle.



The minimum toe clearance position (MTC) is calculated as Zt (minTC(i)) – Zt(TO(i)). MTC from the left or the right foot are equally considered as MTC.

After comparison analysis showing that the MTC obtained during the swing phase of the left foot was not significantly different from the MTC obtained during the swing phase of the right foot each subject with at least five gait events where MTC (whether from the left or right foot) was available, was considered for processing.

After processing and calculation, the MTC is expressed as the mean MTC value (MeanMTC), the median MTC values (MedMTC), the minimal MTC values (MinMTC), the standard deviation of MTC values (SDMTC), the interquartile range of MTC values (IQRMTC), the variance of MTC values (VarMTC) and the coefficient of variation of the MTC values (CVMTC). The delta1 MTC = the maximal value of all MTC of the same subjects – MeanMTC (expressed in mm). The delta2 MTC = Mean MTC – Min MTC.

4.5. Additional gait patterns

According to Montero-Odasso et al.(Montero-Odasso, Sarquis-Adamson et al. 2017), and in order to assess the walking profile changes occurring between the comfortable walking condition and the dual task walking condition, the "DTW cost" was calculated for each gait parameter as follows: dual task cost parameter = [(CW gait parameter – DTW gait parameter) / CW gait parameter] x100 (expressed in %) where a "positive value" means a higher gait parameter value during CW than in DTW and a "negative value" means a higher gait parameter value during DTW than CW. Indeed "a positive DTW cost value" for a gait parameter means that for this parameter, the performance during DTW is lower than during CW, and a "negative DTW cost value" means that the gait parameter performance increased during DTW. For example, a gait speed DTW cost = 33.2 % means that gait speed during DTW is 33.2 % lower than gait speed during CW. In the same way, a symmetry DTW cost = -23.5 % means that the symmetry in DTW increases by 23.5% compared to symmetry in CW.

Similarly, and in order to assess the walking profile changes occurring between the CW condition and the FW condition, the authors calculated "FW improvement". Indeed, the gait parameter FW improvement is calculated as follows: gait parameter FW improvement = [(FW gait parameter – CW gait parameter)/ CW gait parameter] x 100 (expressed in %), where a "positive value" means a higher gait parameter value during FW than during CW, and a "negative value" means a higher gait parameter value during CW than FW. For example, a "gait speed change in FW" = 33.2 % means that the subject increased their gait speed by 33.2 % in FW compared to their own gait speed in CW.

4.6. Brain MRI

In order to investigate relationships between fall incidence, gait pattern and brain structure volumes, volunteers underwent structural brain MRI within a few weeks after their inclusion in the study. Acquisition sequences used involved multiparametric T1 weighted sequences, T2 weighted and Flair sequences and diffusion sequences.

Since the data from brain MRI exams are not shown in this work, brain MRI acquisition methods are not further detailed.

4.7. Neuropsychological evaluation

In order to examine the relationship between fall incidence, gait pattern and cognitive performances, volunteers were asked to undergo neuropsychological evaluation focusing on episodic memory, executive function, visuo-spatial abilities and attentional resources.

As the data relative to neuropsychological evaluation are not shown in this work, the evaluation methods are not further detailed.

4.8. Follow-up

During follow-up, we recorded fall(s), missteps, ADL and IADL dependencies, hospitalization, need for external help in daily life, institutionalization and death. At inclusion, all volunteers received a booklet explaining the goals of the study, the events to record, and their operational definitions. Finally, the names of the investigator and the department organizing the study and their contact details were indicated. The follow-up booklet and its contents was systematically shown and detailed by the investigator to each participant. The investigator paid attention to use the same terms for standardized recommendations and comments. A fall was defined as an unexpected event in which the participant comes to rest on the ground, floor, or lower level (Lamb, Jørstad-Stein et al. 2005). Regarding the potential consequences of only one fall, no distinction was made in this project between volunteers who experienced a single fall and those who experienced more than one fall. Furthermore, we chose to consider the term “misstep” (regardless of whether it was a slip or a trip) as a gait event to be recorded during follow up. A misstep was defined as a trip, slip, or other loss of balance in which recovery occurred to prevent a fall (Srygley, Herman et al. 2009).

Follow-up included phone contact every three months in order to record fall(s) history. The volunteers were asked to note every fall and to detail the fall circumstances. Every three months, an occupational therapist from the geriatrics department contacted each volunteer by phone to ask about fall(s) history, and fixed a date for the next phone contact to be sure the volunteer would be available. People who reported at least one fall during the follow-up period were considered as fallers.

4.9. Ethical considerations

The study was approved by the university's ethics review board and all participants signed a consent form prior to participating. The papers relative to ethical considerations are available in the appendices section (appendix 2).

4.10. Statistical analyses

Statistical analyses were performed using SAS version 9.4 (SAS Institute Inc., Cary, NC, USA) and MATLAB R2013a (Math Works, Natick, MA). Quantitative parameters are expressed as mean ± standard deviation (SD), 25th percentile (Q1), median, 75th percentile (Q3). Qualitative parameters are expressed as number (percentage). Normality was tested using the Shapiro-Wilk test and by investigating mean and median values, histograms and Quantile-Quantile plots. Homoscedasticity was tested using the Levenne test. Quantitative values were compared by one-way analysis of Variance (ANOVA) or by the Kruskal-Wallis test depending on the normality of their distribution. The association between qualitative parameters was tested using Chi square or Fisher's exact test as appropriate. Results were considered significant for a p-value <0.05, except for the comparison between the subjects who completed follow-up and those who dropped out, where a p-value < 0.001 was considered significant.

The correlation between gait parameters and right leg length or within the gait parameters was tested by Pearson's or Spearman's correlation coefficient according to the linearity of their distribution. Normalisation for right leg length was performed as "normalized parameter = parameter/right leg length (m)" for gait parameters correlated to the right leg length and showing significant differences according to fall incidence.

Interrelation between gait parameters showing a p-value < 0.15 after univariate analysis was tested by correlation analysis of Pearson or Spearman according the symmetry of their distribution. According the number of variables to consider (regarding the number of events), logistic regression was performed including the selected relevant variables.

According to their distribution, the IADL scores were categorized before logistic regression analysis. Indeed most people were autonomous for instrumental activities, while few were

dependant for one or two activities. Furthermore and as previously explained, the IADL score was calculated as the sum of the scores obtained on the items applicable to each subject, divided by the sum of the maximum possible score on the applicable items. Regarding the results obtained, most of the IADLs were equal to 1 while other scores were (7/8= 0.875) or (6/7= 0.857) or (6/8= 0.75). The IADL scores were categorized as "0", meaning the participant was autonomous for the IADL (where the item was applicable); and "1" for participants dependant on at least one instrumental activity.

4.11. Data Mining

Classification trees differentiating fallers and non-fallers after two-year follow-up based on their gait pattern at inclusion were developed using the J48 classifier available on the free downloaded software WEKA (Waikato Environment for Knowledge Analysis) version 3.8.1. The Waikato Environment for Knowledge Analysis (WEKA) is free, open-source software for data mining. WEKA is a product of the University of Waikato (New Zealand) and was first implemented in its modern form in 1997. The software is written in the Java™ language. Weka is a collection of machine learning algorithms for data mining tasks. The algorithms can be applied directly to a dataset. Weka contains tools for data pre-processing, classification, regression, clustering, association rules, and visualization. Several free online courses that teach machine learning and data mining using Weka are available in the free tutorial (Abernethy 2010). The version of Weka used in this study is version 3.8.1. The J48 classification is a supervised learning algorithm, where the class of an instance in the training set is known. J48 uses the training data to construct the decision model for the instance to class assignment and uses the model to predict a class of the instances in the test data. In the case of this study, the classes were defined as “fallers during follow-up” (noted “1”) or “non-fallers during follow-up” (noted “0”). The model was developed using the cross-validation method where the default number of folds is 10. The data are subdivided into 10 segments for the first 10 algorithm runs. In turn, one of the segments is used for the test data and the remaining 9 segments are used as training data. On the last run, the entire dataset is used as training data. The algorithm output shows the average of 11 runs. By default, the setting evaluation options involve a confidence factor of 0.25, a pruned option, and a minimum number of instances that should fall in each leaf node of the classification tree = 2.

At the end of the process the model obtained is shown as a classification tree and the performance is detailed. Using the J48, several performance indices were available. However, based on their clinical relevance, the following performance measurements were noted: accuracy (the number of correctly classified instances expressed in percentage); sensitivity, which measures the proportion of positives (the fallers in this study) that are correctly identified as fallers (also called the true positive rate or the recall); specificity, which measures the proportion of negatives (non-fallers) that are correctly identified as non-fallers (also called the true negative rate); positive predictive value (PPV), which is the proportion of positive results that are true positives; negative predictive value (NPV), which is the proportion of negative results that are true negatives; the ROC AUC value (Receiver Operating Characteristic Area Under the Curve) value, also called ROC area, which ranges from 0.5 (meaning discriminatory ability between fallers and non-fallers is similar to that due to chance) to 1 (meaning a perfect discriminatory capacity), and the PR AUC (Precision Recall Area Under the Curve). The ROC curve plots the true positive rate against the false positive rate, while a PR curve plots precision against recall. In case of an imbalanced data set, where the true negatives are more numerous than true positives, the use of the PR curve is recommended (Davis Jesse and 2006). Last but not least, for each classification node leading to a classification as “fallers” or “non-fallers”, the model gives two numbers in brackets; the first number is the total number of instances (weight of instances) reaching the leaf. The second number is the number (weight) of those instances that are misclassified. In case of a missing attribute value; the model will end up with fractional instances at the leaf. When splitting on an attribute, where some of the training instances have missing values, J48 will divide a training instance with a missing value for the split attribute up into fractional parts proportional to the frequencies of the observed non-missing values. This is discussed in the Witten & Frank Data Mining book as well as Ross Quinlan's original publications on C4.5.

In order to obtain the model with the highest accuracy, the authors proceeded step by step. For each model, the authors showed the accuracy, sensitivity, specificity, PPV, NPV, the ROC AUC and the area under the Precision-Recall Curve (PRC). The performance of each model was individually considered and compared to that of previous models in order to consider the usefulness of the variables included in the model. The first three models involved all gait parameters according to the walking condition in which they were obtained. The first model included the gait parameters obtained in CW, the second model included those obtained in FW and the third model included the gait parameters obtained in DTW. The fourth model included the calculated "FW improvement" for each gait parameter. The fifth included the calculated "DTW cost" for each gait parameter. In the sixth model, we included all the gait parameters obtained in CW, their "DTW cost" and their "FW improvement". At this time, considering the large volume of data involved and to increase the accuracy of the model, the unnecessary attributes were removed. Indeed, in order to select the most informative instances to retain in the model, the authors performed step-by-step manual selection. Considering the accuracy of the "full version" of the model (including all instances), the first instance of the model was removed and model performance was measured again and compared to the performance of the "full version". If the accuracy of the reduced model was higher than the accuracy of the "full version", the gait parameter was definitively left out of the model and the same process was applied to the next gait parameter. If the accuracy of the reduced model was lower than that of the full model, the authors considered this instance as "informative" and kept this instance in the model, and so on for each gait parameter. In the end, the "short version" of the model included only the "informative" gait parameters. Following this process, the sixth model included a "full version" and a "short version". The seventh model included all gait parameters, whatever the walking condition they were obtained in (CW, FW or DTW) and also the "DTW cost" and the "FW improvement". A full version and a short version of the seventh model are

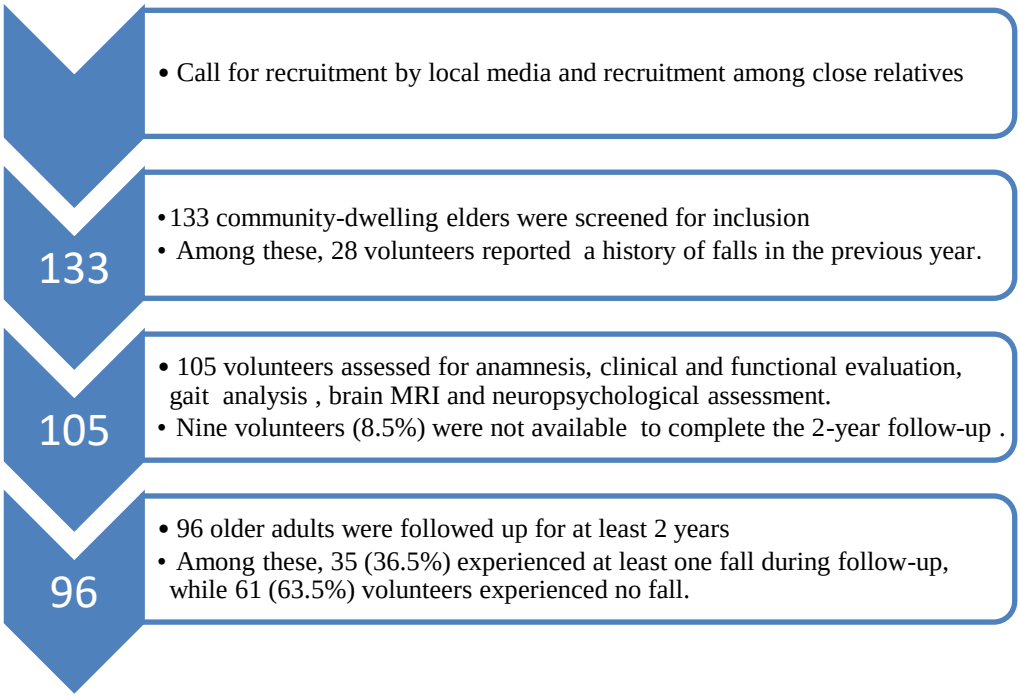
presented. The eighth model also included all gait parameters, whatever the walking condition they were obtained in (CW, FW or DTW) and their “DTW cost” and their “FW improvement” AND age, gender and the right leg length of each subject. A full version and a short version of the eighth model are presented. The ninth model included all gait parameters whatever the walking condition they were obtained in (CW, FW or DTW) and their “DTW cost” and their “FW improvement” AND the SPPB score, the IADL score and the rigidity scored using the UPDRS scale of each subject. A full version and a short version of the ninth model are presented. The tenth model includes the same variables as the ninth model (all gait parameters whatever the walking condition they were obtained in, their DTW cost and FW improvement, the SPPB score, the IADL score and the rigidity scored using the UPDRS scale of each subjects) AND age, gender and right leg length. A full version and a short version of the tenth model are presented.

5. Results

This section will detail the participants included and followed up, the comparison between fallers and non-fallers, the results of the logistic regression analysis addressing the prospective relationship between clinical data or gait parameters and fall(s) during follow-up. Finally, this section will show the classification models obtained using the J48.

5.1. Study participants

One hundred and thirty three volunteers were screened for eligibility between July 2014 and October 2015. Among these, twenty eight (21%) had a history of falls in the previous year, and thus, a total of one hundred and five subjects free of recent fall history were included and considered for the final analyses shown in this work. Two year follow-up was available for 96 subjects (91.5%).



The socio-demographic, clinical and functional characteristics of the 105 participants included in the study are shown in Tables 1 and 2.

During the two-year follow-up period, nine volunteers were censored: 1 died, 1 developed a neoplasm, 1 was diagnosed with dementia at inclusion, 1 was admitted to a nursing home; two volunteers did not leave the study but could not be contacted by telephone; 2 volunteers moved house and 1 person was excluded from follow-up because she fell close to stroke. A comparison of the clinical and functional characteristics of the study participants and those who dropped out is shown in Tables 3 and 4. Briefly, subjects who dropped out reported higher pain scores than people who completed follow up.

Among 96 participants, 48 were women and 48 were men, mean age was 71.3 ± 5.4 years (range 65 to 89 years). Overall, participants were well educated (the average duration of education was 13.03 ± 3.55 years), with low co-morbidities (mean CIRS score 9.42 ± 0.49) and were taking few medications (71% were taking fewer than 5 drugs per day). The vast majority were not frail (90%; mean GFTS and Edmonton score 1.99 ± 1.46), had no nutritional disorder

(mean MNA score 12.81 ± 1.72), most had no mood disorders (15% were depressed, mean GDS score 0.74 ± 0.97) or cognitive disorders (mean MoCA score 26.89 ± 2.75). Fifty percent reported memory complaints, 25% reported walking complaints and 35% reported a fear of falling, while 95% reported feeling well, and 90% were satisfied with their overall quality of life. Clinical assessment confirmed that the participants were independent in the activities of daily living (mean ADL score 6.21 ± 0.41; 87% lived at home without any help), were not frail and performed the functional tests with satisfying results (90% performed the Timed Up and Go test in less than 11 seconds, mean FGA score 26.8 ± 2.92, mean SPPB score 10.41 ± 1.57).

Furthermore, in terms of physical and functional performance, volunteers were non-sarcopenic, and had higher muscle mass, greater muscle strength and better muscle function than the consensual cut-off values used to define sarcopenia. Indeed, the mean skeletal muscle index measured in our sample was 38.0 ± 5.2 in men and 33.3 ± 5.2 in women and thus higher than the threshold of Janssen et al. (Janssen, Heymsfield et al. 2002). Furthermore, the mean grip strength assessed with Martin's Vigorimeter was 72.4 ± 16.1 kPa in men and 51.9 ± 15.9 in women, and was also higher than the threshold values published by Bautmans et al. (even after considering age categories) (Bautmans, Van Puyvelde et al. 2009). Finally, the mean gait speed was 1.29 ± 0.18 m/s, which is higher than the cut-off (≤ 0.8 m/s) used in the definition of sarcopenia issued by the European Working Group on Sarcopenia in Older People (EWGSOP) (Cruz-Jentoft, Baeyens et al. 2010).

5.2. Fall incidence

Among the 96 participants who were followed for at least two years, 35 (36.4%) fell at least once during the follow-up period.

5.3. Univariate analysis of factors related to falls

5.3.1. Comparison of socio-demographic, anamnestic, clinical and functional data

During the two years of follow-up, 18 men and 17 women experienced a fall (p-value = 0.83). Age at inclusion was not different between fallers (72.0 ± 6.9 years) and non-fallers (70.9 ± 4.3 years; p = 0.94). The fallers group included more subjects who lived alone (p = 0.01). There was no difference between groups in terms of tobacco or alcohol consumption; fallers reported 7.8 ± 12.7 pack-years vs 9.0 ± 14.1 pack-years in non-fallers (p= 0.78) and 5 participants among the non-fallers (8.2 %) reported drinking at least 4 doses of alcohol per day compared to 4 among fallers (11.4 %) (p = 0.72). Finally, the daily-life environment and the need for help were also similar between groups: among non-fallers, 54 people had no help, 6 had occasional help and 1 had regular help for housework, while among fallers, 30 had no help, 4 had occasional help and 1 person needed regular help for housework, (p = 1.00).

The medical and surgical data (except CIRS-G score) as well as the usual treatments in both groups are presented in Table 5. All medical data were similar between groups. CIRS score was 9.5 ± 4.8 in non-fallers and 9.3 ± 3.9 in fallers (p=0.82).

Anamnestic data concerning self-reported quality of life are shown in Table 6. There were no significant differences between fallers and non-fallers.

The comparison of clinical and functional characteristics measured at inclusion is shown in Table 7, according to the occurrence of a fall during follow-up. Briefly, both groups were similar.

The table below focuses on variables showing a significant difference according to fall occurrence; there was a statistically significant difference in IADL score, stiffness as assessed with the UPDRS scale and physical performance obtained on the SPPB test.

Variable	Group (n)	Mean ± SD	Q1	Median	Q3	p-value
IADL (score /1)	All (96)	0.99 ± 0.04	1.00	1.00	1.00	
	Non-fallers (61)	**1.00 ± 0.02**	**1.00**	**1.00**	**1.00**	**0.014**
	Fallers (35)	**0.98 ± 0.06**	**1.00**	**1.00**	**1.00**	
Stiffness (score /15)	All (94)	0.62 ± 1.32	0.00	0.00	1.00	
	Non-fallers (60)	**0.45 ± 1.25**	**0.00**	**0.00**	**0.00**	**0.043**
	Fallers (34)	**0.91 ± 1.40**	**0.00**	**0.00**	**2.00**	
SPPB (score /12)	All (96)	10.41 ± 1.57	10.00	11.00	12.00	
	Non-fallers (61)	**10.70 ± 1.42**	**10.00**	**11.00**	**12.00**	**0.015**
	Fallers (35)	**9.89 ± 1.71**	**9.00**	**10.00**	**11.00**	

5.3.2. Comparison of gait parameters obtained from the accelerometric method

Table 8 shows the comparison of gait parameters obtained from the accelerometric method according to fall incidence. Table 9 shows the comparison of the FW and DTW changes calculated for these gait parameters according to fall incidence. The reduced table below shows the statistically significant differences. Briefly, fallers had a significantly lower gait speed in FW and a shorter stride length in CW and in FW than non-fallers. Furthermore, fallers decreased their stride symmetry in DTW whereas non-fallers increased their stride symmetry in DTW.

Variables	Groups	Mean ± SD	Q1	Median	Q3	p-value
FW Gait Speed (m/s)	All (96)	1.70 ± 0.23	1.52	1.72	1.86	
	Non-fallers (61)	**1.74 ± 0.22**	**1.59**	**1.75**	**1.87**	**0.035**
	Fallers (35)	**1.64 ± 0.24**	**1.43**	**1.68**	**1.85**	
CW Stride Length (m)	All (96)	1.34 ± 0.16	1.23	1.37	1.45	
	Non-fallers (61)	**1.37 ± 0.15**	**1.26**	**1.38**	**1.48**	**0.035**
	Fallers (35)	**1.30 ± 0.17**	**1.16**	**1.27**	**1.45**	
FW Stride Length (m)	All (96)	1.55 ± 0.24	1.38	1.58	1.71	
	Non-fallers (61)	**1.60 ± 0.24**	**1.41**	**1.62**	**1.73**	**0.010**
	Fallers (35)	**1.47 ± 0.23**	**1.28**	**1.42**	**1.66**	
Symmetry DTW cost (%)	All (n=95)	-5.93 ± 36.62	-28.11	1.43	22.31	
	Non-fallers (n=60)	**-13.61 ± 40.54**	**-41.56**	**-11.83**	**18.44**	**0.022**
	Fallers (n=35)	**7.24 ± 23.94**	**-15.42**	**7.32**	**26.07**	

In view of the correlation between gait speed and stride length, and right leg length, a comparison of normalized gait parameters was performed according to fall incidence. As shown below, after normalization for the right leg length, FW gait speed and CW stride length were similar in both groups, whereas FW stride length remained significantly different. Participants who experienced a fall during follow-up had a shorter FW stride length than those who did not fall. No normalisation was applied to the DTW symmetry cost.

Normalized gait parameters	**Groups (n)**	**Mean ± SD**	**Q1**	**Median**	**Q3**	**p-value**
FW Gait speed normalized for right leg length (/s)	All (n=96)	2.02 ± 0.26	1.80	2.02	2.20	
	Non-fallers (61)	2.04 ± 0.26	1.87	2.05	2.19	0.19
	Fallers (35)	1.97 ± 0.27	1.75	1.93	2.21	
CW Stride length normalized for right leg length (m)	All (96)	1.59 ± 0.16	1.48	1.59	1.68	
	Non-fallers (61)	1.60 ± 0.15	1.49	1.61	1.68	0.18
	Fallers (35)	1.56 ± 0.17	1.44	1.58	1.68	
FW Stride length normalized for right leg length (m)	All (96)	1.84 ± 0.27	1.64	1.81	2.00	
	Non-fallers (61)	**1.88 ± 0.28**	**1.70**	**1.87**	**2.00**	**0.046**
	Fallers (35)	**1.77 ± 0.24**	**1.61**	**1.75**	**1.95**	

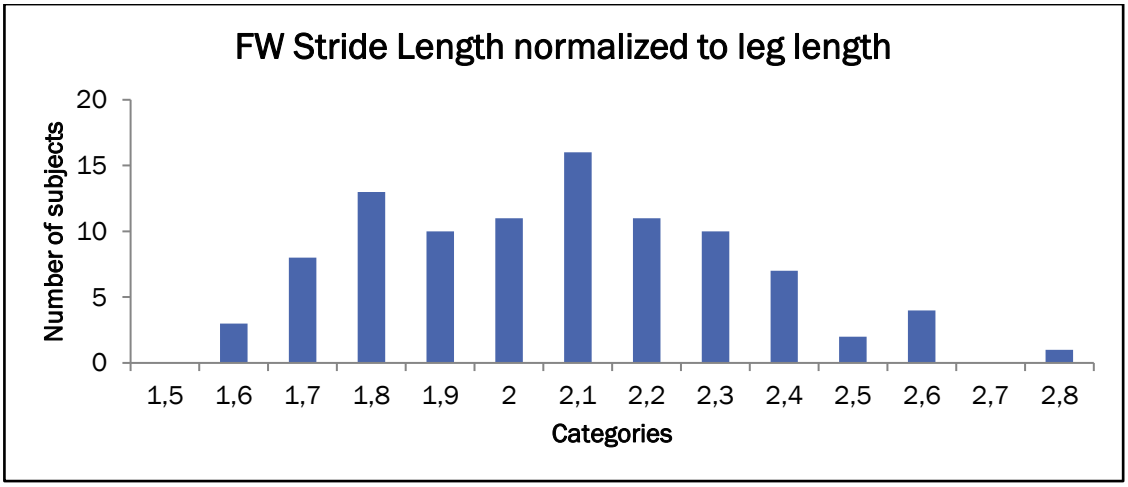
In order to better visualize the difference between groups according to gait performance, figures 1 and 2 show the distribution histograms and box plot of the FW stride length normalized for right leg length, and the stride symmetry DTW cost respectively. Considering their distribution, means and medians, the two variables were considered as having a normal distribution.

In figure 2, the blue line indicates zero stride symmetry DTW cost (0%) to highlight that the median stride symmetry DTW cost of non-fallers is below the blue line, meaning the non-fallers have a negative stride symmetry DTW cost, i.e. they increase their stride symmetry in DTW compared to CW. Conversely, the median stride symmetry DTW cost of fallers is above the blue line, indicating that fallers have a positive stride symmetry DTW cost (i.e. they decrease their stride symmetry in DTW compared to CW).

Figure 1:

Distribution histogram of the FW Stride Length normalized for leg length

And box plot of the same variable according to the falls during the follow-up



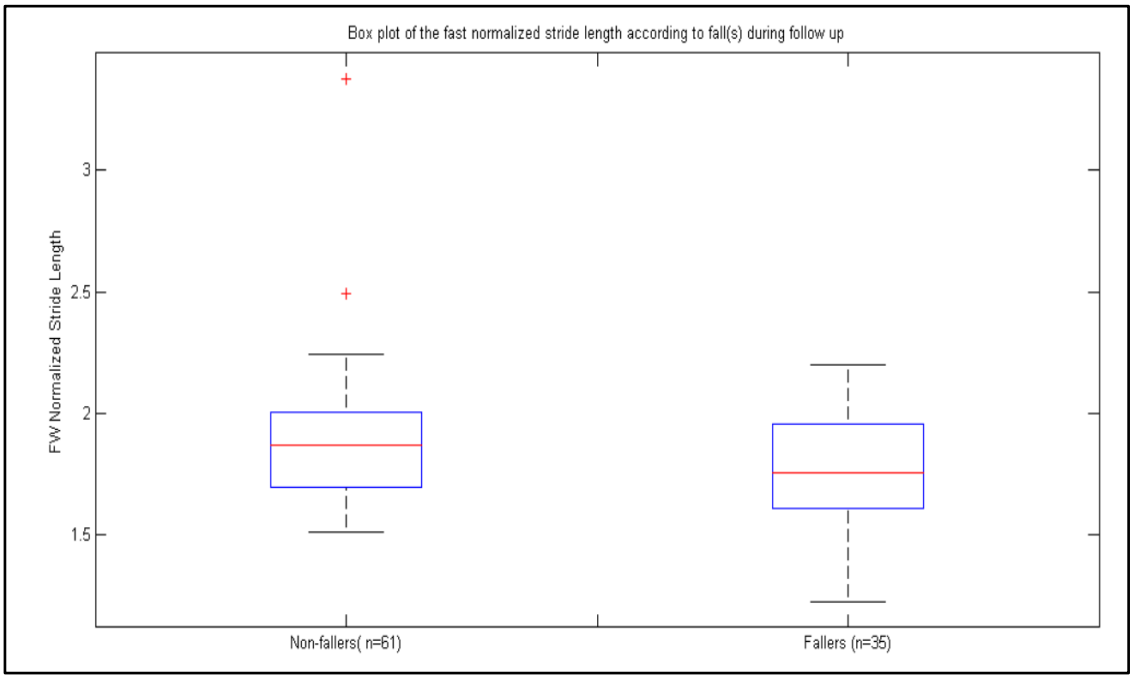
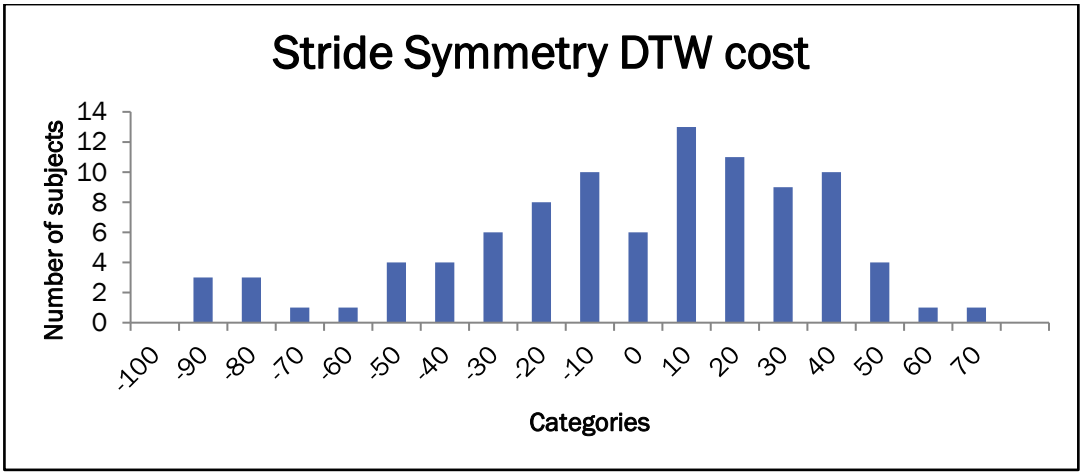
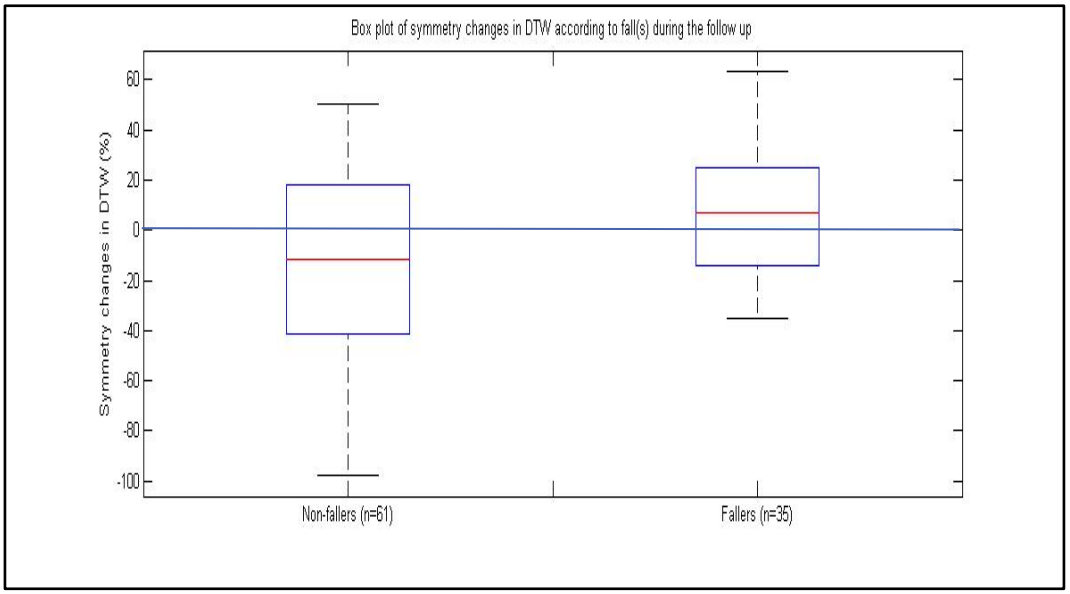


Figure 2:

Distribution histogram of the Stride Symmetry DTW cost

And box plot of the same variable according to the falls during the follow-up





5.3.3. Comparison of gait parameters obtained from the opto-electronic method

Table 10 shows the comparison of all gait parameters obtained from the opto-electronic method in CW, FW and DTW. All parameters were similar between the two groups. Although the differences were not statistically significant, some trends were apparent. Mean MTC values (Mean MTC), median MTC values (Med MTC) and minimal MTC values (Min MTC) were lower in fallers than in non-fallers, whatever the walking condition. The parameters measuring intra-subject variability of the MTC, such as the standard deviation of MTC values (SD MTC), the interquartile range of MTC (IQR MTC), the variance of MTC (Var. MTC) and the coefficient of variation of the MTC (CV MTC) were similar in the two groups.

Changes in MTC during FW and DTW are shown in Table 11. All parameters were similar between groups. Most fallers and non-fallers had an increase in Mean MTC, a decrease in Min MTC and an increase in CV MTC in FW. Conversely in DTW, most fallers and non-fallers had a decrease in Mean MTC, an increase in Min MTC and a decrease in CV MTC.

5.4. Correlation between the right leg length and gait parameters recorded in CW

The gait parameters recorded in CW that were significantly correlated with right leg length were: gait speed ($p = 0.0009$, $r = 0.33$), stride length ($p < 0.0001$, $r = 0.56$), stride frequency ($p = 0.0071$, $r = -0.27$), stride regularity, ($p = 0.039$, $r = 0.21$), Mean MTC ($p = 0.014$, $r = 0.26$), Med MTC ($p = 0.022$, $r = 0.24$), Min MTC ($p = 0.0005$, $r = 0.36$) and CV MTC ($p = 0.0002$, $r = -0.39$).

5.5. Logistic regression analysis according to fall(s) during follow-up

The variables included in the logistic regression analysis were the symmetry DTW cost, FW stride length normalized to the leg length, stiffness according to the UPDRS scale, SPPB score and categorized IADL score.

Ninety-three observations were used for regression analysis; three observations were not used due to missing values for explanatory variables. The number of events was 34 (34 fallers and 59 non-fallers).

After logistic regression analysis, the symmetry DTW cost (unit measure in percent) was shown to be significantly related to the risk of falls, with an odds ratio (OR) = 1.018 (95% confidence interval (CI) 1.002-1.033), p-value = 0.027.

5.6. The use of WEKA

The performances of all models obtained are shown in Table 12.

As shown in Table 12, models 1 to 6 had mediocre to low performances. Accordingly, they were processed only to identify the variables to include in the more complete models. The overall performance of models 1 to 6 is shown and but the classification trees yielded by these models are not detailed. However, the findings from these models concerning the relationship between gait parameters or walking conditions and model performance will be discussed later.

As shown below, models 1, 2 and 3 were processed successively; respectively involving all gait parameters obtained in comfortable walking condition (CW), fast walking condition (FW) and dual task walking condition (DTW). Models 1 to 3 showed poor accuracy, low sensitivity and good to excellent specificity. The second model involving the gait parameters obtained in FW yielded the best performance.

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
CW Model 1	56%	3%	87%	11%	61%	0.42	0.49
FW Model 2	70%	29%	93%	71%	70%	0.52	0.56
DTW Model 3	62%	17%	88%	46%	65%	0.60	0.62

The performance of models 4 to 6 is shown in the table below.

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
FW improvement = Model 4	58%	3%	90%	14%	62%	0.42	0.49
DTW cost = Model 5	59%	34%	74%	43%	66%	0.57	0.60
FW + DTW cost = Model 6 full version	73%	37%	93%	76%	72%	0.67	0.71
Model 6 short version	76%	22%	92%	77%	76%	0.64	0.65

The fourth model including the FW improvement in gait parameters showed very poor sensitivity and high specificity. The fifth model included the DTW cost of gait parameters and showed similar accuracy, higher sensitivity and lower specificity than the fourth model. The full version of the sixth model, which included FW gait parameters and DTW cost gait parameters, showed better accuracy, higher sensitivity and specificity, higher positive and negative predictive values than the previous models. The short version of the sixth model showed the highest accuracy, PPV and NPV of all six models; however, the sensitivity and specificity of the short version of the sixth model were lower than those of the second model.

Models 7 to 10 will be further shown in an “original version” classification tree (i.e. the classification tree directly obtained from WEKA), an “enhanced version” classification tree (the original version classification tree modified to show gait parameter names and misclassification rates, see below) and a “reader-friendly version” classification tree, which makes the classification tree easier to read and use. Based on their usefulness for obtaining the final model, models 7 and 8 will be described but not detailed. In view of the high performance of models 9 and 10, these models will be described and further detailed.

Two points should be noted regarding the classification nodes. First, for each node leading to a classification into fallers or non-fallers, the classification tree gives the number of participants classified (the first number in brackets). In case of two numbers in the brackets, the second number corresponds to the number of people incorrectly classified (i.e. total number of classified people/ number of misclassified people). This data is retained in the full text of this section. Numbers that are not integers mean that the gait parameter is not available for all the people classified in this class. Secondly, in case of “negative” DTW cost values, the names of the classification nodes were changed to make them easier to understand. For example, and as in the first classification node, a “negative cost of symmetry” has been rephrased as an “increase in symmetry” and the arithmetic relationship has been inverted for easier reading and interpretation.

The seventh model included all gait parameters (whatever the walking condition in which they were obtained) and their changes during FW and DTW. The performances of the full and short versions are shown below. The “original”, “enhanced” and “reader friendly” versions of the classification tree obtained from the short version of Model 7 are respectively shown in Figures 3, 4 and 5.

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 7 full version	61%	54%	66%	48%	71%	0.66	0.66
Model 7 short version	80%	69%	87%	75%	83%	0.79	0.77

The performances of the short version are higher than those of the full version of the seventh model. Furthermore, the performances of the seventh model short version are higher than those of the sixth model short version.

The eighth model included all gait parameters (whatever the walking condition in which they were obtained), their FW and DTW changes as well as age, gender and right leg length of the volunteers. The “original”, “enhanced” and “reader friendly” versions of the classification tree obtained from model 8 short version are respectively shown in Figures 6, 7 and 8. The performances of the full and short versions of model 8 are shown below.

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 8 full version	61%	57%	54%	47%	72%	0.66	0.65
Model 8 short version	76%	51%	90%	75%	76%	0.76	0.76

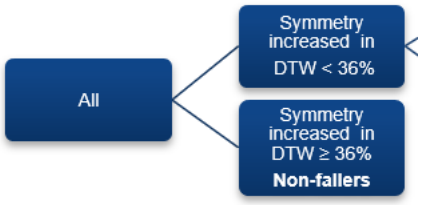
The performances of the short version of the model 8 were better than those of the full version of model 8. The specificity of the short version of model 8 was higher that of the short version of model 7, whereas the accuracy, sensitivity and negative predictive value of the short version of model 8 were lower than those of the short version of model 7.

The ninth model included gait parameters obtained in all walking conditions, their FW and DTW changes and the clinical or functional variables that were significantly different between fallers and non-fallers by univariate analysis (i.e. IADL score, SPPB score and stiffness scored using the UPDRS). The "original", "enhanced" and "reader friendly" versions of the classification tree obtained from model 9 short version are shown in Figures 9, 10 and 11 respectively. The performances of the full and short versions of model 9 are shown below.

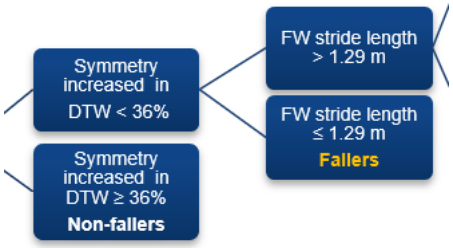
Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 9 full version	59%	46%	67%	44%	68%	0.57	0.59
Model 9 short version	82%	69%	90%	80%	83%	0.80	0.79

The performances of the short version were better than those of the full version. All the performances of model 9 were better than those of model 8 (except for specificity, which was similar in both models). Similarly, all the performances of model 9 were better than those of model 7 (except for sensitivity, which was similar in both models).

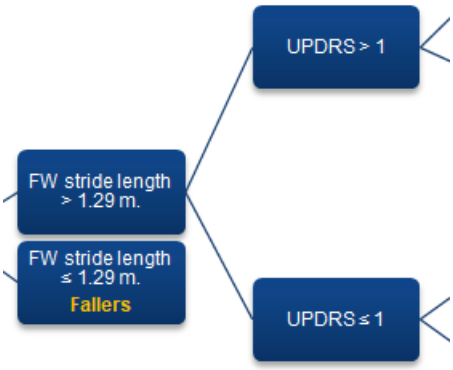
As explained before, model 9 will be detailed using short extracts showing the classification nodes and threshold values used by J48 to classify subjects into fallers or non-fallers.



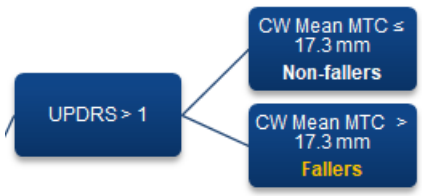
The first extract shows that the first classification node used by J48 is the symmetry DTW cost ("increased" when there is a "negative cost value"). According to this classification node, participants who present a symmetry increase ≥ 36% in DTW are classified as non-fallers. Furthermore, as shown in the "enhanced" version of the classification tree (Figure 10), no participant was misclassified at this step (17.18). As used in the first classification node, the percentage increase in symmetry in DTW makes it possible to identify non-fallers.



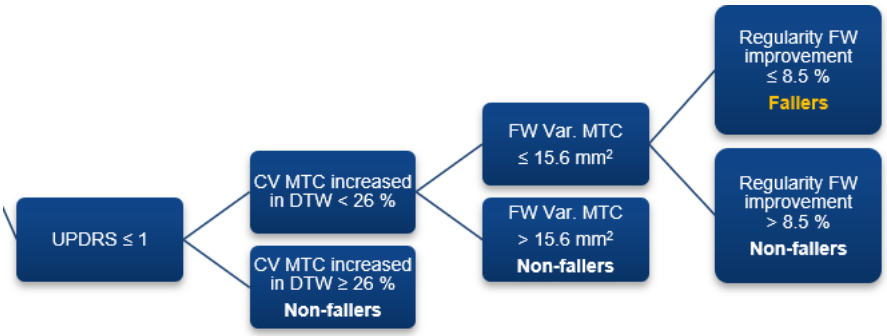
The second classification node shows that subjects who do not have an increase in symmetry ≥ 36% are considered for their FW stride length. Indeed, those with a FW stride length ≤ 1.29 m are classified as fallers, while those who have a FW stride length > 1.29 m will be considered in the next classification node. The "enhanced version" of the model shows that participants are classified as fallers at this stage, without misclassification (11.0). As used in the second classification node, the FW stride length makes it possible to identify fallers.



The next node evaluates stiffness; the model differentiates subjects without stiffness (≤ 1 according the UPDRS scale) from subjects with stiffness (> 1 according the UPDRS scale). This node does not allow any classification of participants.



Among those with stiffness, those who have a CW Mean MTC > 17.3 mm are classified as fallers, without misclassification (7.58). The subjects who have a CW Mean MTC ≤ 17.3 mm are classified as non-fallers, albeit with some misclassification, meaning that fallers were under-recognized and incorrectly classified as non-fallers by the model (5.42/0.42). According to the following extract, among people without stiffness, subjects who increased the coefficient of variation of MTC in DTW ≥ 26 % (negative cost in DTW in the original version of the model) are classified as non-fallers (17.2/0.63), while those who did not increase the same parameter were considered for the variance of MTC in FW.



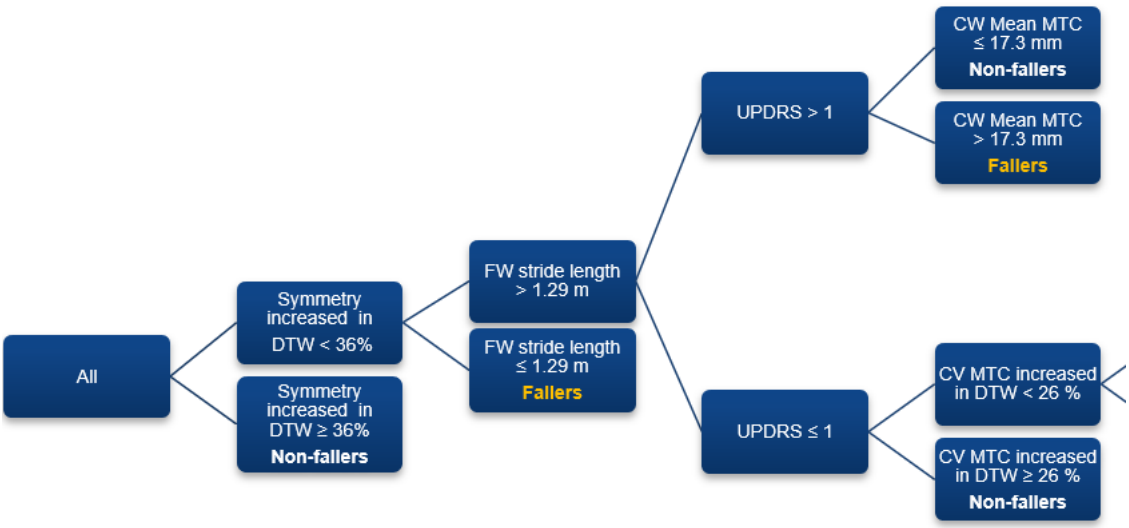
Indeed, people without stiffness, without an increase of ≥ 26 % in CV MTC in DTW and without an MTC Variance > 15.6 mm^2 in FW are considered for the percentage of regularity improvement in FW; those who have an improvement in regularity > 8.5 % in FW are classified as non-fallers (5.24/1.0), while those who do have an improvement in regularity ≤ 8.5 % in FW are classified as fallers (14.4/2.36). Of note, the last three classification nodes involving FW MTV variance and regularity FW improvement showed misclassification (see enhanced version of the ninth model, Figure 10).

The tenth model included gait parameters obtained in all walking conditions, their FW and DTW changes, IADL score, SPPB score and stiffness scored using the UPDRS as well as age, gender and right leg length of the volunteers. The "original", "enhanced" and "reader friendly" versions of the classification tree obtained from model 10 short version are shown in Figures 12, 13 and 14 respectively. The performances of the full and the sort versions of the tenth model are shown below.

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 10 full version	69%	46%	67%	4%	68%	0.58	0.59
Model 10 short version	84%	80%	87%	78%	88%	0.84	0.83

The performances of the short version of model 10 were better than those of the full version. Overall, the performances of the short version of model 10 were better than those of the previous models. Indeed, the performances of model 10 were better than those of model 7, model 8 (except for specificity), and model 9 (except for specificity and positive predictive value). As for model 9, model 10 will be detailed using short extracts showing the classification nodes and threshold values used by J48 to classify subjects into fallers or non-fallers. Moreover, the two models will be compared, and their differences discussed in the next section.

As shown in the extract below, the first nodes are similar to those of model 9.



The first four nodes of the classification tree involve the increase in symmetry in DTW, FW stride length, presence of stiffness, CW Mean MTC, and the increase in CV MTC in DTW. However, the inclusion of gender in this model changed the last nodes; two changes in particular were observed. The first change was that, rather than considering all subjects who had a variance of MTC in FW > 15.6 mm^2 as non-fallers (as in model 9), model 10 considered the variable delta1 MTC in DTW (where delta1 MTC = the maximal value of all MTC of the same subject – Mean MTC); subjects were classified as fallers when the delta 1 MTC in DTW was > 11.6 mm (2.35/0.31) while those with a delta 1 MTC in DTW ≤ 11.6 mm were classified as non-fallers (15.63/0.29). According to the enhanced version of the classification tree obtained from model 10, and displayed in Figure 13, the number of people classified as fallers based on the delta1 in DTW is similar to the number of people previously misclassified as "non-fallers" based on the FW MTC variance in model 9.

The second change concerns those who had a FW MTC variance ≤ 15.6 mm^2; indeed, rather than being considered in terms of the regularity of FW improvement (as in model 9), these subjects were considered in terms of FW Mean MTC. Subjects with FW Mean MTC ≤

18.5 mm were classified as fallers (8.29/0.45), while those with FW Mean MTC > 18.5 mm were subsequently considered for gender; women were classified as non-fallers (4.41/1.0), while men were considered in terms of the symmetry DTW cost: those with a symmetry DTW cost ≤ 22 % were classified as fallers (4.21/0.21) while those with a lower symmetry DTW cost were classified as non-fallers (2.73/0.21). As shown in Figure 13, the number of misclassifications for these last three classification nodes was less than 1.

The figure 15 and 16 show ROC are and PRC are related to model 10.

Figure 15: Receiver Operating Characteristic Curve

Where false positive rate is on x axis while true positive rate is on y axis

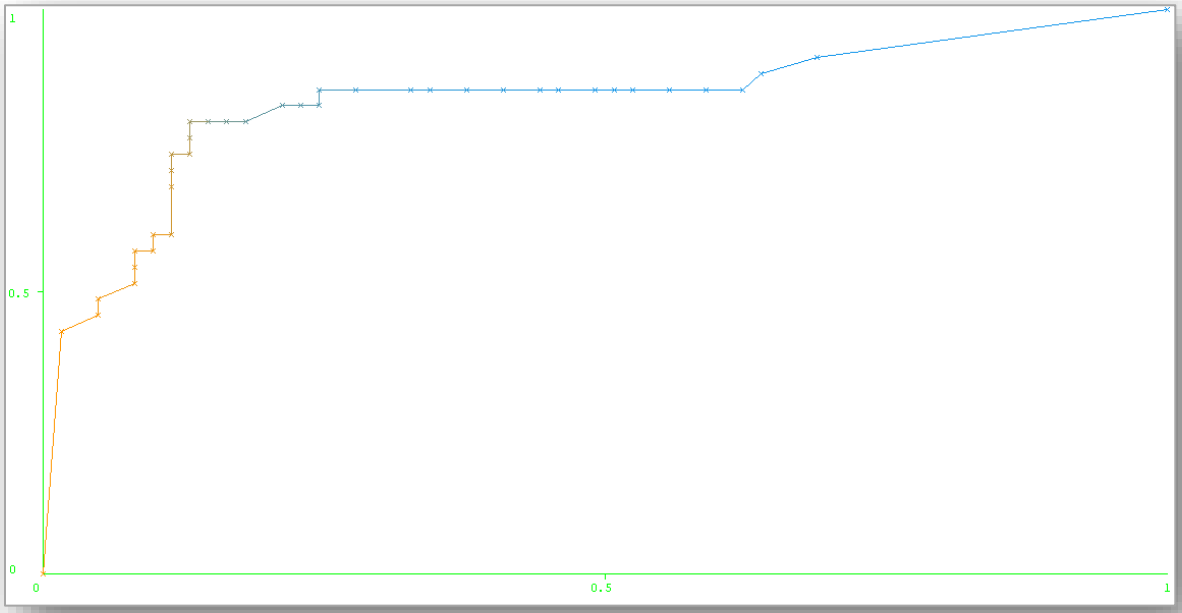
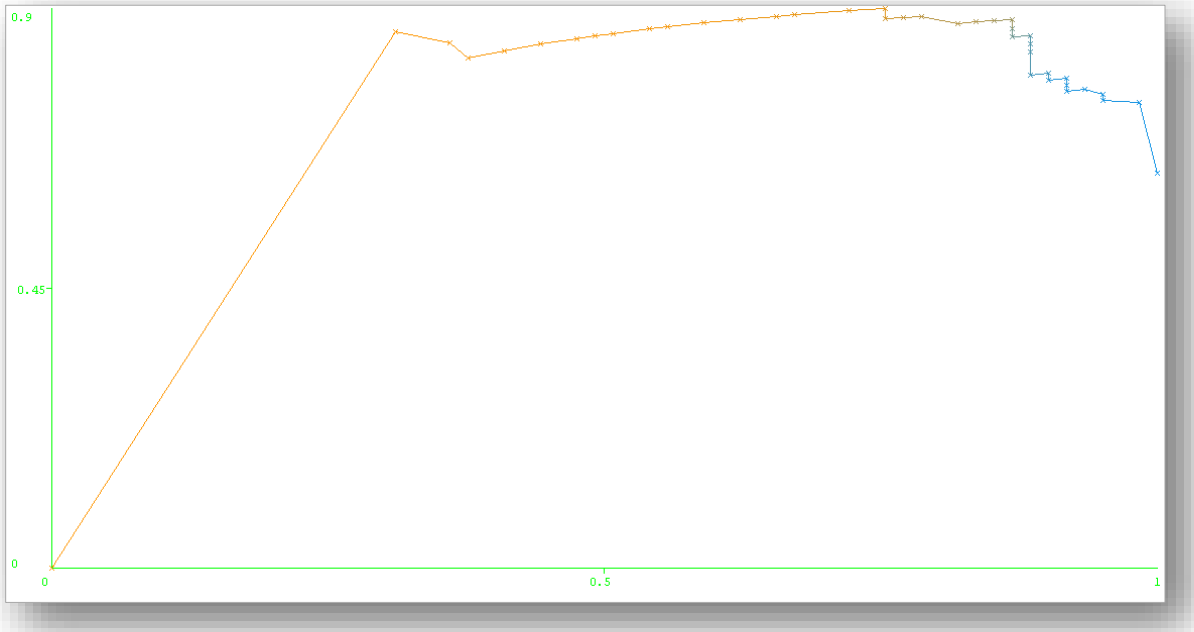


Figure 16: Precision Recall Curve

Were recall is on x axis while precision is on y axis



6. Discussion

The main goal of this exploratory prospective study was to investigate whether, among healthy older adults, the assessment of gait patterns could be useful to discern people at risk of future fall(s). After two years of follow-up, one third of the volunteers had fallen at least once. At inclusion, fallers had a significantly lower comfortable gait speed and shorter stride length in comfortable and fast walking conditions and higher symmetry DTW cost than non-fallers. After adjustment to the right leg length, the FW normalized stride length and the symmetry DTW cost remain different between fallers and non-fallers. Logistic regression analysis showed that higher stride symmetry DTW cost was significantly associated with a higher fall risk. Furthermore, the use of the J48 classifier showed that the symmetry DTW cost, the FW stride length, the stiffness and the MTC values (mean and variability measures) could be useful for the evaluation of the individual future fall risk. Used in this study, the data mining highlighted the usefulness of gait patterns to identify people at risk for future fall(s). Indeed, in this cohort, a classifier tool considering several gait parameters and clinical measures simultaneously made it possible to identify 80% of future fallers. This result is encouraging, and tends to confirm the utility of considering gait patterns for the assessment of fall risk among older adults.

In order to discuss the results obtained, the next section is dedicated to a discussion of: (1) the participants involved; (2) the fall incidence; (3) the relationship between anamnestic, clinical or functional data and future fall(s); (4) the relationship between gait parameters and future fall(s); (5) the classification trees obtained using WEKA; (6) the overall significance of the results obtained. Finally, the strengths and limitations of this exploratory study will be discussed.

Before discussing the results, a comment has to be made about the choice between a p-value < 0.001 and a p-value < 0.05 to be considered as statistically significant. Indeed, after ANOVA analyses, and considering the number of comparisons performed, we considered the need to apply Bonferroni correction. According to a recent review of the use of Bonferroni correction (Armstrong 2014), there is a lack of consensus about the need to apply correction in scientific papers, and the use of the Bonferroni method should be considered in light of the goals of the study, the number of comparisons planned, and the interpretation of the significance obtained. In our study, concerning the comparison between the participants who were followed up and those were excluded, and according to Armstrong's recommendations: "A Bonferroni correction should be considered if: (1) it is imperative to avoid a type I error. (2) a large number of tests are carried out without pre planned hypotheses in an attempt to establish any results that may be significant".

In view of these recommendations, when comparisons would assume the difference between groups is significant, we decided that a p-value < 0.001 should be used as the threshold to consider a difference as statistically significant.

However, concerning the functional data and the gait parameters, and again in line with the recommendations of Armstrong: "No correction would be advised in the following circumstances: if a study is exploratory involving post-hoc testing of unplanned comparisons which are regarded as hypotheses for further investigation", we therefore decided to accept a p-value < 0.05 as being statistically significant. Indeed, this study is an exploratory study where the comparative analyses will be followed by a logistic regression analysis exploring further the relationship between functional data/gait parameters and future fall(s).

6.1. Participants

One hundred thirty-three volunteers were considered for inclusion. Since the clinical endpoint considered in this part of the study was future fall(s), subjects with a recent history of falls were not considered for analysis to avoid bias. However, these 28 participants with a recent fall history were fully assessed using the same material and methods, and their performances will be useful to discuss the second clinical outcome (cognitive decline) in further analyses. Thus, although 133 participants were assessed in this longitudinal study, only the gait performances of the subjects without a recent fall history were considered in the results shown here.

As detailed in the results section and related tables (Tables 1-4), 96 participants were included in the present analysis, and followed-up for a period of 2 years. All were younger than 75 years (which is the age beyond which older adults are considered for geriatric care), independent, robust and non-sarcopenic, which may seem disappointing given that the study was evaluating fall risk in this population. However, in our opinion, this sample truly represents the people who should be targeted for fall prevention programs, to avoid the burden of negative consequences linked with falls (including the first fall). From an epidemiological point of view, the robustness of the sample could be the result of a selection bias. Indeed, recruitment was very quick among these well-educated, fit, independent and well-informed participants, who were also attentive to health related issues. Furthermore, the profile of our study population could also explain the low rate of censored.

The comparison of the participants censored with the participants who completed follow-up showed no significant differences. Thus, according the participants censored have similar profile to those who were considered for analysis, and therefore, the risk of obtaining significantly different results is likely to be small.

Unfortunately, the number of subjects included in this study and, consequently, the number of falls observed during follow-up was relatively low, thus limiting the power of the statistical analysis. Further studies with larger sample sizes are warranted to confirm our findings.

6.3. Fall incidence

During the two years of follow-up, 36% of participants experienced at least one fall. This rate is similar to that reported in the study by Verghese et al on fall risk in community-dwelling older adults, reporting a fall rate of 38% after a mean follow-up period of 20 months (Verghese, Holtzer et al. 2009). In the study by Srygley et al., 25.6% of the participants reported at least 1 fall during the one-year follow-up (Srygley, Herman et al. 2009). However, other prospective studies have shown different results. Laessoe et al. reported a fall incidence of 15% per year, but the authors underlined that the follow-up was not sufficient to ensure complete fall event collection (Laessoe, Hoeck et al. 2007). In contrast, Callisaya et al. reported a fall incidence of 45% over a one-year follow-up period. However, the characteristics of the participants included in both studies were quite similar. These different results support the idea that, even in similar participants and using same definition of the negative outcome, the varying fall rates are probably related to different levels or types of physical activity, and different daily-life environments, which were not taken into account in these studies.

6.4. The relationship between anamnestic, clinical or functional data and fall(s)

In our cohort, fallers and non-fallers were similar in terms of age, gender, medical history and drug treatments, habits and co-morbidities. The only socio-demographic factor that was statistically significant different was marital status. This observation warrants specific comment. Indeed, comparison showed that there were more single persons in the faller group compared to the non-faller group. However, I chose not to consider this socio-demographic variable in further analysis. Indeed, initially the question about marital status was not considered for the evaluation of the future fall(s) risk but rather for the purposes of describing the socio-demographic characteristics of the sample. Any discussion of the relationship between marital status and future fall(s) risk would require taking into account other relevant data related to the daily life, such as social support and mood, in order to more accurately consider the real impact of marital status on social and functional aspects. The following two examples, although they may seem exaggerated, illustrate the importance of consider not only marital status, but also other functional aspects. For example, is it correct to compare, based on their marital status, a man who lost his wife two months previously to disease, a woman who left her husband twenty years ago because he was a drunkard? In the same way, is it correct to compare, based on their marital status, a man living maritally who bears the responsibility for the household and cares for his partner at home because she suffers from dementia with anosognosia, with a woman who lives maritally with a dynamic and sporty man who enjoys participating in leisure activities and city trips? It is probably better to assess the links between marital status and fall risk with more comprehensive anamnesis. Thus, for single subjects, relevant questions might include: “For what reason are you living alone? Is it by choice?”, “How long have you been living alone?”, “How do you feel now that you are alone?”, “Do you

have better or lower quality of life?", "Have you more or fewer social relationships?", "Have you more or fewer physical activities?". For people living maritally, relevant questions could include "Are you mostly satisfied to be living maritally?", "Are you mostly happy to be living with your current partner?", "Who does the housework and the gardening?"… Unfortunately, because the marital status was initially not considered as potentially linked with future fall(s), the circumstances were not sufficiently documented and therefore, in view of this limitation, I chose to not include this variable either in the logistic regression or in the J48 classifier. Nonetheless, we cannot exclude the existence of an independent relation between marital status and fall risk, but the design of this study does not make it possible to answer this question; this would nevertheless be an interesting research topic to address in future research.

In terms of clinical characteristics and functional performances, the two groups were also similar except for IADL score, SPPB score and stiffness scored based on the UPDRS scale. However, even though the p-value was statistically significant (<0.05), the differences between fallers and non-fallers were not clinically relevant. As previously explained, and according to Armstrong, if the IADL, SPPB and UPDRS were to be used to significantly differentiate future fallers from other participants, then Bonferroni correction would be need to be applied (Armstrong 2014). In this scenario, considering a p-value < 0.001 as significant, then none of these variables would be considered as statistically significant. However, as this exploratory study includes logistic regression analysis, Bonferroni correction was not applied.

Concerning the relationships between anamnestic, clinical or functional data and fall incidence, in this study, fallers had a lower IADL score, corresponding to less autonomy in the instrumental activities of daily living, than non-fallers. This relationship between lower IADL score and fall risk was also suggested in the systematic review and meta-analysis (based on 3,747 indexed articles published between 1981 and 2007, of which 129 reported data on social-demographic risk factors for falls) (Bloch, Thibaud et al. 2010). According to this meta-

analysis, among community-dwelling elders living at home, impairment of one or more IADL was associated with a higher fall risk (OR 2.10, 95% CI 1.68-2.64). In our study, the IADL score was not associated with future fall risk after logistic regression. In our opinion, the absence of a significant association in this study does not mean that no association exists. Indeed, our results are probably due to a lack of power resulting from the relatively small number of people dependent for IADLs and the low number of falls.

Furthermore, in our study, subjects who went on to fall during the follow-up period of two years had a lower SPPB score at inclusion than those who did not fall. However, this link was no longer significant after multivariate analysis. Our results are similar to those of Quadri et al. who included 144 patients aged 70 or older (60 men and 84 women, mean age 78.7 ± 5.6 years) in a one-year observational study where fall(s) were one of the outcomes (Quadri, Tettamanti et al. 2005). As in our study, after univariate analysis, a lower score in SPPB was associated with higher fall incidence, but the relation did not remain significant after multivariate analysis. Similar results were also shown by Ward et al. who evaluated 755 Boston area community-dwelling adults (mean age 78.1 ± 5.4 years, 64.1% women) with or without fall history in a 4-year prospective study assessing injurious fall incidence (Ward, Leveille et al. 2015). In this study, the SPPB score was not predictive of injurious falls; however, the group with the poorest chair stand performance (≥16.7s) had the greatest hazard of injurious falls compared to any other groups. Unfortunately, the component scores of the SPPB were not recorded in our study, precluding comparison with these data. Substantial differences also limit the comparison between the two studies. First, the American study included fallers and people less physically robust than those included in our study, which could lead to substantial differences in the results; e.g. in the Ward' study, 32 subjects included (4.3%) had an SPPB score = 1–3/12, 68 (9.1%) a score = 4–6/ 12, 204 (27.3%) a score = 7–9 /12, and 443 (59.3%) a score = 10–12. Secondly, Ward's study considered injurious falls as defined by fractures,

sprains, dislocations; pulled or torn muscles, ligaments, or tendons; or by seeking medical attention, while falls were defined as an unintentionally coming to rest on the ground or another lower level not resulting from a major health event (e.g. myocardial infarction) or an overwhelming external hazard (e.g. vehicular accident), whereas our study considered accidental fall(s), meaning that syncopal falls or falls leading to a medical check-up were not considered in the follow-up). Finally, and given the importance of the individual components of the SPPB score in the American study (e.g. the chair stand performance), the usefulness of these components should be considered in future works involving robust older adults.

Finally, to the best of our knowledge, no similar study including robust community older subjects in a prospective study evaluating fall(s) incidence has assessed the possible relationship between stiffness, as measured using the UPDRS scale, and the risk of falls. Since the UPDRS scale is most often used in studies focusing on neurological disorders, cardiovascular disease or burden, it is unsurprising that it is not reported in studies involving robust older adults. However, considering that the stiffness accompanying high level gait disorders (Giladi, Herman et al. 2005) is associated with increasing age, it seems logical to measure it among the subjects involved in this study, which aimed to investigate whether gait patterns could help clinicians to detect people at risk of fall(s). In our cohort, after multivariate analysis, stiffness was not found to be related to the risk of falls. However, stiffness was used by the classifier tool as one of the earlier classification nodes. Indeed, when considered at an individual level, stiffness together with consideration of CW Mean MTC value helped to identify future fallers. In our opinion, these results support the idea that stiffness should be systematically considered in clinical assessment, even in the absence of neurological or cardiovascular disease or burden, in order to highlight emerging high level gait disorders.

To summarize, our results, in line with previous studies, did not show a strong relationship between clinical or functional data and future fall risk. Of course, this could be partially explained by the robustness of the subjects included in our sample, and the low number of events. Nevertheless, one third of the subjects included fell at least once during follow-up. The absence of statistically significant differences between groups based on clinical and functional data does not mean that no differences exist. Studies with larger sample sizes are warranted to confirm our findings. Moreover, it would also be interesting to consider data related to indoor and outdoor physical activities, daily-life environment and fall context.

6.5. Gait parameters and their relationship with fall incidence

Univariate analysis according to fall occurrence showed that subjects who fell during follow-up had lower FW gait speed, shorter CW and FW stride length and higher symmetry DTW cost than those who did not fall during follow-up. After regression analysis, higher stride symmetry DTW cost was significantly associated with a higher future fall(s) risk.

Before discussing the results showed in the results section, this chapter will discuss the methodological limitations meet starting the logistic regression analyses and the attempt to deal with. After this, the chapter will discuss the gait parameter and their relationship with future fall(s) one by one also integrating the previous reflection and their substantial inputs.

6.5.1. Comments concerning the methodological aspects

The logistic regression performed in this study suffers from some limitations that merit comment. Firstly, the number of events was low in absolute value (35 fallers after two years of follow-up) and therefore, the number of variables that can be included in the analysis is limited by the low number of events.

Consequently, a selection of the variables to be included was performed. First, only variables with a p-value <0.05 by univariate analysis were considered for further analysis. Then, to further reduce the number of candidate variables, only gait parameters with a p-value <0.05 after normalization for the right leg length were considered. Thus, 6 variables remained eligible for inclusion in the regression analysis, namely: marital status, IADL, SPPB score, stiffness, FW normalized stride length, and stride symmetry DTW cost. As explained previously, marital status was not considered for analysis. Finally, considering that each of the five other variables could potentially contain complementary information related to the future fall(s) risk, I chose to include all 5 variables in the logistic regression. Indeed, even though these variables were

correlated to each other, we considered that IADL score, SPPB score and stiffness do not contain the same information regarding the participants' characteristics, and thus, these variables would likely not have the same relationship with the future fall(s) risk.

Considering the normalized FW stride length and the stride symmetry DTW cost, these two variables could appear to yield different information related to the future fall(s) risk. Indeed, additional analyses were performed to further explore the relationship between the stride length and the stride symmetry and other anamnestic, clinical, functional characteristics and gait parameters. Actually, comparison analysis was performed according to quartiles of gait parameter performances.

Concerning the symmetry DTW cost (see appendix 3), twenty-four (25%) participants had stride symmetry DTW cost ≤ -28.1 % (Q1) while twenty-four (25%) participants had stride symmetry DTW cost ≥ 22.3 % (Q3). No anamnestic, clinical or functional characteristic was significantly different according to the quartile groups. The gait parameters showing a significant difference according to quartiles were CW stride symmetry ($p< 0.0001$) and DTW stride symmetry ($p< 0.0001$). Moreover, correlation analysis was also performed between CW stride symmetry and other anamnestic, clinical or functional variables and other gait parameters, confirming that CW stride symmetry was only correlated with other stride symmetry measures.

Concerning normalized FW stride length, 24 participants (25%) had a normalized FW stride length ≤ 1.64 (Q1), while 25 participants (26%) have a normalized FW stride length ≥ 2 (Q3). Compared to the participants who had a lower normalized FW stride length, those who have a higher normalized FW stride length were more often men (80 % of men versus 12.5% of men in the lower normalized FW stride length group, $p < 0.0001$), less often had a fear of falling (12 % had fear of falling versus 42% in the lower normalized FW stride length group, $p = 0.0008$). Concerning the quantitative variables (please see appendix 4), participants who had

a lower normalized FW stride length had significant lower scores on the Edmonton scale, higher grip strength and grip work, and higher functional performances on the FGA test and on the SPPB test. Furthermore, all gait speed measures were significantly different according to the FW stride length quartiles with a p-value < 0.0001, except for gait speed DTW cost. In the same way, all stride length measures were significantly different according to FW stride length quartiles with a p-value < 0.0001 except for stride length DTW cost. In our opinion, these analyses suggest that normalized FW stride length and stride symmetry DTW cost carry different information. Finally, in view of the distribution of the IADL scores, this variable was categorized into a dichotomous variable (totally independent for IADL, vs not totally independent for IADL).

Thus, the variables included in the logistic regression analysis were categorized IADL, stiffness, SPPB score, normalized FW stride length and stride symmetry DTW cost.

Furthermore, in line with the consensual recommendation that one variable should be included in the multivariate model for every 10 events, the number of variables that we could theoretically include in the logistic regression was only 4 (Motulsky 2002). A choice has to be made between risking overfitting of the model, or excluding the categorized IADL. Considering that both of these choices would mean limitations requiring further discussion, we performed both analyses, namely logistic regression analysis including the categorized IADL score (5 variables), as shown in the results section; and a second analysis, not including the categorized IADL score (4 variables); both models yielded the same results (see appendix 5).

Moreover, most of the variables considered for analysis were correlated (see appendix 6), which induces major limitations due to co-linearity. Actually, as shown in Appendix 6, IADL and stiffness and also FW normalized Stride length and SPPB are significantly correlated

in this sample. Furthermore, stiffness and SPPB also show substantial correlation. In view of this correlation, one solution might be to include variables with a p-value < 0.15 by univariate analysis in order to consider variables carrying "complementary" and "non-redundant" information concerning the future fall risk. Unfortunately, in this case, the number of variables to consider would be high: 20 variables. Considering these limitations, a stepwise method was used. Actually, a total of twenty variables was included in the stepwise method; 2 qualitative variables (anxiety and the feeling of dependency), and 18 quantitative variables, namely: IADL score, COVI score, Edmonton scale score, score obtained in the MoCA item assessing language, grip strength, muscular muscle mass, stiffness according the UPDRS scale score, SPPB score, CW and FW gait speeds, CW and FW stride lengths, the DTW stride frequency, DTW symmetry, gait speed DTW costs, stride length FW improvements, stride frequency DTW costs and symmetry DTW cost. Moreover, age, gender and right leg length were forced in the model as explicative variables. Eighty-one observations were used for regression analysis (fifteen observations were not used due to missing values for explanatory variables) and the number of events was 31. According to the stepwise method performed, longer FW stride length seemed have a protective effect against future falls with an OR of 0.026 (CI 95% 0.002-0.312, p-value = 0.004) when stride length was expressed in meters, and an OR of 0.70 (CI 95% 0.54-0.89, p-value=0.004) when stride length was expressed in decimetres. Although this model could be contested regarding the disproportion between the number of variables and the number of events, this result suggests that assessing gait parameters in challenging walking conditions could be useful to discern people at risk for future fall(s).

To summarize, the logistic regression analyses realized, whatever with or without including the IADL score, show that stride symmetry DTW cost is significantly and independently associated with the future fall(s) risk, while the stepwise method suggests that that gait parameter independently associated with future fall(s) risk is FW stride length, instead

of the stride symmetry DTW cost. Although the results obtained are not similar, these results are, in our opinion, not in opposition. Indeed, the variables included in the two analyses were not the same: in the restricted regression analysis, the normalized FW stride length was included in order to keep the number of included variables as lower as possible; including one variable (normalized FW stride length) instead of two variables (leg length and FW stride length). This choice could influence the results obtained. While the more restricted model seems to be more robust in terms of the number of variables included and the number of events observed, it could be influenced by the choice to include normalized FW stride length. On the other hand, the stepwise method including 20 variables for 31 events could be debatable in terms of robustness. As previously explained, the sample size and the low number of events are probably the most important limitations in this study. Thus, in the absence of consensual results, in our opinion, the FW stride length and the stride symmetry DTW cost both seem to be important. Indeed, the classification models obtained using the J48 classifier confirm the importance of considering both gait parameters together to identify people at risk for future fall(s).

After these considerations concerning the regression analyses, their results and limitations, this chapter will discuss the relationships between gait parameters and future fall(s) according to univariate and regression analyses and the comparison with available literature.

6.5.2. The gait parameters and their changes according walking conditions

A. Gait speed

Concerning participants who went on to fall during follow-up had a lower FW gait speed than those who did not fall during follow-up, while CW and DTW gait speeds were similar. Whatever the walking condition, gait speed and its change according to walking conditions were not associated with future fall(s) risk. Although a larger sample could bring to light a significant relationship that could be masked in this study, it is important to note that the available literature tends to confirm our results.

Indeed, concerning *CW gait speed*, Scott et al. (Scott, McLaughlin et al. 2015) included 135 women (mean age 76.7 years; range 70-92) in a 4 year longitudinal study in which gait parameters were assessed at inclusion using an electronic walkway, and failed to show a significant difference between future fallers and non-fallers in terms of CW gait speed at inclusion. Two other prospective studies also tend to confirm our results. First, the study by Maki et al. (Maki 1997) showed that slower gait speed is in fact more indicative of fear of falling than fall risk. Secondly, the study by Hausdorff et al. (Hausdorff, Rios et al. 2001) involving fifty-two subjects (36 women, 16 men) with a mean age of 80 ± 6 years in a one-year longitudinal study, also confirmed than fallers and non-fallers had similar CW gait speed. Finally, another prospective study, published by Verghese et al. (Verghese, Holtzer et al. 2009) and including 597 adults (mean age 80.5 years) showed a significant prospective relationship between slower CW gait speed and higher risk of fall(s) in the fully adjusted model. In fact, the sample included in the Verghese 'study was probably less robust than our sample, as the low mean gait speed of the Varghese' cohort (92.8 ± 24.1 cm/s.) suggests. Actually, the prospective relationship between CW gait speed and fall risk could probably be explained by the low mean

CW gait speed showed in Verghese' study. Indeed, the prospective Boston MOBILIZE study including a cohort of 765 community-dwelling women and men, mainly aged 70 years or older and in which, over 4.3 years, 1737 falls were recorded, showed that although low gait speed could be indicative of a higher fall risk, comfortable gait speeds ≥ 1.3 m/s do not seem to have a prospective relationship with future fall(s)(Kelsey, Procter-Gray et al. 2012). The finding that CW gait speed in our cohort was 1.64 ± 0.24 m/s in fallers and 1.74 ± 0.22 m/s in non-fallers could explain why CW gait speed did not show any prospective association with future fall(s) in our study.

Concerning *FW gait speed*, our results are similar to those of the TASCOG study, a one-year follow-up study involving 176 adults aged 60-86 years, living at home, without walking aids and non-demented (Callisaya, Blizzard et al. 2012). The TASCOG study assessed FW gait speed, cadence, step length and walk ratio at inclusion and their changes in CW and FW expressed in percentage. The authors considered three groups (rather than two groups in our study), namely a group with no fall during follow-up, a group with a single fall during follow-up, and a group with multiple falls. The study did not include comparison of gait parameters according to fall occurrence, thus limiting comparison with our results. However, after adjustment for age, sex, height and weight, no prospective relationship was found between FW gait speed at inclusion and occurrence of falls during follow-up. Thus, these findings tend to confirm our results.

Concerning *DTW gait speed*, no association was found in our study between DTW gait speed and future fall(s) risk. It is noteworthy that none of the gait parameter obtained in DTW showed a relationship with future fall(s). These results are in line with the available literature and, as previously detailed in the rationale section, three recent literature reviews (Zijlstra, Ufkes et al. 2008, Beauchet, Annweiler et al. 2009, Muir-Hunter and Wittwer 2016) have recently underlined that the usefullness of the dual task walking condition for the detection of

future fallers among community-dwelling older adults remains unclear, and additional prospective studies are warranted. In our opinion, the failure to observe any significant impact in this study supports the idea that the dual task walking condition is not useful to detect people at risk of future fall(s) among robust older adults living independently at home, and who have no recognised cogitive disorders, as previously suggested by Beauchet et al. (Beauchet, Annweiler et al. 2009).

The gait speed FW improvement and DTW cost were not signicantly different between fallers and non-fallers in our cohort. The results obtained in terms of gait FW improvement in the TASCOG study confirm the absence of a propective relation between fall(s) and gait speed FW improvement recorded at baseline (Callisaya, Blizzard et al. 2012). To the best of our knowledge, no published data are avalaible investigating the relation between gait speed DTW cost and prospective fall risk among healthy older adults.

B. Stride length

Briefly, in this study, subjects who fell during follow-up had a lower CW and FW stride length than those who did not fall, while DTW stride length was similar. After adjustment for leg length, only the normalized FW stride length showed a significant difference between groups. Logistic regression did not confirm the significant association between normalized FW stride length and future fall risk. But the stepwise method suggests a higher FW stride length was significantly and independently associated with a reduced fall risk.

Concerning *CW stride length*, our results are confirmed by available literature. Indeed, in the studies previously cited, Scott et al.(Scott, McLaughlin et al. 2015) and Verghese at al. (Verghese, Holtzer et al. 2009) failed to show any significant correlation after adjustment.

Concerning *FW stride length*, and *stride length FW improvement*, our results are partially confirmed by the TASCOG study, where the FW step length (expressed in cm) was significantly associated with the risk of experiencing multiple falls (RR 0.95; 95% CI 0.89-0.99) after adjustment, albeit not with the risk of a single fall. Similarly, the step length change (between CW and FW), expressed in %, was not associated with occurrence of a single fall, but was significantly associated, after adjustment, with multiple falls (RR 0.55; 95% C-I 0.36-0.84). Unfortunately, our limited sample size precluded any sub-group analysis according to the number of falls.

Furthermore, considering the results of the TASCOG study and the results obtained in our study using a stepwise method, the FW stride length seems to be protective against future fall(s). This may suggest that FW stride length is a marker of robustness. In order to highlight this potential relationship in our sample, additional comparisons were performed by quartiles of FW stride length. The hypothesis that FW stride length could be a marker of robustness could be supported by previous knowledge. From a biomechanical point of view, the stride length depends on the dynamic balance resources, the articular amplitude and the strength of several

muscles. Indeed, during the gait cycle, and especially during the swing phase (on which the stride length directly depends), an efficient dynamic postural balance, which can face even stressful walking conditions, is needed to keep the gait pattern safe without becoming "cautious". Moreover, during the swing phase, several muscles are called into action; the upper body moves forwards, the trunk twists about a vertical axis, the shoulder girdle rotates in the opposite direction to the pelvis, the arms swing out of the phase with the legs, the spinal muscles are selectively activated to keep the head moving as little as possible, and the hip flexes and extends according the swing phase steps. The strength of the ilio-psoas, gluteus maximus, quadriceps and tibial anterior muscles ensure a maximal forwards movement of the body, increasing the stride length (Levine, Richards et al. 2012). Finally, the articular amplitude of the joints of the body segments involved in the swing phase could limit or conversely allow the swing movement of the leg during the swing phase. Furthermore, considering that all these systems may potentially suffer from age related disorders, co-morbidities, medico-surgical events and drugs, it is not surprising to note that more robust people show higher FW stride length.

Additionally, the anamnestic, clinical and functional data available in this study also support the idea that FW stride length could be a marker of robustness. Indeed, additional comparisons were performed according to FW stride length performances. Among the 96 participants followed for two years, 25 participants (26%) had a FW stride length ≤ 1.38 meters (Q1), while 24 participants (24%) had a FW stride length ≥1.71 (Q3). Participants who had the highest FW stride length were often men (86% of men versus 24 % among those who had lower FW stride length, $p < 0.0001$), took fewer medications (92% took less than 5 medications daily versus 48% among those who had lower FW stride length, $p = 0.0009$), less often had a fear of falling (86 % without fear of falling versus 40 % among those who had lower FW stride length $p = 0.0006$). Regarding the quantitative data (appendix 7), participants who had a higher FW stride

length had a lower score on the Edmonton scale, stronger grip strength and more powerful grip work, even after correction for body height, greater skeletal muscle mass and higher functional performance on the FGA and SPPB tests.

In our opinion, these results suggest the usefulness of considering FW stride length as a marker of robustness and suggest that it should be considered in the clinical setting, but also in further research studies involving older adults.

Concerning the normalized FW stride length, to the best of our knowledge, no study to date has investigated the relationship between FW stride length normalized to the leg length and future fall(s) risk among older adults thus precluding comparisons.

Concerning *DTW stride length* and *stride length DTW cost*, our results are similar to those of previous studies and review concluding that stride length obtained in DTW is not useful to identify future fallers among healthy older people without cognitive decline (Zijlstra, Ufkes et al. 2008, Beauchet, Annweiler et al. 2009, Muir-Hunter and Wittwer 2016).

C. Stride frequency

Our results are in line with those of Scott et al. (Scott, McLaughlin et al. 2015) and Verghese et al. (Verghese, Holtzer et al. 2009) who also failed to show a significant relation between CW cadence (expressed in step/min. rather than in stride/s as in our study) and futures fall(s). The TASCOG study showed a significant relation between FW cadence (also expressed in step/min.) and multiple falls (RR 1.09; C-I 1.03-1.16) and between FW cadence changes (expressed in %) (RR 1.05; C-I 1.01-1.10), but no significant association with the risk of a single fall (Callisaya, Blizzard et al. 2012). As explained before, this result is different to those obtained in our study, and could be linked to classification of participants into three groups (no fall, a single fall, multiple falls) in the TASCOG study. As previous similar studies assessing gait parameters in DTW, the DTW stride frequency and the stride frequency DTW cost not seem useful to discern future fallers among older people without cognitive disorders (Zijlstra, Ufkes et al. 2008, Beauchet, Annweiler et al. 2009, Muir-Hunter and Wittwer 2016).

D. Stride regularity

In our study, the univariate and multivariate analyses failed to show any significant relationship between stride regularity and future fall(s). Of course, a larger sample size could highlight a significant difference that remains unrecognized in this study. Actually, some studies have suggested a relationship between falls and stride regularity. Indeed, Auvinet et al., previously showed, using the Locometrix® device, that people who fell had significantly lower stride regularity than those who did not fall (Auvinet, Berrut et al. 2003). Unfortunately this preliminary study did not make it possible to draw robust conclusion regarding the prospective relationship with future fall(s). Several years later, the results of a two-year prospective observational study involving 259 home-dwelling people aged 66 to 75 years who had never fallen (Mignardot, Deschamps et al. 2014), were published, but did not perform any comparisons according to the stride regularity recorded at baseline and the fall incidence during follow-up. Only principal component analyses results were published restricting the comparison with our results.

Furthermore, after consideration of other gait parameters with similar significance, and similar extraction and calculation processes, the cranio-caudal stride regularity computed by Bautmans et al. (Bautmans, Jansen et al. 2011), as an unbiased autocorrelation coefficient according to Moe-Nilssen and Helbostad (Moe-Nilssen and Helbostad 2004) and the stride-to-stride variability calculated by Senden et al. (Senden, Savelberg et al. 2012) based on cranio-caudal accelerations, were considered for comparison. These cross-sectional studies (both including people with recent fall history) also observed that older people at risk for falls have low stride regularity. Nevertheless, in view of the lack of available studies highlighting a prospective relationship between lower stride regularity and higher future fall risk, the stride regularity should be considered as a marker of fall history rather than a marker of future fall(s).

E. Stride symmetry

In this exploratory study, fallers had positive mean stride symmetry DTW cost, meaning that they decreased their stride symmetry during DTW, while non-fallers had negative mean stride symmetry DTW cost, meaning that their stride symmetry increased in DTW. Moreover, the stepwise method shows higher stride symmetry cost are significantly and independently associated with an increased risk of future fall(s).

The measurement of gait changes between the comfortable walking condition and the dual task walking condition is an original concept first published by Montero-Odasso. Indeed, according to this author, the type and amplitude of the gait changes measured from the comfort through the dual task walking condition have to be considered in addition to gait performances obtained in CW or DTW alone (Montero-Odasso, Sarquis-Adamson et al. 2017). Unfortunately, to the best of our knowledge, no similar data (regarding the symmetry DTW cost) is available in the published literature for strict comparison.

However, a cross sectional study including 15 idiopathic fallers and 11 healthy controls, confirmed that idiopathic fallers had a less symmetric gait pattern in DTW than in CW (Yogev, Plotnik et al. 2007).

Furthermore, if we assume that stride symmetry assesses a similar gait component to step regularity, then our results are in line with Bautmans et al. who compared gait patterns of 40 older adults who had a high fall risk (mean age 80.6 ± 5.4 years) with those of 41 non-faller older adults (mean age 79.1 ± 4.9 years) founding that older adults at risk of fall had less step regularity than non-fallers (Bautmans, Jansen et al. 2011).

In addition, in a one year cohort study including 319 community-dwelling older adults (mean age 75.5 ± 6.9), gait symmetry, assessed as harmonic ratio measured in three axes and in daily-life walking conditions, was found to have a negative relationship with future fall risk (van Schooten, Pijnappels et al. 2016).

Actually, stride symmetry could be linked to the automatic stepping activity coming from central pattern generators (CPGs)(Yogev, Plotnik et al. 2007). Our results suggest that healthy older people walking in the dual task condition benefit from the locomotor rhythmic stepping control. In our opinion, this “automatic” gait pattern behaviour is supposed to enable the subject to adequately perform the additional cognitive task while at the same time, ensuring the distance is covered.

Our hypothesis is partially confirmed by a study of 100 healthy older adults assessed for gait and cognitive performances during single or dual-tasking (Beauchet, Allali et al. 2010). In this study, while most of the subjects had higher stride time variability and worse backward counting performance while dual-tasking, a limited number of subjects significantly reduced their gait variability and simultaneously improved their backward counting performance while walking compared to sitting. This behaviour was called the "magnet effect" because of the hypothesis that in some subjects, dual tasking leads to a global positive effect as much on reducing gait variability as in cognitively performing. In our opinion, even if the data observed in our study do not allow us to confirm the hypothesis formulated by Beauchet et al., this “magnet effect” could perhaps be a marker of healthy gait pattern ageing.

Unfortunately, as previously detailed, comparative analysis according to stride symmetry DTW cost performance in this sample did not yield a deeper understanding of the potential determinants of stride symmetry DTW cost. Considering the available literature linking gait symmetry with CGP activity, data related to central nervous systems structures and cognitive functions could be more relevant to take into account.

F. MTC values

In this study, all MTC values (whatever Mean MTC, Min MTC or the values of variability) and their changes in FW and DTW were similar between fallers and non-fallers. To the best of our knowledge, this is the first prospective study focusing on prospective relationships between MTC values and future fall risk among older adults without a recent fall history and living independently at home. Thus, no similar study is available at this time for comparison.

Several points are noteworthy. Firstly, considering the sample size, the absence of significant difference both by univariate analysis and logistic regression and the stepwise method, does not mean that no relationship actually exists between MTC values and future fall(s) risk. Moreover, in our opinion, one of the most important points in the further exploration of the relationship between MTC and future fall(s) risk is the circumstances of the fall. Indeed, in our study, no distinction was made between participants who tripped (with or without an obstacle on the ground), slipped (with or without an obstacle on the ground), had a stepping reaction or lost their balance. However, some falls, especially those due to tripping, could be linked to MTC values. Thus, since our study did not distinguish trips and slips, the relationship between MTC values and fall(s) could be unrecognized.

6.6. The use of WEKA

The general linear model seems to fail to understand the complexity of the relationship between gait patterns and fall risk among older adults. Indeed, although the inclusion and exclusion criteria used in his study were designed to select a homogeneous population, one of the major characteristic of the older adult population is their heterogeneity. Moreover, as previously discussed, falls among older adults could be linked to several different mechanisms (and furthermore, different mechanisms could be associated). Thus, falls among older adults are a complicated and multifactorial issue. Considering the heterogeneity of the population included and the complexity of the issue addressed, it appeared logical to use data mining to better take into account individual gait patterns rather than differences between groups, with a view to highlighting the potential relationship between gait parameters, walking conditions and future fall(s) risk.

As previously explained, walking is not an automatic task (Hausdorff, Yogev et al. 2005). Indeed, among older adults, one of the priorities of the postural system is to reduce head motion during locomotion (Kavanagh, Barrett et al. 2005). According to Menz et al. adopting a more conservative gait pattern during ageing may be a compensatory strategy to ensure that the head and pelvis remain stable (thereby preventing falls) (Menz, Lord et al. 2003). As previously outlined, individual gait patterns are the result of the relationship between the central nervous system and the musculoskeletal system, of which coordination and perfect functioning are necessary to obtain motion control that is as efficient as possible. Given that age-related effects on the central nervous and musculoskeletal systems could affect postural balance efficacy, it could be assumed that, in case of adaptive resources, the native gait profile is adapted to compensate for age-related effects, thus keeping the walk safe and as efficient as possible.

In light of this hypothesis, the use of a tool that considers individual gait profile seems to be more informative than group analysis.

Actually, the use of the J48 was inspired by a German study comparing the usefulness of functional balance and gait tests with those of gait parameters obtained using an accelerometric method to discern faller or demented subjects among older adults (Gietzelt, Nemitz et al. 2009, Marschollek, Nemitz et al. 2009, Marschollek, Rehwald et al. 2011, Marschollek, Rehwald et al. 2011, Gietzelt, Wolf et al. 2013). Through several published papers, this German team demonstrated the utility of considering gait patterns (Marschollek, Rehwald et al. 2011), presented the WEKA (Marschollek, Rehwald et al. 2011) and showed that the best classification rate was obtained using the J48 classification tool (Gietzelt, Wolf et al. 2013). Considering the similarities between the German study and our own research, after consideration of the available literature concerning the WEKA (currently more than 140 published references available on the website), and considering that the WEKA software makes supervised machine learning easy to use for people who wish to apply it to their own data, we chose to apply this approach to our data. Given the step by step selection of attributes involved in the short versions of the models (in order to increase the performance of each model), we chose to select one by one the attributes to be included, in order to check the utility of each (one by one) in the model, retaining those that were useful (i.e. variables whose removal led to a decrease in model accuracy) and to avoid retaining variables that were not useful (i.e. those whose removal led to similar or higher model accuracy). Other methods, that are faster to use, are suggested by the software. In order to choose the best way to select the attribute to retain in the short version, we compared our step by step methods with the methods suggested by the software. When the number of attributes was small, the methods suggested by the software were more useful. However, when the number of attributes increased, the usefulness of the step by

step method become clear (data not shown). For this reason, we chose to apply this step by step attribute selection method.

When applied to our data, the classification tool confirmed that gait patterns could be useful to discern, among older people without a history of falls, those who were at risk of future fall(s). The models obtained also showed the interest of considering gender, clinical factors (stiffness) and functional aspects (IADL and SPPB scores) of the subjects to be classified. Indeed, the J48 classifier processed two classification trees; i.e. the ninth model, which was shorter and easier to use, and made it possible to identify 24 fallers among 35; and the tenth model, which was longer but performed better, making it possible to identify 28 fallers out of 35.

As in the German study, our results show the interest of adding clinical or functional variables to gait pattern to increase the performance of the classification tree. Indeed, the classification tree involving only gait patterns in the German study (as in the seventh model of our study), showed similar performance to the short version of the seventh model in our study with an accuracy of 80% (80% in our study), a sensitivity of 58% (69% in our study), a specificity of 96% (87% in our study), a positive predictive value of 92% (75% in our study), a negative predictive value of 77% (83% in our study) and a ROC area of 0.83 (0.79 in our study). After adding data related to physical performance, the performances of the German classification tree increased (as did those of the tenth classification tree here) with an accuracy of 78 % (versus 84% in our study), a sensitivity of 74% (versus 80% in our study), a specificity of 82% (versus 87% in our study), a positive predictive value of 74% (versus 78 % in our study), a negative predictive value of 74% (versus 88% in our study) and an AUC of 0.87 (versus 0.84 in our study). The German study and our own study both support the idea that gait pattern, clinical and physical performance included in a classification tree could be useful to identify among older adults those who are at risk of future fall(s).

6.6.1. Use of the J48 classifier

The J48 classification tree is a supervised learning algorithm based on hierarchical relationships between gait parameters and the class (fallers or non-fallers). Thus, the first variable the model uses is the gait parameter that makes it possible to classify as many people as possible in one of the two classes. In fact, the classifier tests the ability of each gait parameter to classify fallers and non-fallers; the tool then chooses as first node the gait parameter that classifies the larget number of subjects as faller or non-fallers. After this step, the classifier explores the data to find the second variable to use as a node to classify as many people as possible, and so on. In our opinion, this orginal approach to processing makes it possible to move away from the usual "groups-based analysis" and explore unknown links between the gait parameters composing the individual gait patterns. Morevoer, some parameters involved as classification nodes make it possible to classify people as fallers or non-fallers, whereas other gait parameters classify people in only one of the classes or not at all. In our opinion, this supports the idea some gait parameters are directly and strongly linked to fall(s), while others have a more complex relationship with fall risk, and need to be balanced or compensated by other gait parameters.

In case of missing values, instances for which the relevant attribute value is missing are notionally split into pieces, one piece for each branch, in the same proportion as the known instances go down the various branches (Witten I. 2005). This automated calculation process applied by the J48 explains why some classification rates shown in brackets in the "enhanced" version of the models are not integers.

Finally, a 10% cross-validation method is used to measure the performance of each model obtained. As previously explained, during the first 10 runs of the algorithm, 10% of the data are used for the test data and the remaining 90% are used as the training set. On the tenth iteration, the entire dataset is used as training data. The eleventh run takes into account the

entire data set as the test set. The algorithm output shows the average of eleven runs. The standard way of predicting the error rate of a learning technique given a single, fixed sample of data is to use stratified 10-fold cross-validation. One may ask, why 10? Extensive tests on numerous datasets, with different learning techniques, have shown that 10 is about the right number of iterations to get the best estimate of error, and there is also some theoretical evidence that backs this up. Although these arguments are by no means conclusive, and debate continues to rage in machine learning and data mining circles about what the best schema for evaluation is, 10-fold cross-validation has become the standard method in practical terms. Tests have also shown that the use of stratification improves results slightly. Thus the standard evaluation technique in situations where only limited data is available is stratified 10-fold cross-validation (Witten I. 2005).

To summarize, the classification nodes were chosen based on the ability to classify the highest number of people, thereafter the performances of the model was evaluated using cross validation processes. This validated process strongly support the robustess of the models obtained.

6.6.2. Models obtained and process used

The ten models were obtained one by one, including gait parameters or clinical variables step by step, and taking into account the performances obtained on each previous model to select the variables to include in the next one. Indeed, considering that the highest sensitivity is obtained in the fifth model, and the highest specificity in the second model, the sixth model included the FW gait parameters (included in the second model) and the DTW cost gait parameters (included in the fifth model). Thus, in our opinion, the most useful knowledge coming from the six first models is that FW seems to be helpful in identifying non-fallers (improving the specificity and the negative predictive value in the second model) while the DTW costs improve sensitivity and seem to be helpful in identifying fallers. Indeed, this potential relationship between FW gait performances and non-fallers is also supported by the previously explained relationship between the FW stride length and robustness.

Another lesson coming from these first six models is that the selection of attributes one by one made it possible to obtain a short version model with the highest performances in terms of the number of correctly classified instances (accuracy).

The seventh model included all gait parameters obtained (whatever the walking condition) and their FW improvement and DTW cost. Once again, the performances of the six previous models obtained oriented the choice of walking conditions to consider. Given that the one-by-one attribute selection process was efficient, all gait parameters were included. In fact, the short version of the seventh model had better performances than the six models previously obtained, but remained insufficient in terms of sensitivity and thus, further models were processed to improve the sensitivity and positive predictive value.

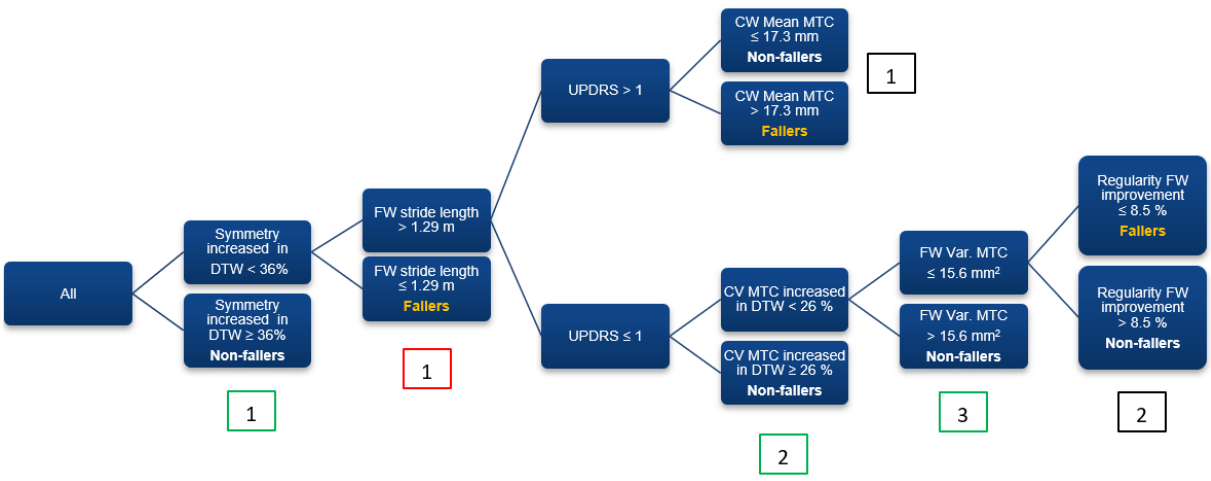
The eighth model, including all gait parameters obtained (whatever the walking condition) and their FW improvement and DTW cost as well as age, gender and right leg length, had higher specificity than the seventh model. The one by one attribute selection process

confirmed that age and leg length are useless attributes. Gender appears useful in helping the model to perform better, even though it was not involved in a classification node. The ninth and the tenth models were developed in view of these previous findings.

The ninth model included all the gait parameters (whatever the walking condition in which they were obtained), their changes compared to CW, as well as IADL score, SPPB score and stiffness based on the UPDRS score. The ninth model appears to perform better and was shorter than the seventh and eighth models, highlighting the utility of adding clinical data. In terms of the additional data, the IADL score was not useful and was removed. The stiffness and SPPB scores increased the performances of the model and the UPDRS score was used as classification node. Indeed, stiffness appears early in the classification tree and its inclusion as a classification node changed the following nodes, suggesting that this parameter is relevant to future fall(s) risk.

After comparison with the previous models obtained, the ninth model confirms the usefulness of the symmetry changes in DTW, FW stride length and stride regularity changes in FW for classification, and suggests considering the utility of Mean MTC in CW and CV MTC changes in DTW to improve classification performances.

Actually, as shown below, three parameters (indicated by the green boxes) seem to be protective against falls, namely high symmetry increase in DTW, high CV MTC increase in DTW and high FW Var. MTC.



The relationship between the increased stride symmetry in DTW has already been discussed above. The measures of the variability of MTC (DTW CV MTC and FW Var. MTC) suggest an inverse relationship between the variability of the MTC and the fall risk. In our opinion, these results could suggest that the variability of MTC is related to efficient postural mechanisms involved in the swing phase to keep the body center of mass in the base of support. Indeed, Schultz et al. have previously shown that mean MTC has no linear relationship with gait speed, step time and step length, suggesting that mean MTC is not associated with the same control mechanisms (Schulz, Lloyd et al. 2010). Moreover, the same study also highlighted that the task-related effect on mean MTC is different according the type of secondary task. This implies that mean MTC could be more closely related to whether the walking condition challenges postural balance mechanisms than the automatic step production processes. Moreover, Hamacher et al. showed that MTC variability, assessed in DTW among older adults, is not related to stride length and stride time variability (Hamacher, Hamacher et al. 2016). Thus, although the study by Schutz et al. assessed mean MTC rather than variability of MTC, these two studies and our results suggest two co-existing postural control mechanisms ensuring maximal head and pelvic stability: one depending on the CPG network production of a rhythmic stepping activity (leading to an increase in stride symmetry and regularity) and the other, probably depending on cortical locomotor control areas, a more complex, less automatic mechanism controlling the swing phase leading to the foot landing pattern. Furthermore, the implication of the second mechanism is also supported by the fact that, among people with stiffness, the last classification node involved MTC values. Unfortunately, without accelerometric data relative to head motion, the design of this exploratory study does not make it possible to confirm our hypothesis, which, in our opinion, warrants exploration in further studies.

Briefly, these comments are in line with previous literature showing that gait variability in healthy older people could be a compensatory strategy to ensure that the head and pelvis remain stable, reducing the fall risk (Woollacott and Tang 1997, Menz, Lord et al. 2003, Kavanagh, Barrett et al. 2005). Thus, among older adults without fall history, the variability of the MTC observed in challenging walking conditions seems to be a marker of the postural mechanisms involved.

Furthermore, in the ninth model, one parameter (indicated in red boxes) seems to lead to fall(s), namely short FW stride length. It is surprising to notice that the ninth model highlights a relationship with shorter FW stride length, while the results of the stepwise method suggested that a longer FW stride length seems to be linked with a reduced fall risk. This relationship between the FW stride length and fall risk, emerging in this model, is probably explained by the fact that the model previously identified a high number of non-fallers based on their symmetry increase in DTW. Indeed, those who are at lowest risk of falling (and probably having the highest FW stride length) had already been classified by the first node.

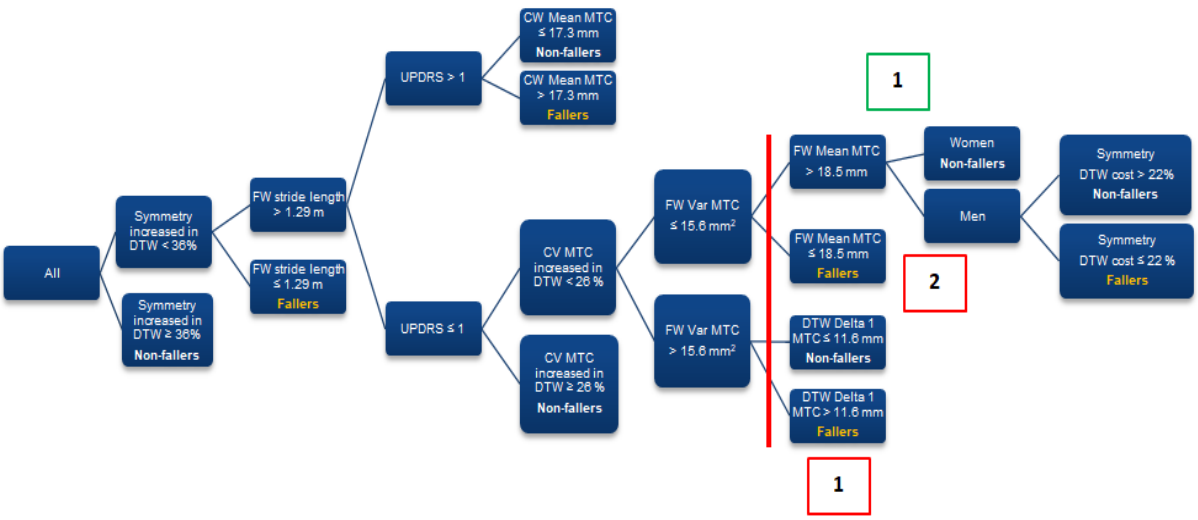
Finally, two black boxes indicate classification nodes involving gait parameters that identified both fallers and non-fallers. At this time, it is not possible to understand why the classification at this node is dichotomist. Several hypotheses could be put forward; such as, the gait parameters involved could lead to the fall, or could be compensatory mechanisms, or the final classification node position explains the dichotomy. Available literature suggests that older adults have tendency to have a more flat-footed landing to ensure a more stable pattern (Winter, Patla et al. 1990). The link between a shorter CW Mean MTC and non-fallers seems to be based on a compensatory mechanism. Moreover, among people suffering from stiffness that reduces postural reactions, a decrease in the distance between the toe and the ground could reduce the time needed to put the foot down on the ground in case of postural adaptations. Thus, although in people without stiffness, high mean MTC seems intuitively protective, among

people with stiffness, a reduced Mean MTC could be an adaptation of the individual gait pattern to ensure a more stable foot landing. Finally, among those who did not sufficiently increase the CV of MTC in DTW or the FW variance MTC, a higher stride regularity improvement in FW helps to discern non-fallers, and seems to be a compensatory mechanism. Once again, the use of a classification tool makes it possible to observe gait compensation mechanisms and supports the idea of two co-existing postural control mechanisms (one to ensure stride symmetry and regularity, and the other to ensure balance aspects related to swing phase). In this context, given that the univariate analysis showed a dichotomisation in the regularity of FW improvement among fallers and non-fallers (Table 9; some subjects showed a "positive" regularity FW improvement, while others showed a "negative" regularity FW improvement, regardless of whether they were fallers or not) and in order to deepen our knowledge of potential compensatory gait mechanisms, an additional comparison was performed between volunteers who had a "positive" regularity FW improvement and those who had a "negative" regularity FW improvement (see appendix 8). Briefly, data concerning the regularity FW improvement were available for 96 people. Among these, 57 (59.4%) had a "positive" regularity FW improvement, meaning that they increased the regularity in FW, while 39 (40.6%) had a "negative" FW regularity improvement meaning that they decreased their regularity in FW. No anamnestic, clinical or functional variables were significantly different according to the regularity FW improvement; only certain gait parameters were different between these two groups. Indeed, those who had a positive FW regularity improvement had higher FW regularity, and higher FW Min MTC, while those with a negative FW regularity improvement had lower Min MTC in FW. We could hypothesise that the positive regularity FW improvement could be protective gait behaviour in order to increase the FW regularity and the FW Min MTC, i.e. to increase the distance between the toe and the ground during the swing phase, thus reducing the

risk of tripping. However, the p-value (< 0.05 but > 0.001) does not enable us to consider these differences as significant. A larger sample could confirm our hypothesis.

To summarize, the ninth models suggests the following: - a greater increase in symmetry in DTW is protective against fall risk; - among those who are unable to increase symmetry sufficiently, a FW stride length lower than 1.29 m leads to a fall; - among those whose FW stride length is ≥ 1.29 m, the model considers stiffness. In case of stiffness, a reduced CW MTC is protective against a fall, allowing a quick stepping reaction. In the absence of stiffness, higher MTC variability during challenging walking conditions is the result of postural adaptation mechanisms targeting the swing phase, ensuring a safer foot landing and a more stable walking pattern. At the same time, among those who do not activate postural mechanisms controlling the swing phase aspects, the automatic step production ensures a regular walking pattern.

As the eighth model showed that adding data relative to gender tended to increase the specificity of the model. We processed the tenth model involving all gait parameters (whatever the walking condition they were obtained in), their changes compared to CW, IADL score, SPPB score and stiffness based on the UPDRS score, as well as age, gender and right leg length. The tenth model showed the best performances overall, classifying 27 fallers out of 35 (i.e. 8 false negatives), and 53 non-fallers out of 61 (i.e. 8 false positives). Thus, the involvement of gender in this model made it possible to discern more fallers than in the ninth model, explaining the higher sensitivity and negative predictive value. As in the eighth model age and leg length were uninformative. However, the involvement in the same model of gender and data relative to functional status and physical performances led to better performances than the ninth model, and final classification node modification. Indeed, the tenth model considers the FW Mean MTC and the Delta1 MTC in DTW rather than the regularity FW improvement.



The figure above shows the tenth model, the red line identifies the classification node changes linked to the inclusion of gender in the model. As shown in the figure, one parameter seems to be protective (in the green boxes); namely a high Mean MTC in FW among women (once again, among people without stiffness high Mean MTC values seem to be protective),

while two red boxes indicate parameters linked to the fall risk; namely, a high delta1 MTC (Max MTC value – Mean MTC value) in DTW and lower FW Mean MTC.

Regarding the first red box, the model shows that among people whose FW variance MTC is high, people who have a Delta1 MTC > 11.6 mm in the dual task walking condition were classified as fallers. Because the delta1 MTC appears after consideration of the increase in CV MT in DTW, the involvement of the delta 1 MTC is, in our opinion, more likely linked to a reduced mean MTC value rather than to high MTC variability. Unfortunately, the design of this study does not allow us to confirm this hypothesis. Given the number of misclassified instances, the number of fallers identified using the delta1 MTC is similar to the number of instances previously misclassified by the ninth model.

Regarding the second red box and according to the model, among people with a lower increase in CV MTC in DTW (and even in case of high MTC variance in FW) a lower FW Mean MTC leads to fall(s). Once again, the model suggests that among people without stiffness, higher mean MTC values protect against fall(s).

Unfortunately, the reason why the last classification node used symmetry DTW cost to detect fallers among men remains unclear. Considering this is the last classification node, we may suppose that additional gait patterns or clinical data could lead to other classifications.

6.6.3. Summary of knowledge from the models

Although our results warrant confirmation in a larger cohort, the results obtained are encouraging. In fact, the application of a classifier tool to our data made it possible to obtain a classification model that correctly identified 28 future fallers (among a total of 35 future fallers) and 53 non-fallers (among the 61 non-fallers). In this context, several models confirm the importance of considering stride symmetry DTW cost, FW stride length, stiffness and mean and variability measures of MTC. Furthermore, the models show that clinical and functional data have additional utility in increasing the performance of the models obtained.

In our opinion, the models obtained using the J48 support the following posits: - gait pattern assessment could help to identify, among older adults, those who are at risk of future fall(s); - individual gait patterns are the result of several gait components that balance each other, and produce a walk that is as efficient as possible; - gait parameters have to be considered in terms of individual gait patterns rather than for any overall group effect.

6.7. Overall significance of the results obtained whatever walking condition, instrumental method used for data acquisition or statistical approach

Since the main goal of this study was to investigate whether gait pattern assessment could be useful to identify, among older adults without a fall history, those at risk of future fall(s), the results obtained in this study are encouraging. Indeed, at the level of the whole cohort and according to logistic regression analysis, a greater FW stride length seems to be associated with a lower risk of fall(s). Furthermore, the use of a classifier tool, addressing the same hypothesis at the individual level, suggests a compensatory relationship between gait parameters, supports the idea of two co-existing postural balance mechanisms, and confirms the utility of considering symmetry DTW cost, FW stride length, mean and variability measures of MTC when assessing individual gait patterns in different challenging walking conditions.

Actually, even if the models shown here need to be validated in other samples, the use of a classification tool on our data made it possible to obtain models that classified subjects as "future fallers" or "non-future-fallers" based on gait performances recorded at inclusion. These models also support the idea that gait parameters are related each other in a dynamic process that involves robustness, weakness and compensatory behaviours linked to the individual profile of the walker. Indeed, compensatory behaviours exist and could be observed daily by clinicians during their practice. The study by Levinger et al. illustrates one of the compensatory behaviours that older adults may adopt. In this study, the authors investigated how the MTC of older adults with osteoarthritis compared to MTC of controls. Surprisingly, MTC were similar in both groups and the authors identified the compensatory behaviours adopted by people with osteoarthritis to keep safe (Levinger, Lai et al.). Unfortunately, most of the time, the heterogeneity of these behaviours limits their systematic assessment and understanding. As suggested in this exploratory research work, no single gait parameter alone is sufficient to

detect, among robust older adults, people who are at risk of future fall(s). Furthermore, this works also supports the idea that the relationship between gait parameters and fall risk does not seem to be linear, probably because of individual compensation mechanisms. This hypothesis was also suggested in the study of Callisaya et al. in which some people in the fastest quartile of gait speed in FW were at increased of fall risk while the majority of the people in the second and the third quartiles of gait speed were at lower risk (Callisaya, Blizzard et al. 2012). Therefore, further research is needed to better understand the underlying relationships between gait parameters and future fall risk and in this context, data mining use could be useful.

6.8. Strengths and limitations of this work

6.8.1. The strengths

One of the main strengths of this study is his relatively well documented sample and outcomes. Indeed, initial phone contact, anamnesis, clinical exam and functional evaluation guarantee the absence of subjects who presented exclusion criteria. Furthermore, anamnestic, clinical and functional data obtained at inclusion made it possible to discuss the gait patterns obtained and their relationship with future fall(s). Finally, considering only gait parameters show significant differences between fallers and non-fallers among a fully documented cohort strongly supports the idea that gait pattern assessment is more useful than other tools (considered in this work) to detect people at risk of future fall(s) among robust older adults.

In our opinion, a further strength of this study is the low rate of drop outs. Indeed, fewer than 10 percent of those included did not complete follow-up to two years. After comparison, subjects who did not complete the follow0up were not significantly different from those who did. The reduced dropout rate and the similarity between those included and those who dropped out suggests that fall(s) incidence among the people who left the study iw likely similar, and thus, no substantial missing information or data would change the results or the conclusions of this study. Indeed, the time spent at inclusion to explain the goals of this study, and the relevance of the follow-up also explain the high number of negative outcomes reported and supports the relevance of the results shown in this study. Indeed, at inclusion, all volunteers received a follow-up book explaining the goals of the study, as well as the definition of falls and missteps, with a page dedicated to each phone contact in order to note negative outcome circumstances and comments. Finally, the contact details of the main investigator and the occupational therapist in charge of phone contacts were given. During follow-up, volunteers were contacted every 3 months to record outcomes, answer some questions and fix the date for the next phone

contact in order to maximize the answer rate and reduce loss of contact. Unfortunately, despite the rigour of this process, data from two volunteers were not considered for analysis and follow-up was interrupted.

The methods used during acquisition and extraction data strongly suggest that the results obtained can be considered as robust. Indeed, data acquisition was performed in a laboratory environment dedicated to movement analyses, the equipment used is validated, installed and used by experienced investigators, and all gait analyses were supervised by the main investigator.

Moreover, the use of three different walking conditions and the calculation of the gait changes occurring between CW and FW (i.e. gait parameter FW improvement) or between CW and DTW (i.e. gait parameter DTW cost) is one of the originalities of this experimental research. To the best of our knowledge, this study is the first to show so many gait parameters in three walking conditions and their changes according to the walking condition. Moreover, systematic randomisation of the FW and DTW rules out any fatigue effect that could be due to performance of the same walking condition. The advantages of considering these different walking conditions are numerous. First, the use of the comfortable walking condition makes it possible to obtain the spontaneous gait pattern and reference data among relatively robust people without a fall history. The spontaneous gait pattern and its changes in FW and DTW could further be addressed to deepen our knowledge of individual gait patterns and their changes in case of stressful walking conditions. Second, the use of the fast walking condition makes it possible to show that gait performances in this walking condition could be useful for the evaluation of fall risk among robust people, arguing in favour of its systematic use in clinical setting. Thirdly, the use of the dual task walking condition confirms that gait parameters obtained in DTW are less useful than among older people with cognitive decline. Given that serial subtraction of seven from 100 leads to the greatest deterioration in gait performances as

shown by Muir et al. (Muir, Speechley et al. 2012), the cognitive task was probably challenging enough to mobilise attentional resources of the volunteers. However, given that the subjects included were relatively robust and free from cognitive disorders, the attentional resources they had should have been sufficient to perform the task and ensure safe walking patterns, even among those who would fall in the next two years. Additional data relative to the neuropsychological performances of the volunteers could be helpful to confirm this hypothesis. Unfortunately, at this time, these data remain unavailable. Furthermore, while the gait pattern obtained in DTW seems to be less useful than hoped, the dual task costs calculated in this study appear to be useful to detect people at risk for future falls (symmetry DTW cost and CV MTC DTW cost). In our opinion, these results illustrate that gait patterns have to be considered as dynamic processes where gait changes are at least as important as gait performance.

In addition, the systematic anthropometric measurements make it possible to consider individual physical characteristics when discussing gait patterns and their relationships with fall(s) and misstep(s), but also show that many gait parameters were correlated to the leg length. This supports the idea that cohort studies including people who have substantial anthropometric differences should consider these relationships.

Finally, although this study is not the first to apply data mining to gait patterns, the use of a classifier tool offered by the free platform WEKA is undoubtedly another originality of this study and supports the idea that gait parameters could have non-linear relationships with future fall(s). The use of data mining makes it possible, despite the reduced sample size and the limited number of falls observed during follow-up, to further explore the relationships between gait parameters and future fall(s) risk. Indeed, considering each gait pattern individually, the J48 overcomes the limitations of comparison and regression analyses in a small sample. The systematic and logical approach used to obtain the models, and the use of a 10-fold cross validation process strongly support the robustness of the models obtained. The models obtained

support the idea of two co-existing balance mechanisms, where one is linked to step production and gait pattern regularity, and the other is linked to the control of the swing period in more challenging walking conditions. Moreover, the models obtained also suggest non- linear relationships between gait parameters and future fall risk, suggesting co-existing compensation mechanisms.

6.8.2. The limitations

Unfortunately, this study has some limitations that deserve to be underlined. First, the sample size is small. Indeed, a larger sample would make more closely resemble the characteristics of the real population of healthy older adults. Thus, the absence of significant relationships between some gait parameters and future falls does not mean that no significant relationship exists. Furthermore, a larger sample would make it possible to observe a higher number of falls and include more explicative variables in the logistic regression.

Second, although this study only considered accidental fall(s) and the follow-up made it possible to exclude falls due to organic disorders such as cardiac or neurologic disorders, our study did not take into account the number of falls (i.e. no difference was made between people who fell once and those who fell several times), the fall consequences, or the context in which the fall occurred (e.g. at home, in the street, on snow, in a dual task situation, in more complex situations, during challenging physical activity, tripping, slipping or losing balance). Consequently, the relationship between MTC value and future fall(s) was not further discussed.

Moreover, this study did not take into account the cognitive performance during the dual task walking condition. Thus, we were unable to identify whether priority was equally given to the cognitive and the motor task, or whether the priority was different among fallers and non-fallers.

Furthermore, although the gait parameter changes between CW and FW or the DTW were compared in terms of fall(s) or missteps(s) incidence, no comparison was made between the walking conditions. Actually, the comparisons made considered the outcomes as the main point of view. Additional analyses should consider the comparison of gait parameters according to the walking condition used.

Last but not least, the acquisition protocol did not make it possible to obtain robust data concerning the double support, stance and swing times and Locometrix® only gives mean

values, and thus, stride to stride variability measures were not available for analysis. This technical limitation in obtaining gait variability measurements (except for MTC) is in our opinion the main limitation of this study. Indeed, the choice to use an accelerometric method and an opto-electronic method in this study was based on consideration of the main advantages of each method, their validation among older adults, their availability in the University of Liège and the previous experiments with this equipment. Using these two instrumental methods simultaneously appeared to be complementary, obtaining mean values on long walking distances, using the Locometrix, and stride by stride analysis using the opto-electronic method. Fortunately, an ambulatory accelerometer-based system used with a signal processing algorithm to automatically extract the temporal gait events involved in the estimation of the phase/sub-phase durations is under development and validation in the University of Liège and further work should confirm, using this software, an accelerometric method allows stride-by-stride analysis.

7. Conclusions

In conclusion, the results obtained in this work support the idea that gait pattern assessment could be useful to detect, among healthy older adults, those at risk of future falls, especially the symmetry DTW cost and the FW stride length.

Moreover, this study shows the utility of considering data mining processes to assess individual gait patterns. Actually, the use of a classifier tool on our data made it possible to obtain a model based on gait patterns and data related to clinical and functional characteristics of the participant that detected more than 80% of people who would go on to fall during follow-up. In our opinion, this study strongly suggests that a better understanding of this complex issue in a heterogeneous population requires consideration of individual profiles integrating gait performance but also clinical and functional measurements (as usually done in a comprehensive geriatric approach).

Of course, our results warrant confirmation in a larger sample and the models obtained need to be validated in terms of the relevant gait parameter to consider in classification nodes, the threshold values used and the performances obtained.

However, considering the limitations but also the strengths of this exploratory research work, we consider the results obtained to be robust and encouraging, and justifying further studies in this field of research.

8. Perspectives

Based on these encouraging results, further studies should be planned to confirm the usefulness of gait patterns to identify people at risk of future fall(s). In fact, based on the results obtained in this exploratory study, further studies should consider calculation of statistical power in order to include enough participants to ensure adequate statistical power.

Moreover, the inclusion of a larger sample would allow the step by step development and validation of a signal processing algorithm initially started in this study among a small proportion of volunteers. Indeed, one engineer working in the INTELSIG Laboratory for the Department of Electrical Engineering and Computer Science, in University of Liège (M. Boutaayamou), has developed and validated an ambulatory accelerometer-based system used with a signal processing algorithm to automatically extract the temporal gait events involved in the estimation of the phase/sub-phase durations, namely heel strike (HS), toe strike (TS), toe-off (TO), maximum of heel clearance (MHC), and maximum of toe clearance (MTC) making it possible to obtain the durations of stride, stance, swing, and double support phases. Data relative to the minimal toe clearance are also available for calculation. The main advantages of foot worn devices are to obtain gait sub-phase information and to record gait patterns over longer distances than using a walkway (limited to 8 meters). In this context, 38 volunteers from this study were asked to participate in the development process of this new equipment and the validation step has shown a promising degree of accuracy/precision compared to reference data.

Therefore, further studies should use these foot-worn accelerometers in addition to accelerometers in the lumbar position and on the head to assess gait patterns, and especially gait variability. In our opinion this acquisition protocol could assess the potential contribution

of these additional gait parameters in future fallers, and address the relationship between MTC values (mean, min and variables assessing the variability of the MTC).

Currently, the University of Liège offers specific dedicated structures (i.e. the laboratory analyses of human movement, LAHM) and specialised human resources with different backgrounds (medical doctors, physiotherapists, and engineers) focusing on the development of gait analysis tools and methods making it possible to use gait pattern assessment to prevent, care for and support healthy ageing. In my opinion, the results obtained and the resources available suggest further ambitious research works have to be planned in this area.

Furthermore, after follow-up, 80 people from among the 96 initially included, were assessed for anamnestic, clinical, and functional evaluation and for gait parameters (according to the acquisition data protocol used at inclusion). Based on these data, the gait parameters of people who have fallen could be compared with the gait parameters of those who have not fallen in order to better understand the consequence of the first fall event on the gait pattern.

Finally, as previously explained, this study initially considered two main outcomes, namely falls and cognitive decline, and explores two potential health-related biomarkers, namely gait performance and brain structure resources. The current work reports only the gait analysis and relationship with fall(s), but additional data remain available for analyses; namely the brain MRI acquired at inclusion and the neuropsychological assessment recorded at inclusion and at the end of the follow-up period. Below, we show the main analyses that have to be done to use available data and explore the relationship with falls and cognitive decline. First, considering data related to brain MRI, a brain MRI is available for 95 out of the 96 non fallers. Then, next steps have to consider the relationships between data related to brain imagery, those related to neuropsychological performances and gait performances, and the fall(s) incidence or/and the cognitive decline occurring during follow-up. A multivariate model is warranted, and the potential usefulness of PRoNTo (Pattern Recognition for Neuroimaging Toolbox) has to be

addressed in the aftermath of this study. PRoNTo is a user-friendly and open-source MATLAB toolbox that could make machine learning modeling available to neuroimaging data. As explained by Schrouff et al., in PRoNTo, brain scans are treated as spatial patterns and statistical learning models are used to identify statistical properties of the data that can be used to discriminate between experimental conditions or groups of subjects (classification models) or to predict a continuous measure (regression models) (Schrouff, Rosa et al. 2013). Applied to the brain MRI data acquired among our sample, PRoNTo could be used to obtain a classification model based on fall(s) during follow-up. Moreover, the brain volumes obtained using VBM analyses could be considered as additional variables to include in the classification model obtained using WEKA, in order to show how these brain volumes could help the model to better classify people as fallers or non-fallers.

In addition, considering the volunteers have been assessed for mood and cognitive performances at inclusion and after follow-up, data relative to cognitive performances at inclusion and after two years are available for univariate and multivariate analyses in terms of their relationship with gait patterns and future fall(s). Data mining resources should investigate the same relationships in a complementary and individual approach.

9. Tables and figures

This section includes the following tables:

- Table 1: Characteristics of the 105 volunteers included: qualitative variables
- Table 2: Characteristics of the 105 volunteers included: quantitative variables
- Table 3: Comparison of characteristics of the 96 volunteers followed and the 9 volunteers censored: qualitative variables
- Table 4: Comparison of characteristics of the 96 volunteers followed and the 9 volunteers censored: quantitative variables
- Table 5: Comparison of medico-surgical data according to fall incidence
- Table 6: Comparison of anamnestic data according to fall incidence
- Table 7: Comparison of clinical and functional data according to fall incidence
- Table 8: Comparison of gait parameters obtained using the accelerometric method according to fall incidence
- Table 9: Comparison of FW and DTW changes of gait parameters obtained using the accelerometric method according to fall incidence
- Table 10: Comparison of gait parameters obtained using the opto-electronic method according to fall incidence
- Table 11: Comparison of FW and DTW changes of gait parameters obtained using the opto-electronic method according to fall incidence
- Table 12: Performances of all models obtained using J48

This section includes the following figures

- Figure 3: The raw classification model obtained with WEKA: the seventh model short version
- Figure 4: The raw classification model obtained with WEKA including the gait parameters names: The seventh model short version
- Figure 5: The short version of the seventh model obtained using WEKA
- Figure 6: The raw classification model obtained with WEKA: the eighth model short version
- Figure 7: The raw classification model obtained with WEKA including the gait parameters names: The eighth model short version
- Figure 8: The short version of the eighth model obtained using WEKA
- Figure 9: The raw classification model obtained with WEKA: the ninth model short version
- Figure 10: The raw classification model obtained with WEKA including the gait parameters names: The ninth model short version
- Figure 11: The short version of the ninth model obtained using WEKA
- Figure 12: The raw classification model obtained with WEKA: the tenth model short version
- Figure 13: The raw classification model obtained with WEKA including the gait parameters names: The tenth model short version
- Figure 14: The short version of the tenth model obtained using WEKA

Table 1: Characteristics of the 105 volunteers included: Qualitative variables

Variable	Categories	Number (%)
Gender		
	Men	52 (49.5)
	Women	53 (50.5)
Marital status		
	Single	28 (26.7)
	In couple	77 (73.3)
Alcohol Consumption		
	< 4 doses / day	96 (91.4)
	≥ 4 doses / day	9 (8.6)
Number of drugs		
	< 5 drugs / day	73 (69.5)
	≥ 5 drugs / day	32 (30.5)
Sleeping Pill		
	No	65 (61.9)
	Yes	40 (38.1)
Antidepressant		
	No	91 (86.7)
	Yes	14 (13.3)
Diabetes history		
	No	84 (80.0)
	Yes	21 (20.0)
HTA history		
	No	60 (57.1)
	Yes	45 (42.9)
Stroke history		
	No	102 (97.1)
	Yes	3 (2.9)
TIA history		
	No	101 (96.2)
	Yes	4 (3.8)

Variable	Categories	Number (%)
Hip or knee prosthesis		
	No	98 (93.3)
	Yes	7 (6.7)
Memory complain		
	No	55 (52.4)
	Yes	50 (47.6)
In case of positive answer to the previous question; does exist a functional limitation linked to the memory complain		
	No	88 (83.8)
	Yes	17 (16.2)
Sleep satisfaction		
	No	71 (67.6)
	Yes	34 (32.4)
Depressive feeling		
	No	90 (85.7)
	Yes	15 (14.3)
Anxiety feeling		
	No	64 (61.0)
	Yes	41 (39.0)
Walking complain		
	No	74 (70.5)
	Yes	31 (29.5)
Fear of falling		
	No	71 (67.6)
	Yes	34 (32.4)
Regular Physiotherapeutic Cares		
	No	104 (99.0)
	Yes	1 (1.0)
Well-being feeling		
	Yes	100 (95.2)
	No	5 (4.8)

Variable	Categories	Number (%)
Quality of life satisfaction		
	Yes	95 (90.5)
	No	10 (9.5)
Feeling of dependency		
	No	93 (89.4)
	Yes	11 (10.6)
Daily-life environment and structural help need		
	House or flat without any help	92 (87.6)
	House or flat with punctual help for hard home house or gardening	11 (10.5)
	House or flat with regular help for hard home house or gardening	2 (1.9)
Frailty according to GFST		
	Not frail(s)	94 (89.5)
	Frail(s)	11 (10.5)
Visual deficit (< 3 meters)		
	No	57 (54.3)
	Yes	48 (45.7)
Timed need to realise the Timed Up and Go test (sec.)		
	< 11	93 (88.6)
	≥ 11 and < 20	12 (11.4)
	≥ 20	0
ROMBERG		
	No oscillation	77 (73.3)
	Mild and non-lateralized oscillation	28 (26.7)
	Lateralisation	0

Table 2: Characteristics of the 105 volunteers included: Quantitative variables

Variable	N	Mean	SD	SE	Min	Q1	Median	Q3	Max
Age (years)	105	71.38	5.37	0.52	65.0	67.0	70.0	75.0	89.0
Education (number of years)	105	13.04	3.52	0.34	5.0	11.0	13.0	15.0	25.0
Tobacco consumption (boxes per year)	105	8.56	13.64	1.33	0.0	0.0	0.0	15.0	60.0
EVA Pain score (0-10)	105	0.49	1.30	0.13	0.0	0.0	0.0	0.0	6.4
Physical activity score (0-7)	105	2.97	2.31	0.23	0.0	1.0	3.0	5.0	7.0
ADL (score / 24)	105	6.24	0.45	0.04	6.0	6.0	6.0	6.0	8.0
IADL (score /1)	105	0.99	0.04	0.00	0.8	1.0	1.0	1.0	1.0
GDS (score /4)	105	0.72	0.95	0.09	0.0	0.0	0.0	1.0	4.0
FES (score /64)	105	18.71	3.43	0.33	7.0	17.0	18.0	19.0	34.0
MNA (score /14)	105	12.86	1.68	0.16	7.0	12.0	14.0	14.0	14.0
CIRS-G (score /56)	105	9.45	4.45	0.43	2.0	6.0	9.0	12.0	22.0
Number of diseases scored as severity 3 or 4	105	0.73	1.11	0.11	0.0	0.0	0.0	1.0	6.0
COVI (score 3-15)	105	3.19	0.67	0.07	3.0	3.0	3.0	3.0	7.0
Edmonton (score /17)	105	2.11	1.56	0.15	0.0	1.0	2.0	3.0	7.0
Moca (score /30)	105	26.88	2.72	0.27	13.0	26.0	27.0	29.0	31.0
Moca, visuo-spatial item (score /5)	105	4.13	1.01	0.10	1.0	4.0	4.0	5.0	5.0
Moca, naming item (score /3)	105	2.80	0.49	0.05	1.0	3.0	3.0	3.0	3.0
Moca, attention item (score /6)	105	5.09	1.07	0.10	0.0	5.0	5.0	6.0	6.0

Variable	N	Mean	SD	SE	Min	Q1	Median	Q3	Max
Moca, langage item (score /3)	105	2.67	0.55	0.05	1.0	2.0	3.0	3.0	3.0
Moca, abstraction item (score /2)	105	1.81	0.42	0.04	0.0	2.0	2.0	2.0	2.0
Moca, rappel item (score /5)	105	4.02	1.10	0.11	0.0	4.0	4.0	5.0	5.0
Moca, orientation item (score /6)	105	5.88	0.41	0.04	3.0	6.0	6.0	6.0	6.0
Body weight (kg)	105	73.18	13.28	1.30	43.5	65.0	73.0	80.0	110.0
Body height (cm)	105	167.82	9.88	0.96	147.0	160.0	168.0	176.0	189.0
Waist circumference (cm)	105	94.70	11.70	1.14	65.0	87.0	93.0	102.0	121.0
Hip circumference (cm)	105	103.62	9.14	0.89	75.0	98.0	103.0	108.0	132.0
Waist/Hip ratio	105	0.91	0.08	0.01	0.8	0.8	0.9	1.0	1.1
Right leg length (cm)	105	83.37	9.85	0.96	1.0	80.5	84.0	88.0	98.0
Body Mass Index (kg/m^2)	105	25.92	3.72	0.36	18.6	23.5	25.5	28.2	36.0
Systolic blood pressure (mm Hg)	105	126.78	20.32	1.98	92.0	110.0	124.0	140.0	180.0
Diastolic blood pressure (mm Hg)	104	63.54	12.06	1.18	2.0	60.0	60.0	70.0	100.0
Pulse pressure (mm Hg)	104	63.50	19.02	1.86	30.0	51.0	60.0	74.0	168.0
Grip strength (kPa)	105	62.11	15.70	1.53	34.0	50.0	60.0	72.0	115.0
Fatigue resistance (s)	104	50.05	30.15	2.96	3.0	28.2	42.4	65.8	143.1
Grip work (kPa*s)	104	2409.06	1623.89	159.24	108.0	1218.9	1846.3	3215.7	7352.9
Grip work / Body weight ($kPa*s*kg^{-1}$)	104	32.87	21.52	2.11	2.3	16.8	26.7	42.5	99.4
Grip work / SMM ($kPa*s*kg^{-1}$)	97	90.89	56.38	5.72	8.1	47.9	76.9	124.7	298.3
Grip work / SMI (kPa*s*%)	97	66.92	42.48	4.31	3.7	37.0	53.6	90.0	217.8

Variable	N	Mean	SD	SE	Min	Q1	Median	Q3	Max
Skeletal Muscle Mass, SMM (kg)	97	26.25	5.85	0.59	13.4	21.7	25.1	30.0	44.8
Skeletal Muscle Index SMI (%)	97	35.70	5.36	0.54	23.0	32.2	35.4	38.5	56.0
Stiffness according UPDRS (score /15)	103	0.65	1.36	0.13	0.0	0.0	0.0	1.0	8.0
FGA (score / 30)	105	26.68	3.00	0.29	17.0	25.0	27.0	29.0	30.0
SPPB (score / 12)	105	10.42	1.57	0.15	6.0	10.0	11.0	12.0	12.0

Table 3: Comparison of characteristics of the 96 volunteers followed and the 9 volunteers censored: qualitative variables

Variable	Categories	N	Number (%)	N	Number (%)	Chi² p-value
Gender		9		96		
	Men		4 (44.4)		48 (50.0)	1.00
	Women		5 (55.6)		48 (50.0)	
Marital status		9		96		
	Single		4 (44.4)		24 (25.0)	0.24
	In couple		5 (55.6)		72 (75.0)	
Alcohol		9		96		
	< 4 dosis / day		9 (100.0		87 (90.6)	NA
	≥ 4 dosis / day		0 (0.0)		9 (9.4)	
Number of drugs		9		96		
	< 5 drugs / day		5 (55.6)		68 (70.8)	0.45
	≥ 5 drugs / day		4 (44.4)		28 (29.2)	
Sleep pill		9		96		
	No		5 (55.6)		60 (62.5)	0.73
	Yes		4 (44.4)		36 (37.5)	
Antidepressant		9		96		
	No		8 (88.9)		83 (86.5)	1.00
	Yes		1 (11.1)		13 (13.5)	
Diabetes history		9		96		
	No		6 (66.7)		78 (81.3)	0.38
	Yes		3 (33.3)		18 (18.8)	
High blood pressure history		9		96		
	No		7 (77.8)		53 (55.2)	0.29
	Yes		2 (22.2)		43 (44.8)	
Stroke history		9		96		
	No		7 (77.8)		95 (99.0)	0.019
	Yes		2 (22.2)		1 (1.0)	
TIA history		9		96		
	No		8 (88.9)		93 (96.9)	0.31
	Yes		1 (11.1)		3 (3.1)	
Hip or knee prothesis		9		96		
	No		9 (100.0		89 (92.7)	NA

Variable	Categories	N	Number (%)	N	Number (%)	Chi² p-value
	Yes		0 (0.0)		7 (7.3)	
Memory complain		9		96		
	No		7 (77.8)		48 (50.0)	0.17
	Yes		2 (22.2)		48 (50.0)	
In case of positive answer to the previous question; does exist a functional limitation linked to the memory complain?		9		96		
	No		9 (100.0		79 (82.3)	NA
	Yes		0 (0.0)		17 (17.7)	
Sleep satisfaction		9		96		
	No		8 (88.9)		63 (65.6)	0.27
	Yes		1 (11.1)		33 (34.4)	
Depressive feeling		9		96		
	No		8 (88.9)		82 (85.4)	1.00
	Yes		1 (11.1)		14 (14.6)	
Anxiety feeling		9		96		
	No		8 (88.9)		56 (58.3)	0.087
	Yes		1 (11.1)		40 (41.7)	
Walking complain		9		96		
	No		2 (22.2)		72 (75.0)	0.0026
	Yes		7 (77.8)		24 (25.0)	
Fear of falling		9		96		
	No		6 (66.7)		65 (67.7)	1.00
	Yes		3 (33.3)		31 (32.3)	
Regular physiotherapeutic cares		9		96		
	No		8 (88.9)		96 (100.0	1.00
	Yes		1 (11.1)		0 (0.0)	
Well-being feeling		9		96		
	Yes		9 (100.0		91 (94.8)	NA
	No		0 (0.0)		5 (5.2)	
Quality of life satisfaction		9		96		
	Yes		9 (100.0		86 (89.6)	NA
	No		0 (0.0)		10 (10.4)	

Variable	Categories	N	Number (%)	N	Number (%)	Chi² p-value
Feeling of dependency		8		96		
	No		7 (87.5)		86 (89.6)	1.00
	Yes		1 (12.5)		10 (10.4)	
Daily-life environment and structural help need		9		96		
	House or flat without any help		8 (88.9)		84 (87.5)	1.00
	House or flat with punctual help for hard home house or gardening		1 (11.1)		10 (10.4)	
	House or flat with regular help for hard home house or gardening		0 (0.0)		2 (2.1)	
Frailty according to GFST		9		96		
	Not frail(s)		8 (88.9)		86 (89.6)	1.00
	Frail(s)		1 (11.1)		10 (10.4)	
Visual deficit (3 meters)		9		96		
	No		6 (66.7)		51 (53.1)	0.50
	Yes		3 (33.3)		45 (46.9)	
Timed need to realise the Timed Up and Go test (sec.)		9		96		
	< 11		6 (66.7)		87 (90.6)	0.065
	≥ 11 and < 20		3 (33.3)		9 (9.4)	
ROMBERG		9		96		
	No oscillation		7 (77.8)		70 (72.9)	1.00
	Mild and non-lateralized oscillation		2 (22.2)		26 (27.1)	

Table 4: Comparison of characteristics of the 96 volunteers followed and the 9 volunteers censored: quantitative variables

Variable	Variable	N	Mean ± SD	Q1	Median	Q3	p-value
Age (years)	All	105	71.38 ± 5.37	67.00	70.00	75.00	
	Dropped	9	72.56 ± 5.25	68.00	73.00	75.00	0.37
	Followed	96	71.27 ± 5.40	67.00	69.50	74.50	
Education (number of years)	All	105	13.04 ± 3.52	11.00	13.00	15.00	
	Dropped	9	13.11 ± 3.33	10.00	15.00	15.00	0.86
	Followed	96	13.03 ± 3.55	11.00	13.00	15.00	
Tobacco consumption (boxes/year)	All	105	8.56 ± 13.64	0.00	0.00	15.00	
	Dropped	9	8.67 ± 15.63	0.00	0.00	11.00	0.75
	Followed	96	8.55 ± 13.53	0.00	0.00	16.50	
EVA pain (score 0-10)	All	105	0.49± 1.30	0.00	0.00	0.00	
	Dropped	**9**	**2.28 ± 2.43**	**0.00**	**1.50**	**4.70**	**<.0001**
	Followed	**96**	**0.32 ± 1.01**	**0.00**	**0.00**	**0.00**	
Physical activity (score 0-7)	All	105	2.97 ± 2.31	1.00	3.00	5.00	
	Dropped	9	2.33 ± 2.29	1.00	1.00	3.00	0.35
	Followed	96	3.03 ± 2.31	1.00	3.00	5.00	
ADL (score / 24)	All	105	6.24 ± 0.45	6.00	6.00	6.00	
	Dropped	9	6.56 ± 0.73	6.00	6.00	7.00	0.078
	Followed	96	6.21 ± 0.41	6.00	6.00	6.00	
IADL (score /1)	All	105	0.99 ± 0.04	1.00	1.00	1.00	
	Dropped	9	0.97 ± 0.07	1.00	1.00	1.00	0.079
	Followed	96	0.99 ± 0.04	1.00	1.00	1.00	
GDS (score / 4)	All	105	0.72 ± 0.95	0.00	0.00	1.00	
	Dropped	9	0.56 ± 0.73	0.00	0.00	1.00	0.72
	Followed	96	0.74 ± 0.97	0.00	0.00	1.00	
FES (score / 64)	All	105	18.71 ± 3.43	17.00	18.00	19.00	
	Dropped	9	20.89 ± 5.67	18.00	19.00	22.00	0.22

Variable	Variable	N	Mean ± SD	Q1	Median	Q3	p-value
	Followed	96	18.51 ± 3.11	17.00	18.00	19.00	
MNA (score / 14)	All	105	12.86 ± 1.68	12.00	14.00	14.00	
	Dropped	9	13.33 ± 1.12	13.00	14.00	14.00	0.43
	Followed	96	12.81 ± 1.72	12.00	14.00	14.00	
CIRS-G total (score / 56)	All	105	9.45 ± 4.45	6.00	9.00	12.00	
	Dropped	9	9.78 ± 4.35	8.00	9.00	11.00	0.78
	Followed	96	9.42 ± 4.49	6.00	9.00	12.00	
Number of diseases scored as severity 3 or 4	105	105	0.73 ± 1.11	0.00	0.00	1.00	
	Dropped	9	0.78 ± 1.09	0.00	0.00	1.00	0.82
	Followed	96	0.73 ± 1.12	0.00	0.00	1.00	
COVI (score 3-15)	All	105	3.19 ± 0.67	3.00	3.00	3.00	
	Dropped	9	3.22 ± 0.67	3.00	3.00	3.00	0.86
	Followed	96	3.19 ± 0.67	3.00	3.00	3.00	
Edmonton (score / 17)	All	105	2.11 ± 1.56	1.00	2.00	3.00	
	Dropped	9	3.44 ± 2.07	2.00	4.00	5.00	0.035
	Followed	96	1.99 ± 1.46	1.00	2.00	3.00	
MoCA total (score / 30)	All	105	26.88 ± 2.72	26.00	27.00	29.00	
	Dropped	9	26.78 ± 2.54	26.00	27.00	29.00	0.83
	Followed	96	26.89 ± 2.75	26.00	27.00	29.00	
MoCA visuo-spatial item (score /5)	All	105	4.13 ± 1.01	4.00	4.00	5.00	
	Dropped	9	3.89 ± 1.27	3.00	4.00	5.00	0.60
	Followed	96	4.16 ± 0.99	4.00	4.00	5.00	
MoCA naming item (score /3)	All	105	2.80 ± 0.49	3.00	3.00	3.00	
	Dropped	9	2.78 ± 0.67	3.00	3.00	3.00	0.75
	Followed	96	2.80 ± 0.47	3.00	3.00	3.00	
MoCA attention item (score / 6)	All	105	5.09 ± 1.07	5.00	5.00	6.00	
	Dropped	9	5.11 ± 0.93	5.00	5.00	6.00	0.98
	Followed	96	5.08 ± 1.08	5.00	5.00	6.00	

Variable	Variable	N	Mean ± SD	Q1	Median	Q3	p-value
MoCA language item (score / 3)	All	105	2.67 ± 0.55	2.00	3.00	3.00	
	Dropped	9	2.33 ± 0.50	2.00	2.00	3.00	0.019
	Followed	96	2.70 ± 0.55	2.00	3.00	3.00	
MoCA abstraction item (score / 2)	All	105	1.81 ± 0.42	2.00	2.00	2.00	
	Dropped	9	1.89 ± 0.33	2.00	2.00	2.00	0.57
	Followed	96	1.80 ± 0.43	2.00	2.00	2.00	
MoCA recall item (score / 5)	All	105	4.02 ± 1.10	4.00	4.00	5.00	
	Dropped	9	4.44 ± 0.73	4.00	5.00	5.00	0.24
	Followed	96	3.98 ± 1.12	3.50	4.00	5.00	
MoCA orientation item (score / 6)	All	105	5.88 ± 0.41	6.00	6.00	6.00	
	Dropped	9	5.89 ± 0.33	6.00	6.00	6.00	0.96
	Followed	96	5.88 ± 0.42	6.00	6.00	6.00	
Body Weight (kg)	All	105	73.18 ± 13.28	65.00	73.00	80.00	
	Dropped	9	74.67 ± 17.10	68.50	75.00	77.00	0.73
	Followed	96	73.04 ± 12.97	64.00	73.00	80.50	
Body Height (cm)	All	105	167.82 ± 9.88	160.00	168.00	176.00	
	Dropped	9	165.89 ± 11.82	156.00	163.00	168.00	0.54
	Followed	96	168.01 ± 9.73	160.00	169.00	176.50	
Waist size (cm)	All	105	94.70 ± 11.70	87.00	93.00	102.00	
	Dropped	9	94.33 ± 14.62	87.00	96.00	100.00	0.92
	Followed	96	94.73 ± 11.48	87.00	93.00	102.00	
Hip size (cm)	All	105	103.62 ± 9.14	98.00	103.00	108.00	
	Dropped	9	103.78 ± 10.13	99.00	107.00	112.00	0.96
	Followed	96	103.61 ± 9.10	98.00	102.50	108.00	
Waist/Hip ratio	All	105	0.91 ± 0.08	0.85	0.90	0.97	
	Dropped	9	0.91 ± 0.08	0.85	0.89	0.90	0.76
	Followed	96	0.91 ± 0.08	0.85	0.91	0.97	
Right leg length (cm)	All	105	84.31 ± 5.79	81.00	84.00	88.00	

Variable	Variable	N	Mean ± SD	Q1	Median	Q3	p-value
	Dropped	9	81.94 ± 6.28	78.00	81.00	85.00	0.22
	Followed	96	84.54 ± 5.73	81.00	84.50	88.00	
Systolic blood pressure (mm Hg)	All	105	126.78 ± 20.32	110.00	124.00	140.00	
	Dropped	9	121.33 ± 28.09	102.00	110.00	134.00	0.19
	Followed	96	127.29 ± 19.56	110.00	124.00	140.00	
Diastolic blood pressure (mm Hg)	All	104	63.54 ± 12.06	60.00	60.00	70.00	
	Dropped	9	65.89 ± 7.11	60.00	63.00	70.00	0.36
	Followed	95	63.32 ± 12.43	58.00	60.00	70.00	
Pulse pressure (mm Hg)	All	104	63.50 ± 19.02	51.00	60.00	74.00	
	Dropped	9	55.44 ± 23.94	38.00	48.00	74.00	0.13
	Followed	95	64.26 ± 18.46	52.00	62.00	74.00	
Grip strength (kPa)	All	105	62.11 ± 15.70	50.00	60.00	72.00	
	Dropped	9	61.33 ± 11.66	50.00	62.00	68.00	0.97
	Followed	96	62.18 ± 16.07	50.00	60.00	74.00	
Fatigue resistance (s)	All	104	50.05 ± 30.15	28.19	42.41	65.77	
	Dropped	9	55.58 ± 40.14	26.12	47.25	69.78	0.84
	Followed	95	49.52 ± 29.25	28.43	41.15	63.25	
Grip work (kPa*s)	All	104	2409.1 ± 1623.9	1218.9	1846.3	3215.7	
	Dropped	9	2686.2 ± 2101.5	1237.5	2126.3	4186.8	0.89
	Followed	95	2382.8 ± 1582.9	1200.2	1805.6	3208.1	
Grip work / Body weight ($kPa*s*Kg^{-1}$)	All	104	32.87 ± 21.52	16.82	26.72	42.51	
	Dropped	9	35.05 ± 24.19	17.76	28.45	52.88	0.73
	Followed	95	32.66 ± 21.39	16.45	26.07	42.47	
Grip work / SMM ($kPa*s*kg^{-1}$)	All	97	90.89 ± 56.38	47.91	76.90	124.66	
	Dropped	7	86.62 ± 46.93	62.32	77.19	137.71	0.85
	Followed	90	91.23 ± 57.26	47.55	75.92	124.66	
Grip work / SMI (kPa*s*%)	All	97	66.92 ± 42.48	36.95	53.60	90.02	
	Dropped	7	66.62 ± 48.47	33.58	50.76	93.64	0.98
	Followed	90	66.95 ± 42.29	36.95	54.11	90.02	

Variable	Variable	N	Mean ± SD	Q1	Median	Q3	p-value
Skeletal Muscle Mass SMM (kg)	All	97	26.25 ± 5.85	21.71	25.10	30.03	
	Dropped	7	25.91 ± 9.25	17.00	23.56	33.51	0.87
	Followed	90	26.28 ± 5.58	21.80	25.22	29.72	
Skeletal Muscle Index SMI (%)	All	97	35.70 ± 5.36	32.15	35.41	38.47	
	Dropped	7	34.77 ± 7.58	29.37	36.86	41.88	0.64
	Followed	90	35.77 ± 5.20	32.42	35.32	38.47	
Body Mass Index (kg/m^2)	All	105	25.92 ± 3.72	23.50	25.50	28.20	
	Dropped	9	26.89 ± 4.44	24.20	28.30	28.50	0.42
	Followed	96	25.83 ± 3.66	23.40	25.45	27.85	
Stiffness according UPDRS (score /15)	All	103	0.65 ± 1.36	0.00	0.00	1.00	
	Dropped	9	1.00 ± 1.73	0.00	0.00	1.00	0.57
	Followed	94	0.62 ± 1.32	0.00	0.00	1.00	
FGA (score / 30)	All	105	26.68 ± 3.00	25.00	27.00	29.00	
	Dropped	9	25.33 ± 3.74	24.00	26.00	28.00	0.19
	Followed	96	26.80 ± 2.92	26.00	27.00	29.00	
SPPB (score / 12)	All	105	10.42 ± 1.57	10.00	11.00	12.00	
	Dropped	9	10.56 ± 1.59	9.00	11.00	12.00	0.77
	Followed	96	10.41 ± 1.57	10.00	11.00	12.00	

Table 5: Comparison of medico-surgical data according to fall incidence

Variables	Categories	N-fallers (n=61) Number (%)	Fallers (n=35) Number (%)	Chi² p-value
Number of drugs / day	< 5 /day	44 (72.1)	24 (68.6)	0.71
	≥ 5 / day	17 (27.9)	11 (31.4)	
Sleeping pill	No	41 (67.2)	19 (54.3)	0.21
	Yes	20 (32.8)	16 (45.7)	
Antidepressant	No	51 (83.6)	32 (91.4)	0.36
	Yes	10 (16.4)	3 (8.6)	
Diabetes history	No	52 (85.2)	26 (74.3)	0.19
	Yes	9 (14.8)	9 (25.7)	
HTA history	No	33 (54.1)	20 (57.1)	0.77
	Yes	28 (45.9)	15 (42.9)	
Stroke history	No	60 (98.4)	35 (100.0)	1.00
	Yes	1 (1.6)	0 (0.0)	
TIA history	No	60 (98.4)	33 (94.3)	0.55
	Yes	1 (1.6)	2 (5.7)	
Hip or Knee prosthesis	No	59 (96.7)	30 (85.7)	0.095
	Yes	2 (3.3)	5 (14.3)	
Far vision deficit	No	34 (55.7)	17 (48.6)	0.50
	Yes	27 (44.03)	18 (51.4)	

Table 6: Comparison of anamnestic data according to fall incidence

Variables	Categories	N-Fallers, n=61 Number (%)	Fallers, n=35 Number (%)	Chi² p-value
Memory complain				
	No	30 (49.2)	18 (51.4)	0.83
	Yes	31 (50.8)	17 (48.6)	
Functional limitation linked with memory complain				
	No	51 (83.6)	28 (80.0)	0.66
	Yes	10 (16.4)	7 (20.0)	
Sleep satisfaction				
	No	37 (60.7)	26 (74.3)	0.18
	Yes	24 (39.3)	9 (25.7)	
Depressive feeling				
	No	52 (85.2)	30 (85.7)	0.95
	Yes	9 (14.8)	5 (14.3)	
Anxiety feeling				
	No	39 (63.9)	17 (48.6)	0.14
	Yes	22 (36.1)	18 (51.4)	
Walking complain				
	No	45 (73.8)	27 (77.1)	0.71
	Yes	16 (26.2)	8 (22.9)	
Fear of falling				
	No	42 (68.9)	23 (65.7)	0.75
	Yes	19 (31.1)	12 (34.3)	
Well-being feeling				
	Yes	57 (93.4)	34 (97.1)	0.65
	No	4 (6.6)	1 (2.9)	
Quality of life satisfaction				
	No	55 (90.2)	31 (88.6)	1.00
	Yes	6 (9.8)	4 (11.4)	
Feeling of dependency				
	No	52 (85.2)	34 (97.1)	0.088
	Yes	9 (14.8)	1 (2.9)	

Table 7: Comparison of clinical and functional data according to fall incidence

Variables	Group (n)	Mean ± SD	Q1	Median	Q3	p-value
BMI (kg/m²)	All (96)	25.83 ± 3.66	23.40	25.45	27.85	
	N-fallers (61)	26.19 ± 3.83	23.50	25.60	28.00	0.21
	Fallers (35)	25.21 ± 3.30	23.10	25.00	27.20	
Right leg length (cm)	All (96)	83.51 ± 10.13	81.00	84.00	88.00	
	N-fallers (61)	83.69 ± 11.98	83.00	86.00	88.00	0.82
	Fallers (35)	83.19 ± 5.78	78.00	84.00	87.00	
Pain score (0-10)	All (96)	0.32 ± 1.01	0.00	0.00	0.00	
	N-fallers (61)	0.24 ± 0.73	0.00	0.00	0.00	0.80
	Fallers (35)	0.46 ± 1.36	0.00	0.00	0.00	
PAS score (0-7)	All (96)	3.03 ± 2.31	1.00	3.00	5.00	
	N-fallers (61)	2.97 ± 2.47	1.00	3.00	6.00	0.46
	Fallers (35)	3.14 ± 2.05	1.00	3.00	5.00	
ADL (score /24)	All (96)	6.21 ± 0.41	6.00	6.00	6.00	
	N-fallers (61)	6.18 ± 0.39	6.00	6.00	6.00	0.37
	Fallers (35)	6.26 ± 0.44	6.00	6.00	7.00	
IADL (score /1)	All (96)	0.99 ± 0.04	1.00	1.00	1.00	
	N-fallers (61)	**1.00 ± 0.02**	**1.00**	**1.00**	**1.00**	**0.014**
	Fallers (35)	**0.98 ± 0.06**	**1.00**	**1.00**	**1.00**	
GDS (score /4)	All (96)	0.74 ± 0.97	0.00	0.00	1.00	
	N-fallers (61)	0.62 ± 0.88	0.00	0.00	1.00	0.15
	Fallers (35)	0.94 ± 1.08	0.00	1.00	2.00	
Covi (score /15)	All (96)	3.19 ± 0.67	3.00	3.00	3.00	
	N-fallers (61)	3.13 ± 0.64	3.00	3.00	3.00	0.057
	Fallers (35)	3.29 ± 0.71	3.00	3.00	3.00	
FES (score /64)	All (96)	18.51 ± 3.11	17.00	18.00	19.00	
	N-fallers (61)	18.23 ± 2.61	17.00	18.00	19.00	0.073
	Fallers (35)	19.00 ± 3.82	17.00	19.00	20.00	
MNA (score /14)	All (96)	12.81 ± 1.72	12.00	14.00	14.00	
	N-fallers (61)	12.90 ± 1.72	12.00	14.00	14.00	0.36
	Fallers (35)	12.66 ± 1.73	12.00	13.00	14.00	
Edmonton (/17)	All (96)	1.99 ± 1.46	1.00	2.00	3.00	
	N-fallers (61)	1.84 ± 1.46	1.00	2.00	3.00	0.14
	Fallers (35)	2.26 ± 1.44	1.00	2.00	3.00	
MoCA total (score /30)	All (96)	26.89 ± 2.75	26.00	27.00	29.00	

	N-fallers (61)	27.18 ± 2.00	26.00	28.00	29.00	0.62
	Fallers (35)	26.37 ± 3.69	26.00	27.00	29.00	
MoCA visuo-spatial item (score /5)	All (96)	4.16 ± 0.99	4.00	4.00	5.00	
	N-fallers (61)	4.18 ± 0.99	4.00	4.00	5.00	0.72
	Fallers (35)	4.11 ± 0.99	3.00	4.00	5.00	
MoCA naming item (score /3)	All (96)	2.80 ± 0.47	3.00	3.00	3.00	
	N-fallers (61)	2.85 ± 0.40	3.00	3.00	3.00	0.20
	Fallers (35)	2.71 ± 0.57	3.00	3.00	3.00	
MoCA attention item (score /6)	All (96)	5.08 ± 1.08	5.00	5.00	6.00	
	N-fallers (61)	5.16 ± 0.88	5.00	5.00	6.00	0.81
	Fallers (35)	4.94 ± 1.37	5.00	5.00	6.00	
MoCA language item (score /3)	All (96)	2.70 ± 0.55	2.00	3.00	3.00	
	N-fallers (61)	2.77 ± 0.46	3.00	3.00	3.00	0.13
	Fallers (35)	2.57 ± 0.65	2.00	3.00	3.00	
MoCA abstraction item (score /2)	All (96)	1.80 ± 0.43	2.00	2.00	2.00	
	N-fallers (61)	1.84 ± 0.37	2.00	2.00	2.00	0.41
	Fallers (35)	1.74 ± 0.51	2.00	2.00	2.00	
MoCA recall item (score /5)	All (96)	3.98 ± 1.12	3.50	4.00	5.00	
	N-fallers (61)	4.02 ± 0.96	3.00	4.00	5.00	0.71
	Fallers (35)	3.91 ± 1.38	4.00	4.00	5.00	
MoCA orientation item (score /6)	All (96)	5.88 ± 0.42	6.00	6.00	6.00	
	N-fallers (61)	5.87 ± 0.46	6.00	6.00	6.00	0.83
	Fallers (35)	5.89 ± 0.32	6.00	6.00	6.00	
Grip strength (kPa)	All (96)	62.18 ± 16.07	50.00	60.00	74.00	
	N-fallers (61)	63.77 ± 15.62	52.00	60.00	74.00	0.12
	Fallers (35)	59.40 ± 16.69	48.00	54.00	74.00	
Fatigue resistance (s)	All (96)	49.52 ± 29.25	28.43	41.15	63.25	
	N-fallers (61)	51.61 ± 27.78	30.28	44.42	68.55	0.27
	Fallers (35)	45.93 ± 31.71	23.47	39.97	60.44	
Grip work (kPa x s)	All (95)	2382.8 ± 1582.9	1200.2	1805.6	3208.1	
	N-fallers (60)	2529.8 ± 1537.2	1397.0	1874.7	3559.4	0.24
	Fallers (35)	2130.9 ± 1650.3	774.51	1802.3	2910.6	

Grip work/ body weight (kPa x s/kg)	All (95)	32.66 ± 21.39	16.45	26.07	42.47	
	N-fallers (60)	34.30 ± 20.56	19.17	26.54	44.97	0.21
	Fallers (35)	29.85 ± 22.76	12.55	26.07	42.47	
Grip work / SMM (kPa x s/kg)	All (95)	91.23 ± 57.26	47.55	75.92	124.66	
	N-fallers (60)	91.69 ± 50.18	55.38	74.90	126.43	0.51
	Fallers (35)	90.42 ± 68.64	41.13	76.90	116.76	
Grip work / SMI (kPa x s/%)	All (95)	66.95 ± 42.29	36.95	54.11	90.02	
	N-fallers (60)	68.25 ± 38.28	37.69	54.63	90.16	0.46
	Fallers (35)	64.69 ± 49.00	34.71	53.60	82.54	
Skeletal Muscle Mass, SSM (kg)	All (90)	26.28 ± 5.58	21.80	25.22	29.72	
	N-fallers (57)	27.02 ± 5.63	22.67	25.75	31.01	0.095
	Fallers (33)	24.99 ± 5.34	21.51	24.87	28.23	
Skeletal Muscle Index, SMI (%)	All (90)	35.77 ± 5.20	32.42	35.32	38.47	
	N-fallers (57)	36.26 ± 5.62	33.05	35.75	38.90	0.25
	Fallers (33)	34.93 ± 4.34	32.10	34.41	37.30	
Stiffness according UPDRS (/15)	All (94)	0.62 ± 1.32	0.00	0.00	1.00	
	N-fallers (60)	**0.45 ± 1.25**	**0.00**	**0.00**	**0.00**	**0.043**
	Fallers (34)	**0.91 ± 1.40**	**0.00**	**0.00**	**2.00**	
FGA (score /30)	All (96)	26.80 ± 2.92	26.00	27.00	29.00	
	N-fallers (61)	27.15 ± 2.59	26.00	28.00	29.00	0.24
	Fallers (35)	26.20 ± 3.37	24.00	27.00	29.00	
SPPB (score /12)	All (96)	10.41 ± 1.57	10.00	11.00	12.00	
	N-fallers (61)	**10.70 ± 1.42**	**10.00**	**11.00**	**12.00**	**0.015**
	Fallers (35)	**9.89 ± 1.71**	**9.00**	**10.00**	**11.00**	

Table 8: Comparison of gait parameters obtained using the accelerometric method according to fall incidence

Variables	Groups	Mean ± SD	Q1	Median	Q3	p-value
CW Gait Speed (m/s)	All (96)	1.29 ± 0.18	1.17	1.28	1.41	
	N-fallers (61)	1.31 ± 0.18	1.20	1.28	1.43	0.078
	Fallers (35)	1.24 ± 0.18	1.12	1.25	1.40	
FW Gait Speed (m/s)	All (96)	1.70 ± 0.23	1.52	1.72	1.86	
	N-fallers (61)	**1.74 ± 0.22**	**1.59**	**1.75**	**1.87**	**0.035**
	Fallers (35)	**1.64 ± 0.24**	**1.43**	**1.68**	**1.85**	
DTW Gait Speed (m/s)	All (95)	1.16 ± 0.22	1.02	1.16	1.32	
	N-fallers (60)	1.17 ± 0.24	1.01	1.16	1.33	0.86
	Fallers (35)	1.16 ± 0.19	1.02	1.14	1.32	
CW Stride Length (m)	All (96)	1.34 ± 0.16	1.23	1.37	1.45	
	N-fallers (61)	**1.37 ± 0.15**	**1.26**	**1.38**	**1.48**	**0.035**
	Fallers (35)	**1.30 ± 0.17**	**1.16**	**1.27**	**1.45**	
FW Stride Length (m)	All (96)	1.55 ± 0.24	1.38	1.58	1.71	
	N-fallers (61)	**1.60 ± 0.24**	**1.41**	**1.62**	**1.73**	**0.010**
	Fallers (35)	**1.47 ± 0.23**	**1.28**	**1.42**	**1.66**	
DTW Stride Length (m)	All (95)	1.29 ± 0.25	1.15	1.28	1.43	
	N-fallers (60)	1.31 ± 0.28	1.15	1.28	1.44	0.31
	Fallers (35)	1.26 ± 0.18	1.10	1.27	1.37	
CW Stride Frequency (Stride/s)	All (96)	0.96 ± 0.07	0.93	0.93	1.03	
	N-fallers (61)	0.96 ± 0.07	0.93	0.93	0.98	0.91
	Fallers (35)	0.96 ± 0.08	0.88	0.98	1.03	
FW Stride Frequency (Stride/s)	All (96)	1.09 ± 0.10	1.03	1.07	1.15	
	N-fallers (61)	1.08 ± 0.11	1.03	1.07	1.17	0.39
	Fallers (35)	1.10 ± 0.08	1.03	1.07	1.12	
DTW Stride Frequency (Stride/s)	All (95)	0.91 ± 0.10	0.88	0.93	0.98	

	N-fallers (60)	0.89 ± 0.10	0.83	0.93	0.96	0.14
	Fallers (35)	0.93 ± 0.10	0.88	0.93	1.03	
CW Regularity (dimensionless)	All (96)	304.01 ± 47.24	269.00	310.00	340.00	
	N-fallers (61)	305.28 ± 46.98	271.00	313.00	338.00	0.73
	Fallers (35)	301.80 ± 48.29	262.00	301.00	341.00	
FW Regularity (dimensionless)	All (96)	307.15 ± 53.62	275.00	310.00	346.50	
	N-fallers (61)	311.33 ± 51.87	276.00	311.00	353.00	0.32
	Fallers (35)	299.86 ± 56.55	273.00	310.00	340.00	
DTW Regularity (dimensionless)	All (94)	254.34 ± 62.40	218.00	259.50	300.00	
	N-fallers (59)	248.80 ± 63.71	209.00	253.00	290.00	0.27
	Fallers (35)	263.69 ± 59.85	228.00	265.00	307.00	
CW Symmetry (dimensionless)	All (96)	213.69 ±63.86	178.50	201.50	239.00	
	N-fallers (61)	208.90 ± 59.84	178.00	208.00	239.00	0.64
	Fallers (35)	222.03 ± 70.46	181.00	201.00	239.00	
FW Symmetry (dimensionless)	All (96)	213.73 ± 53.46	180.00	205.50	248.50	
	N-fallers (61)	218.34 ± 54.51	184.00	205.00	253.00	0.27
	Fallers (35)	205.69 ± 51.37	174.00	206.00	223.00	
DTW Symmetry (dimensionless)	All (95)	213.97 ± 64.66	166.00	202.00	243.00	
	N-fallers (60)	224.42 ± 72.55	170.00	208.00	275.50	0.11
	Fallers (35)	196.06 ± 43.60	162.00	197.00	223.00	

Table 9: Comparison of FW and DTW changes of gait parameters obtained using the accelerometric method according to fall incidence

Variables	Group (n)	Mean ± SD	Q1	Median	Q3	P-value
Gait speed DTW cost (%)	All (n=95)	9.76 ± 11.73	1.69	7.69	16.54	
	Non-fallers (n=60)	11.42 ± 12.55	2.43	8.45	18.85	0.14
	Fallers (n=35)	6.92 ± 9.68	0.83	6.90	13.55	
Gait speed FW improvement (%)	All (n=96)	32.84 ± 11.93	26.30	32.22	38.91	
	Non-fallers (n=61)	33.21 ± 10.91	27.97	32.87	38.53	0.69
	Fallers (n=35)	32.20 ± 13.68	22.90	31.69	42.15	
Stride length DTW cost (%)	All (n=96)	4.80 ± 15.40	-0.67	4.41	8.68	
	Non-fallers (n=61)	5.87 ± 18.56	-0.69	4.65	11.76	0.37
	Fallers (n=35)	2.94 ± 7.00	-0.66	4.17	7.38	
Stride length FW improvement (%)	All (n=96)	15.91 ± 12.40	10.62	14.95	19.78	
	Non-fallers (n=61)	17.36 ± 14.20	12.20	16.34	19.84	0.13
	Fallers (n=35)	13.39 ± 7.96	8.51	13.82	18.84	
Stride frequency DTW cost (%)	All (n=95)	5.56 ± 8.88	0.00	5.10	10.20	
	Non-fallers (n=60)	7.12 ± 9.07	0.00	5.24	10.20	0.11
	Fallers (n=35)	2.88 ± 7.98	0.00	4.85	8.04	
Stride frequency FW improvement (%)	All (n=96)	13.90 ± 10.63	9.26	15.05	19.39	
	Non-fallers (n=61)	13.30 ± 11.70	10.75	15.05	18.45	0.93
	Fallers (n=35)	14.96 ± 8.52	9.18	14.29	20.43	
Regularity DTW cost (%)	All (n=96)	17.96 ± 20.34	5.95	16.28	27.25	
	Non-fallers (n=61)	21.01 ± 22.70	5.60	18.79	29.54	0.052
	Fallers (n=35)	12.64 ± 14.16	6.31	12.10	19.57	
Regularity FW improvement (%)	All (n=96)	1.56 ± 13.63	-7.15	1.45	11.15	
	Non-fallers (n=61)	2.67 ± 13.89	-7.40	3.22	12.62	0.29
	Fallers (n=35)	-0.37 ± 13.14	-6.91	0.71	9.64	
Symmetry DTW cost (%)	All (n=95)	-5.93 ± 36.62	-28.11	1.43	22.31	
	Non-fallers (n=60)	**-13.61 ± 40.54**	**-41.56**	**-11.83**	**18.44**	**0.022**
	Fallers (n=35)	**7.24 ± 23.94**	**-15.42**	**7.32**	**26.07**	
Symmetry FW improvement (%)	All (n=96)	5.58 ± 31.39	-13.36	2.25	24.47	
	Non-fallers (n=61)	10.39 ± 32.89	-8.76	10.21	32.11	0.059
	Fallers (n=35)	-2.80 ± 27.04	-20.79	-6.63	16.78	

Table 10: Comparison of gait parameters obtained using the opto-electronic method according to fall incidence

Variables	Groups (n)	Mean ± SD	Q1	Median	Q3	p-value
CW Mean MTC (mm)	All (n=88)	17.60 ± 5.32	14.38	17.04	20.87	
	Non-fallers (n= 55)	17.76 ± 5.16	14.30	17.20	20.85	0.71
	Fallers (n = 33)	17.32 ± 5.66	14.54	16.64	20.89	
FW Mean MTC (mm)	All (n=87)	20.19 ± 6.84	15.45	19.54	24.14	
	Non-fallers (n= 54)	20.73 ± 7.23	16.21	20.61	24.14	0.35
	Fallers (n = 30)	19.32 ± 6.15	14.78	17.98	22.93	
DTW Mean MTC (mm)	All (n=84)	15.20 ± 5.52	11.68	14.24	17.83	
	Non-fallers (n= 54)	15.31 ± 5.50	11.83	14.16	17.65	0.82
	Fallers (n = 30)	15.01 ± 5.65	11.45	15.28	17.83	
CW Med MTC (mm)	All (n=88)	17.57 ± 5.42	14.09	16.88	21.17	
	Non-fallers (n= 55)	17.70 ± 5.20	14.35	16.65	21.05	0.77
	Fallers (n = 33)	17.35 ± 5.85	13.47	16.93	21.29	
FW Med. MTC (mm)	All (n=87)	20.00 ± 6.96	15.13	20.10	24.50	
	Non-fallers (n= 54)	20.78 ± 7.35	16.01	20.22	25.26	0.18
	Fallers (n = 30)	18.72 ± 6.18	13.00	18.18	22.81	
DTW Med. MTC (mm)	All (n=84)	15.09 ± 5.63	11.60	14.40	18.02	
	Non-fallers (n= 54)	15.27 ± 5.58	11.54	14.40	17.36	0.70
	Fallers (n = 30)	14.77 ± 5.80	11.71	14.55	20.00	
CW SD MTC (mm)	All (n=88)	4.44 ± 1.69	3.06	4.38	5.39	
	Non-fallers (n= 55)	4.31 ± 1.78	2.83	4.12	5.42	0.21
	Fallers (n = 33)	4.66 ± 1.54	3.32	4.99	5.35	
FW SD MTC (mm)	All (n=87)	5.00 ± 2.67	3.31	4.49	6.62	
	Non-fallers (n= 54)	5.00 ± 2.79	3.21	4.52	6.47	0.74
	Fallers (n = 30)	4.99 ± 2.49	3.37	4.46	7.07	
DTW SD MTC (mm)	All (n=84)	4.18 ± 1.93	2.71	3.89	5.26	
	Non-fallers (n= 54)	4.18 ± 1.86	2.72	4.12	5.20	0.77
	Fallers (n = 30)	4.17 ± 2.07	2.68	3.59	6.13	
CW IQR MTC (mm)	All (n=88)	6.23 ± 2.92	3.86	5.52	8.06	
	Non-fallers (n= 55)	6.06 ± 2.99	3.60	4.73	8.32	0.36
	Fallers (n = 33)	6.50 ± 2.83	4.99	5.67	7.71	
FW IQR MTC (mm)	All (n=87)	6.99 ± 4.12	4.15	5.95	9.03	
	Non-fallers (n= 54)	7.25 ± 4.40	4.17	6.43	9.61	0.51
	Fallers (n = 30)	6.56 ± 3.65	4.15	5.81	8.80	
DTW IQR MTC (mm)	All (n=84)	5.72 ± 2.91	3.40	5.40	7.22	

Variables	Groups (n)	Mean ± SD	Q1	Median	Q3	p-value
	Non-fallers (n= 54)	5.63 ± 2.75	3.29	5.44	7.11	0.92
	Fallers (n = 30)	5.89 ± 3.22	3.55	5.31	7.80	
CW Var. MTC (mm^2)	All (n=88)	22.57 ± 17.44	9.39	19.16	29.00	
	Non-fallers (n= 55)	21.71 ± 18.66	8.03	16.94	29.39	0.21
	Fallers (n = 33)	23.98 ± 15.35	11.04	24.89	28.61	
FW Var. MTC (mm^2)	All (n=87)	21.01 ± 19.02	7.30	15.27	27.98	
	Non-fallers (n= 54)	21.35 ± 18.92	8.89	17.31	27.39	0.47
	Fallers (n = 30)	20.46 ± 19.46	5.86	12.66	27.98	
DTW Var. MTC (mm^2)	All (n=84)	32.78 ± 37.92	11.40	21.94	44.98	
	Non-fallers (n= 54)	34.22 ± 43.15	11.44	22.17	43.79	0.90
	Fallers (n = 30)	30.19 ± 26.48	11.37	21.59	50.05	
CW COV MTC (%)	All (n=88)	27.18 ± 12.51	18.21	24.84	32.55	
	Non-fallers (n= 55)	25.69 ± 11.62	17.17	24.33	30.94	0.16
	Fallers (n = 33)	29.65 ± 13.69	20.13	27.00	34.07	
FW COV MTC (%)	All (n=87)	29.83 ± 16.17	18.18	26.75	36.96	
	Non-fallers (n= 54)	29.28 ± 13.91	18.23	27.59	35.00	0.90
	Fallers (n = 30)	30.72 ± 19.51	16.79	26.14	39.19	
DTW COV. MTC (%)	All (n=84)	28.25 ± 17.01	14.93	25.28	35.71	
	Non-fallers (n= 54)	28.58 ± 17.69	15.05	24.92	38.67	0.98
	Fallers (n = 30)	27.66 ± 15.98	12.84	27.10	35.59	
CW Min MTC (mm)	All (n=88)	11.89 ± 5.43	8.42	11.65	15.15	
	Non-fallers (n= 55)	12.42 ± 5.49	9.31	11.97	16.09	0.24
	Fallers (n = 33)	11.01 ± 5.29	8.36	10.75	14.48	
FW Min. MTC (mm)	All (n=87)	9.96 ± 5.20	6.88	9.48	12.20	
	Non-fallers (n= 54)	10.06 ± 5.22	6.88	9.13	11.69	0.87
	Fallers (n = 30)	9.80 ± 5.24	6.93	9.88	12.20	
DTW Min. MTC (mm)	All (n=84)	13.97 ± 7.12	8.69	13.50	18.80	
	Non-fallers (n= 54)	14.15 ± 7.37	8.47	13.96	18.62	0.76
	Fallers (n = 30)	13.65 ± 6.77	9.56	12.13	19.86	
CW Delta1 MTC (mm)	All (n=88)	6.16 ± 3.05	3.88	5.33	7.79	
	Non-fallers (n= 55)	6.01 ± 3.33	3.42	5.16	7.25	0.20
	Fallers (n = 33)	6.41 ± 2.55	4.60	6.00	8.26	
FW Delta1 MTC (mm)	All (n=87)	5.82 ± 3.43	3.46	5.05	7.69	
	Non-fallers (n= 54)	5.93 ± 3.38	3.82	5.15	7.83	0.52
	Fallers (n = 30)	5.64 ± 3.55	3.11	4.95	6.91	
DTW Delta1 MTC (mm)	All (n=84)	6.59 ± 3.70	3.69	6.18	8.37	
	Non-fallers (n= 54)	6.47 ± 3.44	3.73	5.84	8.12	0.99

Variables	Groups (n)	Mean ± SD	Q1	Median	Q3	p-value
	Fallers (n = 30)	6.80 ± 4.17	3.51	6.30	9.28	
CW Delta2 MTC (mm)	All (n=88)	5.70 ± 2.52	3.72	5.26	6.84	
	Non-fallers (n= 55)	5.34 ± 2.38	3.47	4.88	6.56	0.089
	Fallers (n = 33)	6.31 ± 2.67	4.05	6.16	7.97	
FW Delta2 MTC (mm)	All (n=87)	5.39 ± 2.50	3.45	5.02	7.50	
	Non-fallers (n= 54)	5.42 ± 2.53	3.45	5.02	7.53	0.81
	Fallers (n = 30)	5.35 ± 2.48	3.47	5.02	7.40	
DTW Delta2 MTC (mm)	All (n=84)	6.08 ± 3.32	3.85	5.65	7.31	
	Non-fallers (n= 54)	6.15 ± 3.50	3.66	5.70	7.27	0.99
	Fallers (n = 30)	5.94 ± 3.01	4.22	5.55	7.36	

Table 11: Comparison of FW and DTW changes of gait parameters obtained using the opto-electronic method according to fall incidence

Variables	Groups (n)	Mean ± SD	Q1	Median	Q3	P-value
Mean MTC FW improvement (%)	All (n=87)	16.52 ± 30.31	-1.03	13.66	27.33	
	Non-fallers (n=54)	16.47 ± 26.24	0.78	15.38	28.16	0.42
	Fallers (n=33)	16.61 ± 36.44	-1.03	9.78	22.02	
Mean MTC DTW cost (%)	All (n=83)	12.75 ± 16.03	3.26	15.08	22.42	
	Non-fallers (n=53)	12.14 ± 16.72	-1.13	14.60	22.42	0.61
	Fallers (n=30)	13.83 ± 14.96	5.32	15.99	21.37	
Min MTC FW improvement (%)	All (n=87)	-4.14 ± 77.69	-39.98	-17.47	6.44	
	Non-fallers (n=54)	-8.52 ± 54.10	-41.67	-17.99	12.74	0.73
	Fallers (n=33)	3.03 ± 106.25	-38.20	-10.66	-2.29	
Min MTC DTW cost (%)	All (n=83)	-35.62 ± 96.77	-57.70	-13.20	9.68	
	Non-fallers (n=53)	-23.65 ± 63.45	-50.33	-11.12	9.68	0.32
	Fallers (n=30)	-56.78 ± 136.15	-60.58	-22.51	6.58	
COV MTC FW improvement (%)	All (n=87)	18.79 ± 61.32	-26.24	6.75	58.41	
	Non-fallers (n=54)	25.67 ± 63.37	-21.71	15.91	64.49	0.12
	Fallers (n=33)	7.52 ± 56.94	-29.92	-5.82	29.89	
COV MTC DTW cost (%)	All (n=83)	-10.12 ± 62.81	-39.82	1.87	39.89	
	Non-fallers (n=53)	-18.30 ± 69.96	-54.98	-3.71	39.89	0.28
	Fallers (n=30)	4.33 ± 45.17	-14.78	3.14	27.21	

Table 12: Performances of all models obtained using J48

Models	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Comfort W = Model 1	56%	3%	87%	11%	61%	0.42	0.49
Fast W = Model 2	70%	29%	93%	71%	70%	0.52	0.56
Dual Task W = = Model 3	62%	17%	88%	46%	65%	0.60	0.62
FW improvement = Model 4	58%	3%	90%	14%	62%	0.42	0.49
DTW cost = Model 5	59%	34%	74%	43%	66%	0.57	0.60
FW + DTW cost = Model 6 full version	73%	37%	93%	76%	72%	0.67	0.71
Model 6 = short version	76%	22%	92%	77%	76%	0.64	0.65
All gait parameters = Model 7 full version	61%	54%	66%	48%	71%	0.66	0.66
Model 7 short version	80%	69%	87%	75%	83%	0.79	0.77
All gait parameters+ Age +gender+ leg length = Model 8 full version	61%	57%	54%	47%	72%	0.66	0.65
Model 8 short version	76%	51%	90%	75%	76%	0.76	0.76
All gait parameters +IADL+ Stiffness +SPPB = Model 9 full version	59%	46%	67%	44%	68%	0.57	0.59
Model 9 short version	82%	69%	90%	80%	83%	0.80	0.79
All gait parameters +Age + gender + leg length + IADL +Stiffness +SPPB = Model 10 full version	69%	46%	67%	4%	68%	0.58	0.59
Model 10 short version	84%	80%	87%	78%	88%	0.84	0.83

Figure 3: The raw seventh model short version

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-13,15-17,19-20,22,25-26,28-29,32-33,35-38-weka.filters.unsupervised.attribute.Remove-R10,12-22-weka.filters.unsupervised.attribute.Remove-R12-14,16-23

Instances: 96

Attributes: 14

x14

x18

x21

x23

x24

x27

x30

x31

x34

x40

x52

x56

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

----

x31 <= -35.981308: 0 (17.18)

x31 > -35.981308

| x14 <= 1.29: 1 (11.0)

| x14 > 1.29

| | x40 <= 3.3: 0 (12.96/0.38)

| | x40 > 3.3

| | | x21 <= -4.672897: 1 (6.91/0.1)

| | | x21 > -4.672897

| | | | x18 <= 0.88

| | | | | x27 <= 12.684366: 1 (8.81/1.0)

| | | | | x27 > 12.684366: 0 (3.53)

| | | | x18 > 0.88

| | | | | x30 <= 188: 0 (12.81)

| | | | | x30 > 188

| | | | | | x30 <= 239

| | | | | | | x21 <= 0: 0 (6.0/1.0)

| | | | | | | x21 > 0

| | | | | | | | x21 <= 8.035714: 1 (7.0)

| | | | | | | | x21 > 8.035714: 0 (4.0/1.0)

| | | | | | x30 > 239: 0 (5.81)

Number of Leaves : 11

Size of the tree : 21

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	77	80.2083 %
Incorrectly Classified Instances	19	19.7917 %
Kappa statistic	0.5649	
Mean absolute error	0.2342	
Root mean squared error	0.4251	
Relative absolute error	50.4204 %	
Root relative squared error	88.2471 %	
Total Number of Instances	96	

=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,869	0,314	0,828	0,869	0,848	0,566	0,787	0,818	0
	0,686	0,131	0,750	0,686	0,716	0,566	0,787	0,674	1
Weighted Avg.	0,802	0,248	0,800	0,802	0,800	0,566	0,787	0,766	

=== Confusion Matrix ===

a	b	<-- classified as
53	8	a = 0
11	24	b = 1

Figure 4: The raw seventh model short version including the gait parameters

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-13,15-17,19-20,22,25-26,28-29,32-33,35-38-weka.filters.unsupervised.attribute.Remove-R10,12-22-weka.filters.unsupervised.attribute.Remove-R12-14,16-23

Instances: 96

Attributes: 14

x14

x18

x21

x23

x24

x27

x30

x31

x34

x40

x52

x56

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

Symmetry DTW cost <= -35.981308: 0 (17.18)

Symmetry DTW cost > -35.981308

| FW Stride Length <= 1.29: 1 (11.0)

| FW Stride Length > 1.29

| | CW Delta 1 MTC <= 3.3: 0 (12.96/0.38)

| | CW Delta 1 MTC > 3.3

| | | Stride Frequency DTW cost <= -4.672897: 1 (6.91/0.1)

| | | Stride Frequency DTW cost > -4.672897

| | | | CW Stride Frequency <= 0.88

| | | | | Regularity FW improvement <= 12.684366: 1 (8.81/1.0)

| | | | | Regularity FW improvement > 12.684366: 0 (3.53)

| | | | CW Stride Frequency > 0.88

| | | | | DTW Symmetry <= 188: 0 (12.81)

| | | | | DTW Symmetry > 188

| | | | | | DTW Symmetry <= 239

| | | | | | | Stride Frequency DTW cost <= 0: 0 (6.0/1.0)

| | | | | | | Stride Frequency DTW cost > 0

| | | | | | | | Stride Frequency DTW cost <= 8.035714: 1 (7.0)

| | | | | | | | Stride Frequency DTW cost > 8.035714: 0 (4.0/1.0)

| | | | | | DTW Symmetry > 239: 0 (5.81)

Number of Leaves : 11

Size of the tree : 21

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	77	80.2083 %
Incorrectly Classified Instances	19	19.7917 %
Kappa statistic	0.5649	
Mean absolute error	0.2342	
Root mean squared error	0.4251	
Relative absolute error	50.4204 %	
Root relative squared error	88.2471 %	
Total Number of Instances	96	

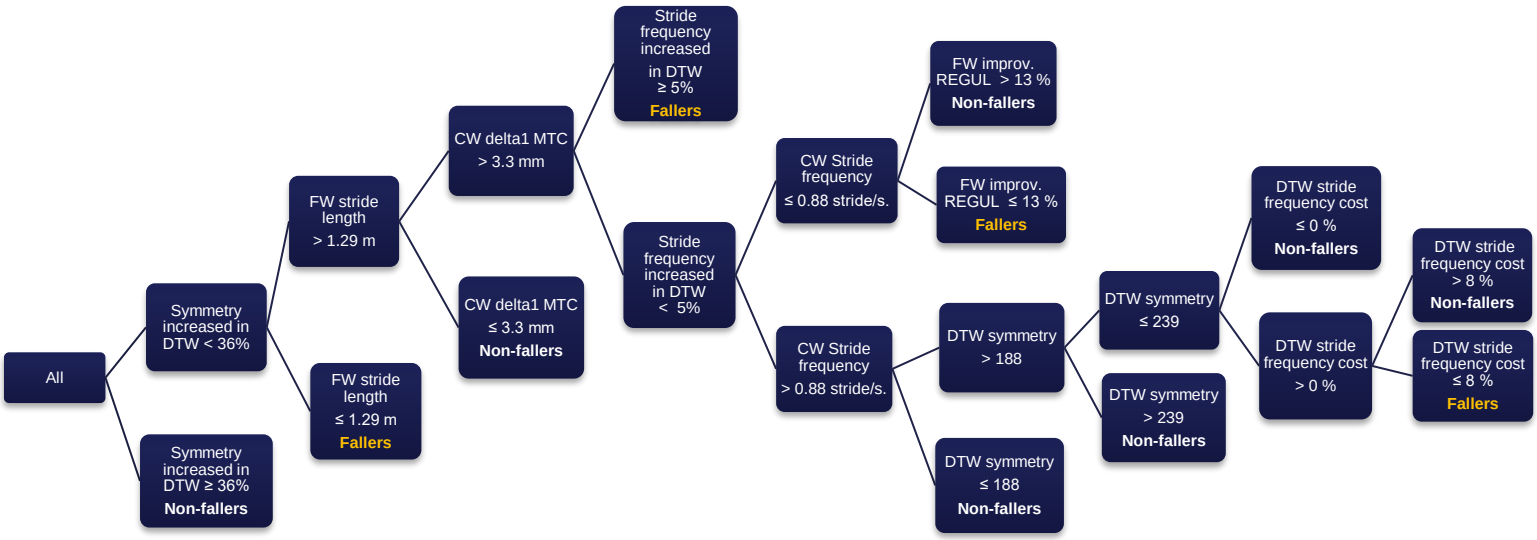
=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,869	0,314	0,828	0,869	0,848	0,566	0,787	0,818	0
	0,686	0,131	0,750	0,686	0,716	0,566	0,787	0,674	1
Weighted Avg.	0,802	0,248	0,800	0,802	0,800	0,566	0,787	0,766	

=== Confusion Matrix ===

a	b	<-- classified as
53	8	a = 0
11	24	b = 1

Figure 5: The short version of the seventh model



Fallers, n=35	True positive = 24	False negative =11
Non-fallers, n=61	True negative = 53	False positive = 8

Model	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 7 " Short version"	**80%**	**69%**	**87%**	**75%**	**83%**	**0.79**	**0.77**

Figure 6: The raw eighth model short version

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-2,4-8,10-12,15-17,19-20-weka.filters.unsupervised.attribute.Remove-R7-11,13-14,17-23-weka.filters.unsupervised.attribute.Remove-R10-11,14-23-weka.filters.unsupervised.attribute.Remove-R13-23

Instances: 96

Attributes: 14

x3

x9

x13

x14

x18

x21

x27

x30

x31

x41

x42

x53

x65

class

Test mode: 10-fold cross-validation

J48 pruned tree

x31 <= -35.981308: 0 (17.18)

x31 > -35.981308

| x14 <= 1.29: 1 (11.0)

| x14 > 1.29

| | x21 <= -4.672897

| | | x65 <= -66.8: 0 (2.56/0.56)

| | | x65 > -66.8: 1 (6.55/0.11)

| | x21 > -4.672897

| | | x18 <= 0.88

| | | | x27 <= 12.684366

| | | | | x31 <= 32.720588: 1 (8.0)

| | | | | x31 > 32.720588: 0 (2.0)

| | | | x27 > 12.684366: 0 (4.71)

| | | x18 > 0.88

| | | | x9 <= 1.75: 0 (17.0)

| | | | x9 > 1.75

| | | | | x30 <= 188: 0 (10.0)

| | | | | x30 > 188

| | | | | | x65 <= -26.4: 0 (3.4/0.2)

| | | | | | x65 > -26.4

| | | | | | | x21 <= 9.708738

| | | | | | | | x53 <= 4

| | | | | | | | | x13 <= 1.45: 0 (2.8)

| | | | | | | | | x13 > 1.45: 1 (3.0)

| | | | | | | | x53 > 4: 1 (5.8)

| | | | | | | x21 > 9.708738: 0 (2.0)

Number of Leaves : 14

Size of the tree : 27

Time taken to build model: 0.03 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	73	76.0417 %
Incorrectly Classified Instances	23	23.9583 %
Kappa statistic	0.4458	
Mean absolute error	0.2643	
Root mean squared error	0.4489	
Relative absolute error	56.9031 %	
Root relative squared error	93.1991 %	
Total Number of Instances	96	

=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,902	0,486	0,764	0,902	0,827	0,462	0,763	0,797	0
	0,514	0,098	0,750	0,514	0,610	0,462	0,763	0,696	1
Weighted Avg.	0,760	0,344	0,759	0,760	0,748	0,462	0,763	0,760	

=== Confusion Matrix ===

a	b	<-- classified as
55	6	a = 0
17	18	b = 1

Figure 7: The raw eighth model short version including gait parameters

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-2,4-8,10-12,15-17,19-20-weka.filters.unsupervised.attribute.Remove-R7-11,13-14,17-23-weka.filters.unsupervised.attribute.Remove-R10-11,14-23-weka.filters.unsupervised.attribute.Remove-R13-23

Instances: 96

Attributes: 14

x3

x9

x13

x14

x18

x21

x27

x30

x31

x41

x42

x53

x65

class

Test mode: 10-fold cross-validation

J48 pruned tree

Symmetry DTW cost <= -35.981308: 0 (17.18)

Symmetry DTW cost > -35.981308

| FW Stride Length <= 1.29: 1 (11.0)

| FW Stride Length > 1.29

| | Stride Frequency DTW cost <= -4.672897

| | | CoV MTC DTW cost <= -66.8: 0 (2.56/0.56)

| | | CoV MTC DTW cost > -66.8: 1 (6.55/0.11)

| | Stride Frequency DTW cost > -4.672897

| | | CW Stride Frequency <= 0.88

| | | | Regularity FW improvement <= 12.684366

| | | | | Symmetry DTW cost <= 32.720588: 1 (8.0)

| | | | | Symmetry DTW cost > 32.720588: 0 (2.0)

| | | | Regularity FW improvement > 12.684366: 0 (4.71)

| | | CW Stride Frequency > 0.88

| | | | FW Gait Speed <= 1.75: 0 (17.0)

| | | | FW Gait Speed > 1.75

| | | | | DTW Symmetry <= 188: 0 (10.0)

| | | | | DTW Symmetry > 188

| | | | | | CoV MTC DTW cost <= -26.4: 0 (3.4/0.2)

| | | | | | CoV MTC DTW cost > -26.4

| | | | | | | Stride Frequency DTW cost <= 9.708738

| | | | | | | | FW SD MTC <= 4

| | | | | | | | | CW Stride Length <= 1.45: 0 (2.8)

| | | | | | | | | CW Stride Length > 1.45: 1 (3.0)

| | | | | | | | FW SD MTC > 4: 1 (5.8)

| | | | | | | Stride Frequency DTW cost > 9.708738: 0 (2.0)

Number of Leaves : 14

Size of the tree : 27

Time taken to build model: 0.03 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	73	76.0417 %
Incorrectly Classified Instances	23	23.9583 %
Kappa statistic	0.4458	
Mean absolute error	0.2643	
Root mean squared error	0.4489	
Relative absolute error	56.9031 %	
Root relative squared error	93.1991 %	
Total Number of Instances	96	

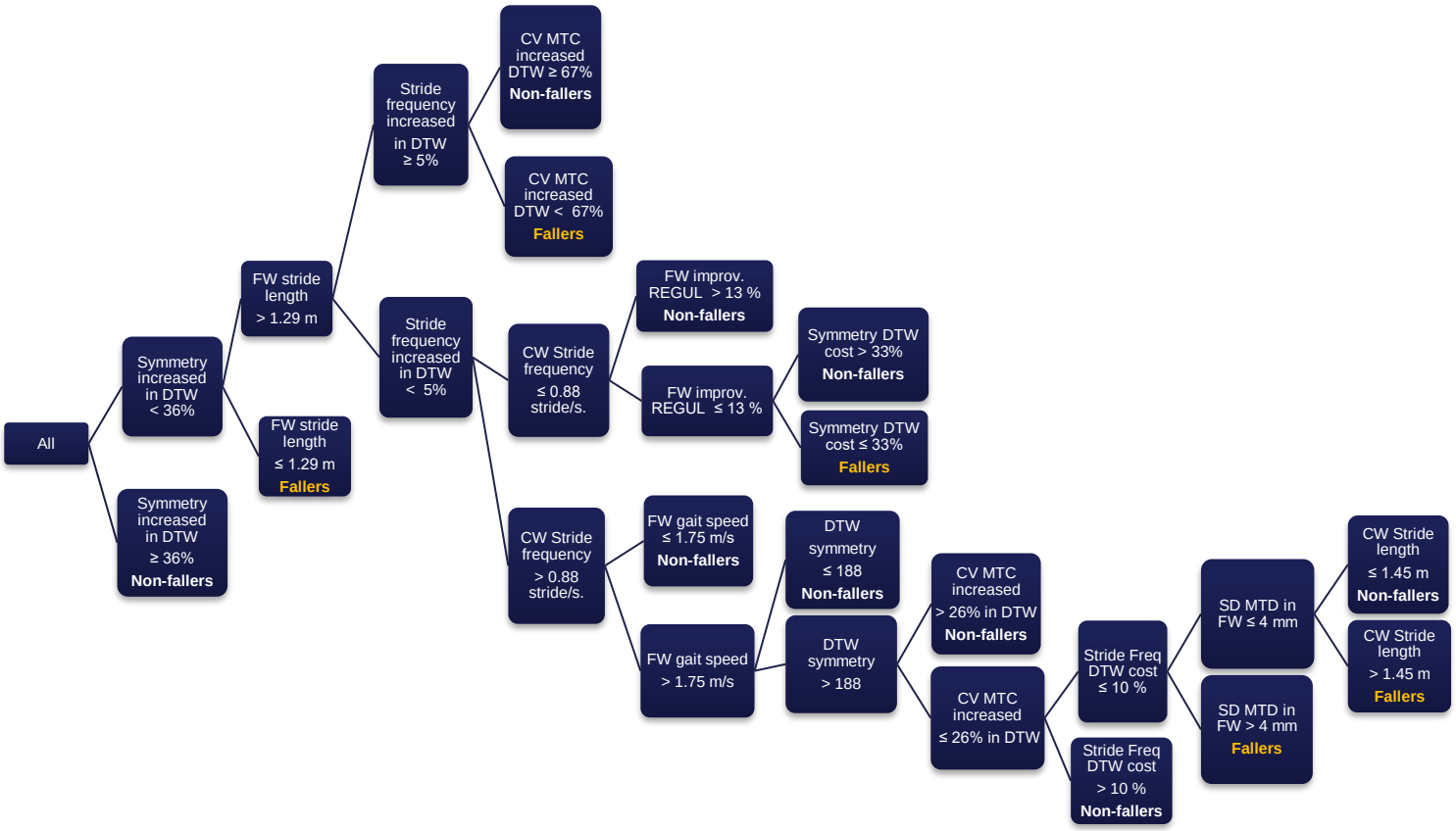
=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,902	0,486	0,764	0,902	0,827	0,462	0,763	0,797	0
	0,514	0,098	0,750	0,514	0,610	0,462	0,763	0,696	1
Weighted Avg.	0,760	0,344	0,759	0,760	0,748	0,462	0,763	0,760	

=== Confusion Matrix ===

a	b	<-- classified as
55	6	a = 0
17	18	b = 1

Figure 8: The short version of the eighth model



Fallers, n=35	True positive = 18	False negative =17
Non-fallers, n=61	True negative = 55	False positive = 6

Model	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 8 short version	**76%**	**51%**	**90%**	**75%**	**76%**	**0.76**	**0.76**

Figure 9: The raw ninth model short version

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-5,8-12,15-17,19-21-weka.filters.unsupervised.attribute.Remove-R7-10,12-13,16,18-19,21-22-weka.filters.unsupervised.attribute.Remove-R13-14,16-26,29-34-weka.filters.unsupervised.attribute.Remove-R16

Instances: 96

Attributes: 19

x6

x7

x13

x14

x18

x22

x27

x30

x31

x33

x36

x39

x42

x54

x55

x63

x64

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

x31 <= -35.981308: 0 (17.18)

x31 > -35.981308

| x14 <= 1.29: 1 (11.0)

| x14 > 1.29

| | x6 <= 1

| | | x65 <= -26.4: 0 (17.2/0.63)

| | | x65 > -26.4

| | | | x55 <= 15.6

| | | | | x27 <= 8.518519: 1 (14.4/2.36)

| | | | | x27 > 8.518519: 0 (5.24/1.0)

| | | | x55 > 15.6: 0 (17.98/2.33)

| | x6 > 1

| | | x33 <= 17.3: 0 (5.42/0.42)

| | | x33 > 17.3: 1 (7.58)

Number of Leaves : 8

Size of the tree : 15

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	79	82.2917 %
Incorrectly Classified Instances	17	17.7083 %
Kappa statistic	0.6058	
Mean absolute error	0.2388	
Root mean squared error	0.4049	
Relative absolute error	51.4195 %	
Root relative squared error	84.0564 %	
Total Number of Instances	96	

=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,902	0,314	0,833	0,902	0,866	0,610	0,801	0,821	0
	0,686	0,098	0,800	0,686	0,738	0,610	0,801	0,739	1
Weighted Avg.	0,823	0,236	0,821	0,823	0,820	0,610	0,801	0,791	

=== Confusion Matrix ===

a	b	<-- classified as
55	6	a = 0
11	24	b = 1

Figure 10: The raw ninth model short version including the gait parameters

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-5,8-12,15-17,19-21-weka.filters.unsupervised.attribute.Remove-R7-10,12-13,16,18-19,21-22-weka.filters.unsupervised.attribute.Remove-R13-14,16-26,29-34-weka.filters.unsupervised.attribute.Remove-R16

Instances: 96

Attributes: 19

x6

x7

x13

x14

x18

x22

x27

x30

x31

x33

x36

x39

x42

x54

x55

x63

x64

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

Symmetry DTW cost <= -35.981308: 0 (17.18)

Symmetry DTW cost > -35.981308

| FW Stride Length <= 1.29: 1 (11.0)

| FW Stride Length > 1.29

| | Stiffness according UPDRS <= 1

| | | CoV MTC DTW cost <= -26.4: 0 (17.2/0.63)

| | | CoV MTC DTW cost > -26.4

| | | | FW Var. MTC <= 15.6

| | | | | Regularity FW improvement <= 8.518519: 1 (14.4/2.36)

| | | | | Regularity FW improvement > 8.518519: 0 (5.24/1.0)

| | | | FW Var. MTC > 15.6: 0 (17.98/2.33)

| | Stiffness according UPDRS > 1

| | | CW Mean MTC <= 17.3: 0 (5.42/0.42)

| | | CW Mean MTC > 17.3: 1 (7.58)

Number of Leaves : 8

Size of the tree : 15

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	79	82.2917 %
Incorrectly Classified Instances	17	17.7083 %
Kappa statistic	0.6058	
Mean absolute error	0.2388	
Root mean squared error	0.4049	
Relative absolute error	51.4195 %	
Root relative squared error	84.0564 %	
Total Number of Instances	96	

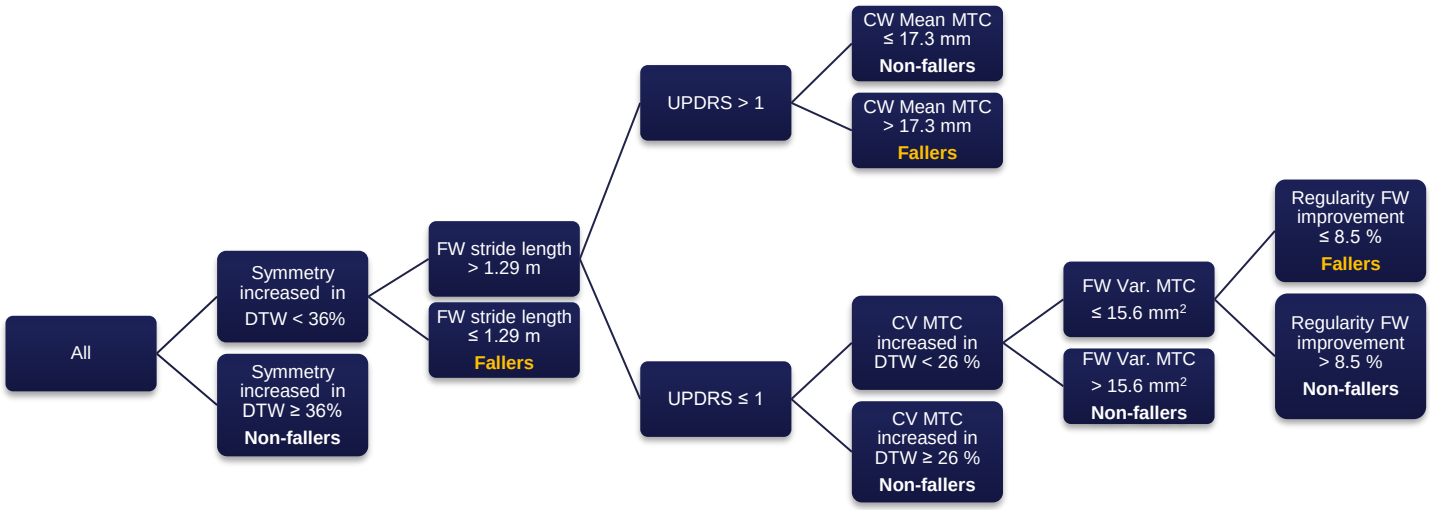
=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,902	0,314	0,833	0,902	0,866	0,610	0,801	0,821	0
	0,686	0,098	0,800	0,686	0,738	0,610	0,801	0,739	1
Weighted Avg.	0,823	0,236	0,821	0,823	0,820	0,610	0,801	0,791	

=== Confusion Matrix ===

a	b	<-- classified as
55	6	a = 0
11	24	b = 1

Figure 11: The short version of the ninth model



Fallers, n=35	True positive = 24	False negative =11
Non-fallers, n=61	True negative = 55	False positive = 6

Model	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
Model 9 short version	82%	69%	90%	80%	83%	0.80	0.79

Figure 12: The raw tenth model short version

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-2,4-5,7-12,15-23-weka.filters.unsupervised.attribute.Remove-R5-10,13,15-23-weka.filters.unsupervised.attribute.Remove-R8-13,15,17-19,21-23-weka.filters.unsupervised.attribute.Remove-R11-14

Instances: 96

Attributes: 14

x3

x6

x13

x14

x30

x31

x33

x49

x51

x55

x63

x64

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

x31 <= -35.981308: 0 (17.18)

x31 > -35.981308

| x14 <= 1.29: 1 (11.0)

| x14 > 1.29

| | x6 <= 1

| | | x65 <= -26.4: 0 (17.2/0.63)

| | | x65 > -26.4

| | | | x55 <= 15.6

| | | | | x51 <= 18.5: 1 (8.29/0.45)

| | | | | x51 > 18.5

| | | | | | x3 <= 0

| | | | | | | x31 <= 22.310757: 1 (4.21/0.21)

| | | | | | | x31 > 22.310757: 0 (2.73/0.21)

| | | | | | x3 > 0: 0 (4.41/1.0)

| | | | x55 > 15.6

| | | | | x49 <= 11.6: 0 (15.63/0.29)

| | | | | x49 > 11.6: 1 (2.35/0.31)

| | x6 > 1

| | | x33 <= 17.3: 0 (5.42/0.42)

| | | x33 > 17.3: 1 (7.58)

Number of Leaves : 11

Size of the tree : 21

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	81	84.375 %
Incorrectly Classified Instances	15	15.625 %
Kappa statistic	0.6648	
Mean absolute error	0.2054	
Root mean squared error	0.3842	
Relative absolute error	44.2357 %	
Root relative squared error	79.7538 %	
Total Number of Instances	96	

=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,869	0,200	0,883	0,869	0,876	0,665	0,839	0,855	0
	0,800	0,131	0,778	0,800	0,789	0,665	0,839	0,791	1
Weighted Avg.	0,844	0,175	0,845	0,844	0,844	0,665	0,839	0,831	

=== Confusion Matrix ===

a	b	<-- classified as
53	8	a = 0
7	28	b = 1

Figure 13: The raw tenth model including the gait parameters

Scheme: weka.classifiers.trees.J48 -C 0.25 -M 2

Relation: T0marchecomplet_sex_ageNaN_120-weka.filters.unsupervised.attribute.Remove-R1-2,4-5,7-12,15-23-weka.filters.unsupervised.attribute.Remove-R5-10,13,15-23-weka.filters.unsupervised.attribute.Remove-R8-13,15,17-19,21-23-weka.filters.unsupervised.attribute.Remove-R11-14

Instances: 96

Attributes: 14

x3

x6

x13

x14

x30

x31

x33

x49

x51

x55

x63

x64

x65

class

Test mode: 10-fold cross-validation

=== Classifier model (full training set) ===

J48 pruned tree

Symmetry DTW cost <= -35.981308: 0 (17.18)

Symmetry DTW cost > -35.981308

| FW Stride Length <= 1.29: 1 (11.0)

| FW Stride Length > 1.29

| | Stiffness according UPDRS <= 1

| | | CoV MTC DTW cost <= -26.4: 0 (17.2/0.63)

| | | CoV MTC DTW cost > -26.4

| | | | FW Var. MTC <= 15.6

| | | | | FW Mean MTC <= 18.5: 1 (8.29/0.45)

| | | | | FW Mean MTC > 18.5

| | | | | | Men

| | | | | | | Symmetry DTW cost <= 22.310757: 1 (4.21/0.21)

| | | | | | | Symmetry DTW cost > 22.310757: 0 (2.73/0.21)

| | | | | | Women: 0 (4.41/1.0)

| | | | FW Var. MTC > 15.6

| | | | | DTW Delta 1 MTC <= 11.6: 0 (15.63/0.29)

| | | | | DTW Delta 1 MTC > 11.6: 1 (2.35/0.31)

| | Stiffness according UPDRS > 1

| | | CW Mean MTC <= 17.3: 0 (5.42/0.42)

| | | CW Mean MTC > 17.3: 1 (7.58)

Number of Leaves : 11

Size of the tree : 21

Time taken to build model: 0.02 seconds

=== Stratified cross-validation ===

=== Summary ===

Correctly Classified Instances	81	84.375 %
Incorrectly Classified Instances	15	15.625 %
Kappa statistic	0.6648	
Mean absolute error	0.2054	
Root mean squared error	0.3842	
Relative absolute error	44.2357 %	
Root relative squared error	79.7538 %	
Total Number of Instances	96	

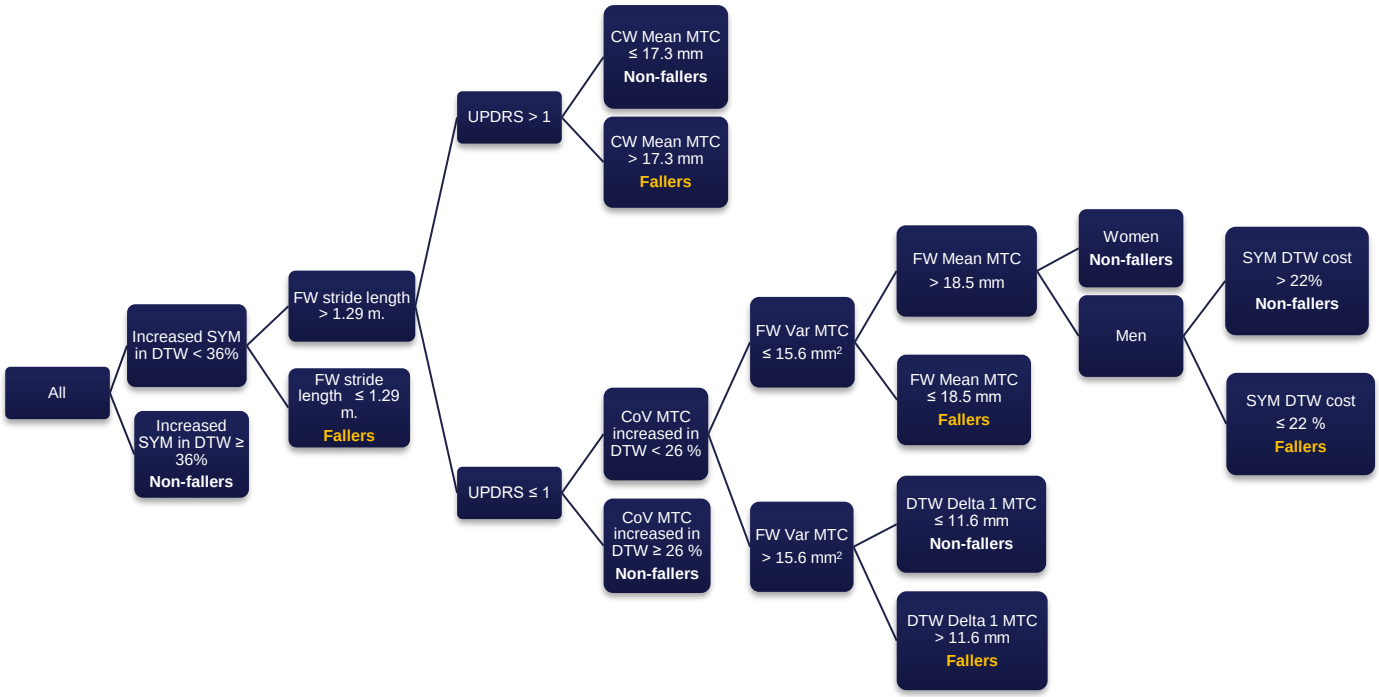
=== Detailed Accuracy By Class ===

	TP Rate	FP Rate	Precision	Recall	F-Measure	MCC	ROC Area	PRC Area	Class
	0,869	0,200	0,883	0,869	0,876	0,665	0,839	0,855	0
	0,800	0,131	0,778	0,800	0,789	0,665	0,839	0,791	1
Weighted Avg.	0,844	0,175	0,845	0,844	0,844	0,665	0,839	0,831	

=== Confusion Matrix ===

a	b	<-- classified as
53	8	a = 0
7	28	b = 1

Figure 14: The short version of the tenth model



Fallers, n= 35	True positive = 28	False negative = 7
Non-fallers, n= 61	True negative = 53	False positive = 8

Model	Accuracy	Sensitivity	Specificity	PPV	NPV	ROC Area	PRC Area
All gait parameters + Age + gender + leg length +IADL+SPPB+UPDRS Model 10 "short version"	84%	80%	87%	78%	88%	0.84	0.83

10. References

Abernethy, M. (2010). "Data mining with WEKA, Part 1: Introduction and regression." Retrieved March, 2018, from https://www.ibm.com/developerworks/library/os-weka1/.

Abram, W., M. Beer, R. Berkow, A. Fletcher and M. Chir (1995). The Merck Manual of Geriatrics**:** 65–78.

Al-Yahya, E., H. Dawes, L. Smith, A. Dennis, K. Howells and J. Cockburn (2011). "Cognitive motor interference while walking: A systematic review and meta-analysis." Neuroscience & Biobehavioral Reviews **35**(3): 715-728.

Alan, R. D. A. and G. B. John (2014). "The effects of walking speed on minimum toe clearance and on the temporal relationship between minimum clearance and peak swing-foot velocity in unilateral trans-tibial amputees." Prosthetics and Orthotics International **39**(2): 120-125.

Allali, G., E. I. Ayers and J. Verghese (2016). "Motoric Cognitive Risk Syndrome Subtypes and Cognitive Profiles." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **71**(3): 378-384.

Allali, G., R. W. Kressig, F. Assal, F. R. Herrmann, V. Dubost and O. Beauchet (2007). "Changes in gait while backward counting in demented older adults with frontal lobe dysfunction." Gait & Posture **26**(4): 572-576.

Allali, G., C. P. Launay, H. M. Blumen, M. L. Callisaya, A.-M. De Cock, R. W. Kressig, V. Srikanth, J.-P. Steinmetz, J. Verghese and O. Beauchet (2017). "Falls, Cognitive Impairment, and Gait Performance: Results From the GOOD Initiative." Journal of the American Medical Directors Association **18**(4): 335-340.

Ambrose, A. F., G. Paul and J. M. Hausdorff (2013). "Risk factors for falls among older adults: A review of the literature." Maturitas **75**(1): 51-61.

Annweiler, C., O. Beauchet, R. Bartha, M. Montero-Odasso and W. T.-W. g. A.-L. f. Knowledge (2013). "Slow gait in MCI is associated with ventricular enlargement: results from the Gait and Brain Study." J Neural Transm **120**(7): 1083-1092.

Annweiler, C., O. Beauchet, R. Bartha, J. L. Wells, M. J. Borrie, V. Hachinski and M. Montero-Odasso (2013). "Motor cortex and gait in mild cognitive impairment: a magnetic resonance spectroscopy and volumetric imaging study." Brain **136**(3): 859-871.

Annweiler, C. and M. Montero-Odasso (2012). "Vascular burden as a substrate for higher-level gait disorders in older adults. A review of brain mapping literature." Panminerva Med **54**(3): 189-204.

Annweiler, C., M. Montero-Odasso, R. Bartha, J. Drozd, V. Hachinski and O. Beauchet (2014). "Association between gait variability and brain ventricle attributes: a brain mapping study." Experimental Gerontology **57**: 256-263.

Armstrong, R. A. (2014). "When to use the Bonferroni correction." Ophthalmic and Physiological Optics **34**(5): 502-508.

Artaud, F., A. Singh-Manoux, A. Dugravot, C. Tzourio and A. Elbaz (2015). "Decline in Fast Gait Speed as a Predictor of Disability in Older Adults." Journal of the American Geriatrics Society **63**(6): 1129-1136.

Auvinet, B., G. Berrut, C. Touzard, L. Moutel, N. Collet, D. Chaleil and E. Barrey (2002). "Falls in the Elderly: the Need for Teamwork Through a Network." Rev Med Ass Maladie **33**.

Auvinet, B., G. Berrut, C. Touzard, L. Moutel, N. Collet, D. Chaleil and E. Barrey (2002). "Reference data for normal subjects obtained with an accelerometric device." Gait & Posture **16**(2): 124-134.

Auvinet, B., G. Berrut, C. Touzard, L. Moutel, N. Collet, D. Chaleil and E. Barrey (2003). "Gait abnormalities in elderly fallers." Journal of Aging and Physical Activity **11**(1): 40-52.

Barak, Y., R. C. Wagenaar and K. G. Holt (2006). "Gait Characteristics of Elderly People With a History of Falls: A Dynamic Approach." Physical Therapy **86**(11): 1501-1510.

Barrett, R. S., P. M. Mills and R. K. Begg (2010). "A systematic review of the effect of ageing and falls history on minimum foot clearance characteristics during level walking." Gait & Posture **32**(4): 429-435.

Bassey, E. J., M. A. Fiatarone, E. F. O'neill, M. Kelly, W. J. Evans and L. A. Lipsitz (1992). "Leg extensor power and functional performance in very old men and women." Clinical Science **82**(3): 321-327.

Bautmans, I., B. Jansen, B. Van Keymolen and T. Mets (2011). "Reliability and clinical correlates of 3D-accelerometry based gait analysis outcomes according to age and fall-risk." Gait Posture **33**(3): 366-372.

Bautmans, I., K. Van Puyvelde and T. Mets (2009). "Sarcopenia and functional decline: pathophysiology, prevention an therapy." Acta Clinica Belgica **64**(4): 303-316.

Bautmans I., M. (2005). "A fatigue resistance test for elderly persons based on grip strength: reliability and comparison with healthy young subjects." Aging Clinical and Experimental Research **17**(3): 217-222

Beauchet, O., G. Allali, C. Annweiler, G. Berrut, N. Maarouf, F. R. Herrmann and V. Dubost (2008). "Does Change in Gait while Counting Backward Predict the Occurrence of a First Fall in Older Adults?" Gerontology **54**(4): 217-223.

Beauchet, O., G. Allali, L. Poujol, J. C. Barthelemy, F. Roche and C. Annweiler (2010). "Decrease in gait variability while counting backward: a marker of "magnet effect"?" J Neural Transm **117**(10): 1171-1176.

Beauchet, O., G. Allali, H. Sekhon, J. Verghese, S. Gillain, J.-P. Steinmetz, R. W. Kressig, J. M. Barden, T. Szturm, C. P. Launay, S. Grenier, L. Bherer, T. Liu-Ambrose, V. L. Chester, M. L. Callisaya, V. Srikanth, G. Léonard, A.-M. De Cock, R. Sawa, G. Duque, R. Camicioli and J. L. Helbostad (2017). "Guidelines for Assessment of Gait and Reference Values for Spatiotemporal Gait Parameters in Older Adults: The Biomathics and Canadian Gait Consortiums Initiative." Frontiers in Human Neuroscience **11**.

Beauchet, O., C. Annweiler, V. Dubost, G. Allali, R. W. Kressig, S. Bridenbaugh, G. Berrut, F. Assal and F. R. Herrmann (2009). "Stops walking when talking: a predictor of falls in older adults?" European Journal of Neurology **16**(7): 786-795.

Beauchet, O. and G. Berrut (2006). "Marche et double tâche : définition, intérêts et perspectives chez le sujet âgé." Psychologie & NeuroPsychiatrie du vieillissement **4**(3): 215-225.

Beaudart, C., D. Maquet, M. Mannarino, F. Buckinx, M. Demonceau, J.-M. Crielaard, J.-Y. Reginster and O. Bruyère (2013). "Effects of 3 months of short sessions of controlled whole body vibrations on the risk of falls among nursing home residents." BMC Geriatrics **13**: 42-42.

Blake, A. J., K. Morgan, M. J. Bendall, H. Dallosso, S. B. Ebrahim, T. H. Arie, P. H. Fentem and E. J. Bassey (1988). "Falls by elderly people at home: Prevalence and associated factors." Age Ageing **17**.

Bloch, F., M. Thibaud, B. Dugué, C. Brèque, A. S. Rigaud and G. Kemoun (2010). "Episodes of falling among elderly people: a systematic review and meta-analysis of social and demographic predisposing characteristics." Clinics **65**(9): 895-903.

Bloch, F., M. Thibaud, C. Tournoux-Facon, C. Brèque, A.-S. Rigaud, B. Dugué and G. Kemoun (2013). "Estimation of the risk factors for falls in the elderly: Can meta-analysis provide a valid answer?" Geriatrics & Gerontology International **13**(2): 250-263.

Bloem, B. R., J. Gussekloo, A. M. Lagaay, E. J. Remarque, J. Haan and R. G. J. Westendorp (2000). "Idiopathic Senile Gait Disorders Are Signs of Subclinical Disease." Journal of the American Geriatrics Society **48**(9): 1098-1101.

Bloem, B. R., V. V. Valkenburg, M. Slabbekoorn and M. D. Willemsen (2001). "The Multiple Tasks Test: Development and normal strategies." Gait & Posture **14**(3): 191-202.

Bohannon, R. W. (1997). "Comfortable and maximum walking speed of adults aged 20-79 years: reference values and determinants." Age Ageing **26**: 15-19.

Bohannon, R. W., A. W. Andrews and M. W. Thomas (1996). "Walking Speed: Reference Values and Correlates for Older Adults." Journal of Orthopaedic & Sports Physical Therapy **24**(2): 86-90.

Bonnet, V., S. Ramdani, C. Azevedo-Coste, P. Fraisse, C. Mazzà and A. Cappozzo (2014). "Integration of Human Walking Gyroscopic Data Using Empirical Mode Decomposition." Sensors **14**(1): 370.

Bootsma-van der Wiel, A., J. Gussekloo, A. J. M. De Craen, E. Van Exel, B. R. Bloem and R. G. J. Westendorp (2003). "Walking and Talking as Predictors of Falls in the General Population: The Leiden 85-Plus Study." Journal of the American Geriatrics Society **51**(10): 1466-1471.

Boutaayamou, M., C. Schwartz, J. Stamatakis, V. Denoël, D. Maquet, B. Forthomme, J.-L. Croisier, B. Macq, J. G. Verly, G. Garraux and O. Brüls (2015). "Development and validation of an accelerometer-based method for quantifying gait events." Medical Engineering & Physics **37**(2): 226-232.

Brach, J. S., S. Perera, S. Studenski, M. Katz, C. Hall and J. Verghese (2010). "Meaningful change in measures of gait variability in older adults." Gait & Posture **31**(2): 175-179.

Brach, J. S., S. Studenski, S. Perera, J. M. VanSwearingen and A. B. Newman (2008). "Stance time and step width variability have unique contributing impairments in older persons." Gait & Posture **27**(3): 431-439.

Brach, J. S., S. A. Studenski, S. Perera, J. M. VanSwearingen and A. B. Newman (2007). "Gait variability and the risk of incident mobility disability in community-dwelling older adults." The journals of gerontology. Series A, Biological sciences and medical sciences **62**(9): 983-988.

Bridenbaugh, S. A. and R. W. Kressig (2011). "Laboratory Review: The Role of Gait Analysis in Seniors' Mobility and Fall Prevention." Gerontology **57**(3): 256-264.

Brodie, M. A. D., H. B. Menz and S. R. Lord (2014). "Age-associated changes in head jerk while walking reveal altered dynamic stability in older people." Experimental Brain Research **232**(1): 51-60.

Brodie, M. A. D., M. Psarakis and P. Hoang (2016). "Gyroscopic corrections improve wearable sensor data prior to measuring dynamic sway in the gait of people with Multiple Sclerosis." Computer Methods in Biomechanics and Biomedical Engineering **19**(12): 1339-1346.

Buckinx F, B. C., Slomian J., Maquet D., Demonceau M., Gillain S., Petermans J., Reginster J. Y., Bruyère O. (2015). "Added value of a triaxial accelerometer assessing gait parameters to predict falls and mortality among nursing home residents: a two-year prospective study." Technology and Health Care **23**: 195-203.

Buckinx, F., C. Beaudart, D. Maquet, M. Demonceau, J. M. Crielaard, J. Y. Reginster and O. Bruyère (2014). "Evaluation of the impact of 6-month training by whole body vibration on the risk of falls among nursing home residents, observed over a 12-month period: a single blind, randomized controlled trial." Aging Clinical and Experimental Research **26**(4): 369-376.

Callisaya, M. L., L. Blizzard, J. L. McGinley and V. K. Srikanth (2012). "Risk of falls in older people during fast-walking – The TASCOG study." Gait & Posture **36**(3): 510-515.

Cappello, A., R. Stagni, S. Fantozzi and A. Leardini (2005). "Soft tissue artifact compensation in knee kinematics by double anatomical landmark calibration: performance of a novel method during selected motor tasks." IEEE Transactions on Biomedical Engineering **52**(6): 992-998.

Casellato, L. (1994). "New Developments in Visual Acuity Charts." Ophthalmologica **208**(2): 95-97

Chiari, L., U. D. Croce, A. Leardini and A. Cappozzo (2005). "Human movement analysis using stereophotogrammetry: Part 2: Instrumental errors." Gait & Posture **21**(2): 197-211.

Cho, C. Y. and G. Kamen (1998). "Detecting Balance Deficits in Frequent Fallers Using Clinical and Quantitative Evaluation Tools." Journal of the American Geriatrics Society **46**(4): 426-430.

Choi, K. and Y. Ko (2015). "Characteristics Associated With Fear of Falling and Activity Restriction in South Korean Older Adults." Journal of Aging and Health **27**(6): 1066-1083.

Clément, J. P., R F Nassif, J M Léger, and F. Marchan (1997). "Development and Contribution to the Validation of a Brief French Version of the Yesavage Geriatric Depression Scale." L'encéphale **23**(2): 91-99.

Craig, J., A. Murray, S. Mitchell, L. Clark, L. Saunders and L. Burleigh (2013). "The high cost to health and social care of managing falls in older adults living in the community in Scotland." Scottish Medical Journal **58**(4): 198-203.

Cruz-Jentoft, A. J., J. P. Baeyens, J. M. Bauer, Y. Boirie, T. Cederholm, F. Landi, F. C. Martin, J. P. Michel, Y. Rolland, S. M. Schneider, E. Topinkova, M. Vandewoude, M. Zamboni and P. European Working Group on Sarcopenia in Older (2010). "Sarcopenia: European consensus on definition and

diagnosis: Report of the European Working Group on Sarcopenia in Older People." Age Ageing **39**(4): 412-423.

Cui, X., C. K. Peng, M. D. Costa, A. Weiss, A. L. Goldberger and J. M. Hausdorff (2014). "Development of a new approach to quantifying stepping stability using ensemble empirical mode decomposition." Gait & Posture **39**(1): 495-500.

D'Ath, P., Katona, P, Mullan, E, Evans, S, and Katona, C (1994). "Screening, Detection and Management of Depression in Elderly Primary Care Attenders. I: The Acceptability and Performance of the 15 Item Geriatric Depression Scale (GDS15) and the Development of Short Versions." Family Practice **11**(3): 260-266.

Davis Jesse, G. M. and (2006). The Relationship Between Precision-Recall and ROC Curves. Proceedings of the 23 rd International Conference on Machine Learning. Pittsburgh.

de Bruin, E. D. and A. Schmidt (2010). "Walking behaviour of healthy elderly: attention should be paid." Behavioral and Brain Functions **6**(1): 1-8.

Delbaere, K., J. C. T. Close, H. Brodaty, P. Sachdev and S. R. Lord (2010). "Determinants of disparities between perceived and physiological risk of falling among elderly people: cohort study." The BMJ **341**: c4165.

Delbaere, K., J. C. T. Close, A. S. Mikolaizak, P. S. Sachdev, H. Brodaty and S. R. Lord (2010). "The Falls Efficacy Scale International (FES-I). A comprehensive longitudinal validation study." Age and Ageing **39**(2): 210-216.

Delbaere, K., G. Crombez, G. Vanderstraeten, T. Willems and D. Cambier (2004). "Fear-related avoidance of activities, falls and physical frailty. A prospective community-based cohort study." Age and Ageing **33**(4): 368-373.

Della Croce, U., A. Leardini, L. Chiari and A. Cappozzo (2005). "Human movement analysis using stereophotogrammetry: Part 4: assessment of anatomical landmark misplacement and its effects on joint kinematics." Gait & Posture **21**(2): 226-237.

Deshpande, N., E. J. Metter, S. Bandinelli, J. Guralnik and L. Ferrucci (2009). "Gait speed under varied challenges and cognitive decline in older persons: a prospective study." Age and Ageing **38**(5): 509-514.

Diener, H. C. and J. Dichgans (1988). Chapter 22 On the role of vestibular, visual and somatosensory information for dynamic postural control in humans. Progress in Brain Research. O. Pompeiano and J. H. J. Allum, Elsevier. **76:** 253-262.

Feder, G., C. Cryer, S. Donovan and Y. Carter (2000). "Guidelines for the prevention of falls in people over 65." BMJ **321**(7267): 1007-1011.

Festa, E. K., W. C. Heindel and B. R. Ott (2010). "Dual-task conditions modulate the efficiency of selective attention mechanisms in Alzheimer's disease." Neuropsychologia **48**(11): 3252-3261.

Fitzpatrick, A. L., C. K. Buchanan, R. L. Nahin, S. T. DeKosky, H. H. Atkinson, M. C. Carlson, J. D. Williamson and f. t. G. E. o. M. S. Investigators (2007). "Associations of Gait Speed and Other Measures of Physical Function With Cognition in a Healthy Cohort of Elderly Persons." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **62**(11): 1244-1251.

Gabell, A. and U. S. L. Nayak (1984). "The effect of age on variability in gait." Journals of Gerontology **39**(6): 662-666.

GaitRite/CIR systems. Retrieved July, 2018, from https://www.gaitrite.com/gait-analysis-software.

Gietzelt, M., G. Nemitz, K. H. Wolf, H. Meyer Zu Schwabedissen, R. Haux and M. Marschollek (2009). "A clinical study to assess fall risk using a single waist accelerometer." Informatics for health & social care **34**(4): 181-188.

Gietzelt, M., K. H. Wolf, M. Kohlmann, M. Marschollek and R. Haux (2013). "Measurement of accelerometry-based gait parameters in people with and without dementia in the field: a technical feasibility study." Methods of Information in Medicine **52**(4): 319-325.

Giladi, N., T. Herman, I. I. Reider-Groswasser, T. Gurevich and J. M. Hausdorff (2005). "Clinical characteristics of elderly patientswith a cautious gait of unknown origin." Journal of Neurology **252**(3): 300-306.

Gillain, S., M. Boutaayamou, C. Beaudart, M. Demonceau, O. Bruyère, J. Y. Reginster, G. Garraux and J. Petermans (2018). "Assessing gait parameters with accelerometer-based methods to identify older adults at risk of falls: a systematic review." European Geriatric Medicine: 1-14.

Gillain, S., M. Dramé, F. Lekeu, V. Wojtasik, C. Ricour, J.-L. Croisier, E. Salmon and J. Petermans (2015). "Gait speed or gait variability, which one to use as a marker of risk to develop Alzheimer disease? A pilot study." Aging Clinical and Experimental Research **28**(2): 249-255.

Gillain, S. and J. Petermans (2013). "Contribution of new techniques to study the gait in old populations." Annals of Physical & Rehabilitation Medicine **56**(5): 384-395.

Gillain, S., E. Warzee, F. Lekeu, V. Wojtasik, D. Maquet, J. L. Croisier, E. Salmon and J. Petermans (2009). "The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects and a comparison with other clinical tests used in single and dual-task conditions." Annals of Physical and Rehabilitation Medicine **52**(6): 453-474.

Goetz, C. G., B. C. Tilley, S. R. Shaftman, G. T. Stebbins, S. Fahn, P. Martinez-Martin, W. Poewe, C. Sampaio, M. B. Stern, R. Dodel, B. Dubois, R. Holloway, J. Jankovic, J. Kulisevsky, A. E. Lang, A. Lees, S. Leurgans, P. A. LeWitt, D. Nyenhuis, C. W. Olanow, O. Rascol, A. Schrag, J. A. Teresi, J. J. van Hilten and N. LaPelle (2008). "Movement Disorder Society-sponsored revision of the Unified Parkinson's Disease Rating Scale (MDS-UPDRS): Scale presentation and clinimetric testing results." Movement Disorders **23**(15): 2129-2170.

Gorton, I., G. E. Hebert and M. E. Gannotti (2009). "Assessment of the kinematic variability among 12 motion analysis laboratories." Gait & Posture **29**(3): 398-402.

Grillner, S. (1985). "Central pattern generators for locomotion with special reference to vertebrates." Annual review of Neurosciences **8**: 233-261

Gschwind, Y. J., I. Wolf, S. A. Bridenbaugh and R. W. Kressig (2011). "Basis for a Swiss perspective on fall prevention in vulnerable older people." Swiss Med Wkly **141**: w13305.

Guralnik, J. M., E. M. Simonsick, L. Ferrucci, R. J. Glynn, L. F. Berkman, D. G. Blazer, P. A. Scherr and R. B. Wallace (1994). "A Short Physical Performance Battery Assessing Lower Extremity Function: Association with Self-reported Disability and Prediction of Mortality and Nursing Home Admission." Journal of Gerontology **49**(2): 85-94.

Hamacher, D., D. Hamacher, F. Herold and L. Schega (2016). "Are there differences in the dual-task walking variability of minimum toe clearance in chronic low back pain patients and healthy controls?" Gait & Posture **49**: 97-101.

Hamacher, D., D. Hamacher and L. Schega (2014). "Towards the importance of minimum toe clearance in level ground walking in a healthy elderly population." Gait & Posture **40**(4): 727-729.

Hartmann, A., S. Luzi, K. Murer, R. A. de Bie and E. D. de Bruin (2009). "Concurrent validity of a trunk tri-axial accelerometer system for gait analysis in older adults." Gait Posture **29**(3): 444-448.

Hartmann, A., K. Murer, R. A. de Bie and E. D. de Bruin (2009). "Reproducibility of spatio-temporal gait parameters under different conditions in older adults using a trunk tri-axial accelerometer system." Gait & Posture **30**(3): 351-355.

Hashish, R., M. E. Toney-Bolger, S. S. Sharpe, B. D. Lester and A. Mulliken (2017). "Texting during stair negotiation and implications for fall risk." Gait & Posture **58**: 409-414.

Hausdorff, J. M. (2005). "Gait variability: methods, modeling and meaning." Journal of NeuroEngineering and Rehabilitation **2**: 19-19.

Hausdorff, J. M., M. E. Cudkowicz, R. Firtion, W. J. I. and A. L. Goldberger (1998). "Gait variability and basal ganglia disorders: Stride-to-stride variations of gait cycle timing in parkinson's disease and Huntington's disease." Movement Disorders **13**(3): 428-437.

Hausdorff, J. M., H. K. Edelberg, S. L. Mitchell, A. L. Goldberger and J. Y. Wei (1997). "Increased gait unsteadiness in community-dwelling elderly fallers." Archives of Physical Medicine and Rehabilitation **78**(3): 278-283.

Hausdorff, J. M., M. E. Nelson, D. Kaliton, J. E. Layne, M. J. Bernstein, A. Nuernberger and M. A. F. Singh (2001). "Etiology and modification of gait instability in older adults: a randomized controlled trial of exercise." Journal of Applied Physiology **90**(6): 2117-2129.

Hausdorff, J. M., D. A. Rios and H. K. Edelberg (2001). "Gait variability and fall risk in community-living older adults: A 1-year prospective study." Arch Phys Med Rehabil **82**.

Hausdorff, J. M., D. A. Rios and H. K. Edelberg (2001). "Gait variability and fall risk in community-living older adults: A 1-year prospective study." Archives of Physical Medicine and Rehabilitation **82**(8): 1050-1056.

Hausdorff, J. M., A. Schweiger, T. Herman, G. Yogev-Seligmann and N. Giladi (2008). "Dual-Task Decrements in Gait: Contributing Factors Among Healthy Older Adults." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **63**(12): 1335-1343.

Hausdorff, J. M., G. Yogev, S. Springer, E. S. Simon and N. Giladi (2005). "Walking is more like catching than tapping: Gait in the elderly as a complex cognitive task." Exp Brain Res **164**.

Hawker, G. A., S. Mian, T. Kendzerska and M. French (2011). "Measures of adult pain: Visual Analog Scale for Pain (VAS Pain), Numeric Rating Scale for Pain (NRS Pain), McGill Pain Questionnaire (MPQ), Short-Form McGill Pain Questionnaire (SF-MPQ), Chronic Pain Grade Scale (CPGS), Short Form-36 Bodily Pain Scale (SF-36 BPS), and Measure of Intermittent and Constant Osteoarthritis Pain (ICOAP)." Arthritis Care & Research **63**(S11): S240-S252.

Heinrich, S., K. Rapp, U. Rissmann, C. Becker and H.-H. König (2010). "Cost of falls in old age: a systematic review." Osteoporosis International **21**(6): 891-902.

Herman, T., A. Mirelman, N. Giladi, A. Schweiger and J. M. Hausdorff (2010). "Executive Control Deficits as a Prodrome to Falls in Healthy Older Adults: A Prospective Study Linking Thinking, Walking, and Falling." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **65A**(10): 1086-1092.

Jackson AS, B. S., Mahar MT, Wier LT, Ross RM, Stuteville JE. (1990). "Prediction of functional aerobic capacity without exercise testing." Medicine and Science in Sports and Exercise **22**(6): 863-870.

Jackson, A. S. and R. M. Ross (1996). "Methods and limitations of assessing functional work capacity objectively." J Back Musculoskelet Rehabil. **6**(3): 265-276.

Jahn, K., J. Wagner, A. Deutschländer, R. Kalla, K. Hüfner, T. Stephan, M. Strupp and T. Brandt (2009). "Human hippocampal activation during stance and locomotion." Annals of the New York academy of sciences: 229-235.

Janssen, I., S. B. Heymsfield, R. N. Baumgartner and R. Ross (2000). "Estimation of skeletal muscle mass by bioelectrical impedance analysis." Journal of Applied Physiology **89**(2): 465-471.

Janssen, I., S. B. Heymsfield and R. Ross (2002). "Low Relative Skeletal Muscle Mass (Sarcopenia) in Older Persons Is Associated with Functional Impairment and Physical Disability." Journal of the American Geriatrics Society **50**(5): 889-896.

Jensen, J. L., L. A. Brown and M. H. Woollacott (2001). "Compensatory Stepping: The Biomechanics of a Preferred Response Among Older Adults." Experimental Aging Research **27**(4): 361-376.

Judge J, D. R., Ounpuu S (1996). "Step length reductions in advanced age: the role of ankle and hip kinetics." J Gerontol A Biol Sci Med Sci **51**(6): 303-312.

Katz, S., A. B. Ford, R. W. Moskowitz, B. A. Jackson and M. W. Jaffe (1963). "Studies of illness in the aged: The index of adl: a standardized measure of biological and psychosocial function." JAMA **185**(12): 914-919.

Kavanagh, J. J., R. S. Barrett and S. Morrison (2004). "Upper body accelerations during walking in healthy young and elderly men." Gait & Posture **20**(3): 291-298.

Kavanagh, J. J., R. S. Barrett and S. Morrison (2005). "Age-related differences in head and trunk coordination during walking." Human Movement Science **24**(4): 574-587.

Kavanagh, J. J. and H. B. Menz (2008). "Accelerometry: A technique for quantifying movement patterns during walking." Gait & Posture **28**(1): 1-15.

Kelsey, J. L., E. Procter-Gray, M. T. Hannan and W. Li (2012). "Heterogeneity of Falls Among Older Adults: Implications for Public Health Prevention." American Journal of Public Health **102**(11): 2149-2156.

Kenny, R., L. Rubenstein, M. E. Tinetti, K. Brewer, K. Cameron, E. Capezuti, D. P. John, S. Lamb, F. Martin, P. Rockey, M. Suther and E. Walker Peterson (2011). "Summary of the Updated American Geriatrics Society/British Geriatrics Society Clinical Practice Guideline for Prevention of Falls in Older Persons." J Am Geriatr Soc **59**: 148-157.

Khandoker, A. H., M. Palaniswami and R. K. Begg (2008). "A comparative study on approximate entropy measure and poincaré plot indexes of minimum foot clearance variability in the elderly during walking." Journal of NeuroEngineering and Rehabilitation **5**: 4-4.

Killeen, T., C. S. Easthope, L. Demkó, L. Filli, L. Lőrincz, M. Linnebank, A. Curt, B. Zörner and M. Bolliger (2017). "Minimum toe clearance: probing the neural control of locomotion." Scientific Reports **7**: 1922.

Ko, S., S. Stenholm, C. W. Chia, E. M. Simonsick and L. Ferrucci (2011). "Gait pattern alterations in older adults associated with type 2 diabetes in the absence of peripheral neuropathy—Results from the Baltimore Longitudinal Study of Aging." Gait & Posture **34**(4): 548-552.

Ko, S., M. I. Tolea, J. M. Hausdorff and L. Ferrucci (2011). "Sex-specific differences in gait patterns of healthy older adults: Results from the Baltimore Longitudinal Study of Aging." Journal of Biomechanics **44**(10): 1974-1979.

Ko, S., S. M. Ling, J. Winters and L. Ferrucci (2009). "Age-Related Mechanical Work Expenditure During Normal Walking: the Baltimore Longitudinal Study of Aging." Journal of biomechanics **42**(12): 1834-1839.

L. Lundin-Olsson, L. N., Y. Gustafson (1997). ""“Stops walking when talking” as a predictor of falls in elderly people." Lancet **349**: 617.

la Fougère, C., A. Zwergal, A. Rominger, S. Förster, G. Fesl, M. Dieterich, T. Brandt, M. Strupp, P. Bartenstein and K. Jahn (2010). "Real versus imagined locomotion: A [18F]-FDG PET-fMRI comparison." NeuroImage **50**(4): 1589-1598.

Laessoe, U., H. C. Hoeck, O. Simonsen, T. Sinkjaer and M. Voigt (2007). "Fall risk in an active elderly population – can it be assessed?" Journal of Negative Results in Biomedicine **6**: 2-2.

Laessoe, U., H. C. Hoeck, O. Simonsen and M. Voigt (2008). "Residual attentional capacity amongst young and elderly during dual and triple task walking." Human Movement Science **27**(3): 496-512.

Lamb, S. E., E. C. Jørstad-Stein, K. Hauer, C. Becker, E. on behalf of the Prevention of Falls Network and G. Outcomes Consensus (2005). "Development of a Common Outcome Data Set for Fall Injury Prevention Trials: The Prevention of Falls Network Europe Consensus." Journal of the American Geriatrics Society **53**(9): 1618-1622.

Lawton MP, B. E. (1969). "Assessment of older people: self-maintaining and instrumental activities of daily living." Gerontologist **9**: 179-186.

Levine, D., J. Richards and M. W. Whittle (2012). Whittle's Gait Analysis.

Levinger, P., D. T. H. Lai, H. B. Menz, A. D. Morrow, J. A. Feller, J. R. Bartlett, N. R. Bergman and R. Begg "Swing limb mechanics and minimum toe clearance in people with knee osteoarthritis." Gait & Posture **35**(2): 277-281.

Linn, B. S., M. W. Linn and L. E. E. Gurel (1968). "Cumulative Illness Rating Scale " J Am Geriatr Soc **16**(5): 622-626.

Lipman, R. (1982). "Differentiating anxiety and depression in anxiety disorders: Use of rating scales." Psychopharmacological Bulletin **18**(4): 69-77

M. Vandresse, J. D., J-M. Paul. (2017). "Population projections 2016-2060." Retrieved October, 2017.

Maki, B. E. (1997). "Gait Changes in Older Adults: Predictors of Falls or Indicators of Fear?" J Am Geriatr Soc **45**(3): 313-320.

Maki, B. E. and W. McIlroy (1996). "Postural control in the older adult." Clin Geriatr Med **12**(4): 635-658.

Maki, B. E. and W. E. McIlroy (2006). "Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention." Age and Ageing **35**(suppl_2): ii12-ii18.

Malouin, F., C. L. Richards, P. L. Jackson, F. Dumas and J. Doyon (2003). "Brain activations during motor imagery of locomotor-related tasks: a PET study." Human Brain Mapping **19**: 47-62.

Marschollek, M., G. Nemitz, M. Gietzelt, K. H. Wolf, H. Meyer Zu Schwabedissen and R. Haux (2009). "Predicting in-patient falls in a geriatric clinic: a clinical study combining assessment data and simple sensory gait measurements." Zeitschrift fur Gerontologie und Geriatrie **42**(4): 317-321.

Marschollek, M., A. Rehwald, K.-H. Wolf, M. Gietzelt, G. Nemitz, H. M. zu Schwabedissen and M. Schulze (2011). "Sensors vs. experts - A performance comparison of sensor-based fall risk assessment vs. conventional assessment in a sample of geriatric patients." BMC Medical Informatics and Decision Making **11**: 48-48.

Marschollek, M., A. Rehwald, K. H. Wolf, M. Gietzelt, G. Nemitz, H. Meyer zu Schwabedissen and R. Haux (2011). "Sensor-based Fall Risk Assessment – an Expert 'to go'." Methods of Information in Medicine **50**(5): 420-426.

McIlroy, W. and B. E. Maki (1996). "Age-related Changes in Compensatory Stepping in Response to Unpredictable Perturbations." The Journals of Gerontology: Series A **51A**(6): M289-M296.

McNeill Alexander, R. (2002). "Energetics and optimization of human walking and running: the 2000 Raymond Pearl memorial lecture." American Journal of Human Biology **14**(5): 641-648.

Menz, H. B., S. R. Lord and R. C. Fitzpatrick (2003). "Acceleration Patterns of the Head and Pelvis When Walking Are Associated With Risk of Falling in Community-Dwelling Older People." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **58**(5): M446-M452.

Menz, H. B., S. R. Lord and R. C. Fitzpatrick (2003). "Acceleration patterns of the head and pelvis when walking on level and irregular surfaces." Gait & Posture **18**(1): 35-46.

Menz, H. B., S. R. Lord and R. C. Fitzpatrick (2003). "Age-related differences in walking stability." Age and Ageing **32**(2): 137-142.

Mignardot, J. B., T. Deschamps, E. Barrey, B. Auvinet, G. Berrut, C. Cornu, T. Constans and L. de Decker (2014). "Gait disturbances as specific predictive markers of the first fall onset in elderly people: a two-year prospective observational study." Front Aging Neurosci **6**: 22.

Miller, M. D., C. F. Paradis, P. R. Houck, S. Mazumdar, J. A. Stack, A. H. Rifai, B. Mulsant and C. F. Reynolds (1992). "Rating chronic medical illness burden in geropsychiatric practice and research: Application of the Cumulative Illness Rating Scale." Psychiatry Research **41**(3): 237-248.

Mills, P. M. and R. S. Barrett (2001). "Swing phase mechanics of healthy young and elderly men." Human Movement Science **20**(4): 427-446.

Moe-Nilssen, R. (1998). "A new method for evaluating motor control in gait under real-life environmental conditions. Part 1: The instrument." Clinical Biomechanics **13**(4): 320-327.

Moe-Nilssen, R. (1998). "A new method for evaluating motor control in gait under real-life environmental conditions. Part 2: Gait analysis." Clinical Biomechanics **13**(4): 328-335.

Moe-Nilssen, R. (1998). "Test-retest reliability of trunk accelerometry during standing and walking." Archives of Physical Medicine and Rehabilitation **79**(11): 1377-1385.

Moe-Nilssen, R. and J. L. Helbostad (2004). "Estimation of gait cycle characteristics by trunk accelerometry." Journal of Biomechanics **37**(1): 121-126.

Montero-Odasso, M., S. W. Muir and M. Speechley (2012). "Dual-task complexity affects gait in people with mild cognitive impairment: the interplay between gait variability, dual tasking, and risk of falls." Arch Phys Med Rehabil **93**(2): 293-299.

Montero-Odasso, M., J. Verghese, O. Beauchet and J. M. Hausdorff (2012). "Gait and Cognition: A Complementary Approach to Understanding Brain Function and the Risk of Falling." Journal of the American Geriatrics Society **60**(11): 2127-2136.

Montero-Odasso, M. M., Y. Sarquis-Adamson, M. Speechley and et al. (2017). "Association of dual-task gait with incident dementia in mild cognitive impairment: Results from the gait and brain study." JAMA Neurology **74**(7): 857-865.

Morrison, A., T. Fan, S. S. Sen and L. Weisenfluh (2013). "Epidemiology of falls and osteoporotic fractures: a systematic review." ClinicoEconomics and Outcomes Research: CEOR **5**: 9-18.

Mortaza N, A. O. N., Mehdikhani N (2014). "Are gait parameters capable of distinguishing a faller than a non-faller elderly?" Eur J Phys Rehabil Med **50**(6): 677-691.

Motulsky, H. (2002). Biostatistique, une approche intuitive. Brussels.

Muir-Hunter, S. W. and J. E. Wittwer (2016). "Dual-task testing to predict falls in community-dwelling older adults: a systematic review." Physiotherapy **102**(1): 29-40.

Muir, S. W., M. Speechley, J. Wells, M. Borrie, K. Gopaul and M. Montero-Odasso (2012). "Gait assessment in mild cognitive impairment and Alzheimer's disease: the effect of dual-task challenges across the cognitive spectrum." Gait Posture **35**(1): 96-100.

Murray MP, K. R., Clarkson BH. (1969). "Walking patterns in healthy old men." J Gerontol **24**(2): 169-178.

Nasreddine, Z. S., N. A. Phillips, V. Bédirian, S. Charbonneau, V. Whitehead, I. Collin, J. L. Cummings and H. Chertkow (2005). "The Montreal Cognitive Assessment, MoCA: A Brief Screening Tool For Mild Cognitive Impairment." Journal of the American Geriatrics Society **53**(4): 695-699.

Nutt, J.C. and P.D. Thompson (1993). "Human walking and higher-level gait dosirders, particularly in the elderly." Neurology **43**: 268-279.

O'Connor, C. M., S. K. Thorpe, M. J. O'Malley and C. L. Vaughan (2007). "Automatic detection of gait events using kinematic data." Gait & Posture **25**(3): 469-474.

Pai, Y., M. Rogers, J. Patton, T. Cain and T. Hanke (1998). "Static versus dynamic predictions of protective stepping following waist-pull perturbations in young and older adults." Journal of Biomechanics **31**(12): 1111-1118.

Podsiadlo, D. and S. Richardson (1991). "The timed "up and go"; A test of basic functional mobility for frail elderly persons." J Am Geriatr Soc **39**.

Quadri, P., M. Tettamanti, S. Bernasconi, F. Trento and F. Loew (2005). "Lower limb function as predictor of falls and loss of mobility with social repercussions one year after discharge among elderly inpatients." Aging Clinical and Experimental Research **17**(2): 82-89.

Remaud, A., C. Thuong-Cong and M. Bilodeau (2015). "Age-Related Changes in Dynamic Postural Control and Attentional Demands are Minimally Affected by Local Muscle Fatigue." Frontiers in Aging Neuroscience **7**: 257.

Rolfson DB, M. S., Tsuyuki RT, Tahir A, Rockwood K. (2006). "Validity and reliability of the Edmonton Frail Scale." Age and Aging **35**(5): 526-529.

Rosenblatt, N. J., A. Bauer and M. D. Grabiner (2016). "Relating minimum toe clearance to prospective, self-reported, trip-related stumbles in the community." Prosthetics and Orthotics International **41**(4): 387-392.

Rubenstein, L. Z. (2006). "Falls in older people: Epidemiology, risk factors and strategies for prevention." Age Ageing **35**.

Rubenstein, L. Z., J. O. Harker, A. Salvà, Y. Guigoz and B. Vellas (2001). "Screening for Undernutrition in Geriatric Practice: Developing the Short-form Mini-nutritional Assessment (MNA-SF)." The Journals of Gerontology Series A: Biological Sciences and Medical Sciences **56**(6): 366-372.

Santhiranayagam, B. K., D. T. H. Lai, W. A. Sparrow and R. K. Begg (2015). "Minimum toe clearance events in divided attention treadmill walking in older and young adults: a cross-sectional study." Journal of NeuroEngineering and Rehabilitation **12**(1): 58.

Scheffer, A. C., M. J. Schuurmans, N. van Dijk, T. van der Hooft and S. E. de Rooij (2008). "Fear of falling: measurement strategy, prevalence, risk factors and consequences among older persons." Age and Ageing **37**(1): 19-24.

Scherder, E., L. Eggermont, D. Swaab, M. van Heuvelen, Y. Kamsma, M. de Greef, R. van Wijck and T. Mulder (2007). "Gait in ageing and associated dementias; its relationship with cognition." Neuroscience & Biobehavioral Reviews **31**(4): 485-497.

Scherder, E., L. Eggermont, C. Visscher, P. Scheltens and D. Swaab (2011). "Understanding higher level gait disturbances in mild dementia in order to improve rehabilitation: 'Last in–first out'." Neuroscience & Biobehavioral Reviews **35**(3): 699-714.

Schrouff, J., M. J. Rosa, J. M. Rondina, A. F. Marquand, C. Chu, J. Ashburner, C. Phillips, J. Richiardi and J. Mourão-Miranda (2013). "PRoNTo: Pattern Recognition for Neuroimaging Toolbox." Neuroinformatics **11**(3): 319-337.

Schulz, B. W., J. D. Lloyd and W. E. Lee (2010). "The effects of everyday concurrent tasks on overground minimum toe clearance and gait parameters." Gait & Posture **32**(1): 18-22.

Schwartz, C., V. Denoël, B. Forthomme, J.-L. Croisier and O. Brüls (2015). "Merging multi-camera data to reduce motion analysis instrumental errors using Kalman filters." Computer Methods in Biomechanics and Biomedical Engineering **18**(9): 952-960.

Scott, D., P. McLaughlin, G. C. Nicholson, P. R. Ebeling, A. L. Stuart, D. Kay and K. M. Sanders (2015). "Changes in gait performance over several years are associated with recurrent falls status in community-dwelling older women at high risk of fracture." Age and Ageing **44**(2): 287-293.

Sekiya, N., Nagasaki, H. (1998). "Reproductibility of the walking patterns of normal young adults: Test-retest reliability of the walk ratio (step-length/step-rate)." Gait Posture **7**(3): 225-227.

Senden, R., H. H. Savelberg, B. Grimm, I. C. Heyligers and K. Meijer (2012). "Accelerometry-based gait analysis, an additional objective approach to screen subjects at risk for falling." Gait & Posture **36**(2): 296-300.

Sheridan, P. L., J. Solomont, N. Kowall and J. M. Hausdorff (2003). "Influence of executive function on locomotor function: Divided attention increases gait variability in alzheimer's disease." J Am Geriatr Soc **51**.

Sint Jan, S. V. and U. D. Croce (2005). Clinical Biomechanics **20**(6): 659-660.

Siu, K.C., L.S. Chou, U. Mayr, P. van Donkelaar and M. H. Woollacott (2009). "Attentional mechanisms contributing to balance constraints during gait: The effects of balance impairments." Brain Research **1248**: 59-67.

Snijders, A. H., B. P. van de Warrenburg, N. Giladi and B. R. Bloem (2007). "Neurological gait disorders in elderly people: clinical approach and classification." The Lancet Neurology **6**(1): 63-74.

Springer, S., N. Giladi, C. Peretz, G. Yogev, E. S. Simon and J. M. Hausdorff (2006). "Dual-tasking effects on gait variability: The role of aging, falls, and executive function." Mov Disord **21**.

Srygley, J. M., T. Herman, N. Giladi and J. M. Hausdorff (2009). "Self-Report of Missteps in Older Adults: A Valid Proxy of Fall Risk?" Archives of Physical Medicine and Rehabilitation **90**(5): 786-792.

Storti, K. L., K. K. Pettee, J. S. Brach, J. B. Talkowski, C. R. Richardson and A. M. Kriska (2008). "Gait speed and step-count monitor accuracy in community-dwelling older adults." Medicine & Science in Sports & Exercise **40**(1): 59-64.

Sun, D. Q., J. Huang, R. Varadhan and Y. Agrawal (2016). "Race and fall risk: data from the National Health and Aging Trends Study (NHATS)." Age and Ageing **45**(1): 120-127.

Taborri, J., E. Palermo, S. Rossi and P. Cappa (2016). "Gait Partitioning Methods: A Systematic Review." Sensors (Basel, Switzerland) **16**(1): 66.

Tanaka, A., H. Okuzumi, I. Kobayashi, N. Murai, K. Meguro and T. Nakamura (1995). "Gait Disturbance of Patients with Vascular and Alzheimer-Type Dementias." Perceptual and Motor Skills **80**(3): 735-738.

Thiem, U., R. Klaaßen-Mielke, U. Trampisch, A. Moschny, L. Pientka and T. Hinrichs (2014). "Falls and EQ-5D rated quality of life in community-dwelling seniors with concurrent chronic diseases: a cross-sectional study." Health and Quality of Life Outcomes **12**: 2-2.

Tinetti, M. E. and C. Kumar (2010). "The patient who falls: “it's always a trade-off”." JAMA **303**(3): 258-266.

Tinetti , M. E., M. Speechley and S. F. Ginter (1988). "Risk Factors for Falls among Elderly Persons Living in the Community." New England Journal of Medicine **319**(26): 1701-1707.

van Schooten, K. S., M. Pijnappels, S. M. Rispens, P. J. M. Elders, P. Lips, A. Daffertshofer, P. J. Beek and J. H. van Dieën (2016). "Daily-Life Gait Quality as Predictor of Falls in Older People: A 1-Year Prospective Cohort Study." PLOS ONE **11**(7): e0158623.

Vellas, B., L. Balardy, S. Gillette-Guyonnet, G. Abellan Van Kan, A. Ghisolfi-Marque, J. Subra, S. Bismuth, S. Oustric and M. Cesari (2013). "Looking for Frailty in Community-dwelling Older Persons: The Gérontopôle Frailty Screening Tool (GFST)." The Journal of Nutrition, Health & Aging. **17**(7): 629-631.

Verghese, J., R. Holtzer, R. B. Lipton and C. Wang (2009). "Quantitative gait markers and incident fall risk in older adults." J Gerontol A Biol Sci Med Sci **64**(8): 896-901.

Verghese, J., J. Mahoney, A. F. Ambrose, C. Wang and R. Holtzer (2010). "Effect of cognitive remediation on gait in sedentary seniors." J Gerontol A Biol Sci Med Sci **65**(12): 1338-1343.

W.Whittle, M. (2007). Methods of gait analysis. Gait analysis an introduction**:** 165.

Wagner, J., T. Stephan, R. Kalla, B. Hartmut, M. Strupp, T. Brandt and K. Jahn (2008). "Mind the bend:cerbral activations associated with mental imagery of walking along a curved path." Exp Brain Res **191**: 247-255.

Wang, J., Y. Wai, Y. Weng, K. Ng, Y.-Z. Huang, L. Ying, H. Liu and C. Wang (2009). "Functional MRI in the assessment of cortical activation during gait-related imaginary tasks." Journal of Neural Transmission **116**(9): 1087-1092.

Ward, R. E., S. G. Leveille, M. K. Beauchamp, T. Travison, N. Alexander, A. M. Jette and J. F. Bean (2015). "Functional Performance As a Predictor of Injurious Falls Among Older Adults." Journal of the American Geriatrics Society **63**(2): 315-320.

Watson, W. L., A. J. Clapperton and R. J. Mitchell (2011). "The cost of fall-related injuries among older people in NSW, 2006–07." New South Wales Public Health Bulletin **22**(4): 55-59.

WHO. (2018, January 2018). "Falls; key facts." Retrieved July 2018, from http://www.who.int/news-room/fact-sheets/detail/falls.

Willemsen, A. T. M., F. Bloemhof and H. B. K. Boom (1990). "Automatic stance-swing phase detection from accelerometer data for peroneal nerve stimulation." IEEE Transactions on Biomedical Engineering **37**(12): 1201-1208.

Windolf, M., N. Götzen and M. Morlock (2008). "Systematic accuracy and precision analysis of video motion capturing systems—exemplified on the Vicon-460 system." Journal of Biomechanics **41**(12): 2776-2780.

Winter (1992). "Foot trajectory in human gait: a precise and multifactorial motor control task." Phys Ther. **72**(1): 45-53.

Winter, D. A., A. E. Patla, J. S. Frank and S. E. Walt (1990). "Biomechanical Walking Pattern Changes in the Fit and Healthy Elderly." Physical Therapy **70**(6): 340-347.

Witten, I. and Franck, E. (2005). Credibility: evaluating what's been learned. Data Mining: Practicale machine learning tools and techniques. Elsevier. San Francisco, Morgan Kaufmann**:** 149-151.

Witten, I. and Franck, E. (2005). Implementations: real machine learning schemes. Data Mining: practicale machine learning tools and techniques. Elsevier. San Francisco, Morgan Kaufmann**:** 191-192.

Wolfson, L. I., R. Whipple, P. Amerman and A. Kleinberg (1986). "Stressing the Postural Response." J Am Geriatr Soc **34**(12): 845-850.

Woollacott, M. and A. Shumway-Cook (2002). "Attention and the control of posture and gait: a review of an emerging area of research." Gait & Posture **16**(1): 1-14.

Woollacott, M. H. and P.-F. Tang (1997). "Balance Control During Walking in the Older Adult: Research and Its Implications." Physical Therapy **77**(6): 646-660.

Wrisley, D. M., G. F. Marchetti, D. K. Kuharsky and S. L. Whitney (2004). "Reliability, Internal Consistency, and Validity of Data Obtained with the Functional Gait Assessment." Physical Therapy **84**(10): 906-918.

Yogev-Seligmann, G., Y. Rotem-Galili, A. Mirelman, R. Dickstein, N. Giladi and J. M. Hausdorff (2010). "How Does Explicit Prioritization Alter Walking During Dual-Task Performance? Effects of Age and Sex on Gait Speed and Variability." Physical Therapy **90**(2): 177-186.

Yogev, G., M. Plotnik, C. Peretz, N. Giladi and J. M. Hausdorff (2007). "Gait asymmetry in patients with Parkinson's disease and elderly fallers: when does the bilateral coordination of gait require attention?" Experimental Brain Research **177**(3): 336-346.

Zijlstra, A., E. D. de Bruin, N. Bruins and W. Zijlstra (2008). "The step length-frequency relationship in physically active community-dwelling older women." European Journal of Applied Physiology **104**(3): 427-434.

Zijlstra, A., T. Ufkes, D. A. Skelton, L. Lundin-Olsson and W. Zijlstra (2008). "Do Dual Tasks Have an Added Value Over Single Tasks for Balance Assessment in Fall Prevention Programs? A Mini-Review." Gerontology **54**(1): 40-49.

Zijlstra, A. and W. Zijlstra (2013). "Trunk-acceleration based assessment of gait parameters in older persons: A comparison of reliability and validity of four inverted pendulum based estimations." Gait & Posture **38**(4): 940-944.

Zijlstra, W. (2004). "Assessment of spatio-temporal parameters during unconstrained walking." European Journal of Applied Physiology **92**(1-2): 39-44.

Zijlstra, W. and A. L. Hof (2003). "Assessment of spatio-temporal gait parameters from trunk accelerations during human walking." Gait & Posture **18**(2): 1-10.

11. Abbreviations list

ADL: Activities of daily living

AUC: Area under the curve

BIA: bioelectrical impedance

CIRS-g: Cumulative Illness Rating Scale geriatric version

CoV MTC: coefficient variation of MTC values

CW: Comfort walking condition

DTW: Dual task walking condition

FES-I: French version of the falls efficacy scale

FGA: Functional gait assessment test

FW: Fast walking condition

GDS-4: Geriatric depression scale short version

GFST: Gerontopôle frailty screening tool

GS: Gait speed

HS: Hell strike

IADL: Instrumental activities of daily living

ICC: Intra-correlation coefficient

IQR MTC: Interquartile range of MTC values

Max: Maximal

Mean MTC: the mean MTC value

Med MTC: the median MTC value

Min: Minimal

Min MTC: the minima MTC value

MNA-14: Mini Nutritional Assessment short version

MoCA: Montreal cognitive assessment

MTC: Minimal Toe Clearance

NPV: Negative predictive value

PASS: Physical Activity Status Scale

PRC: Precision recall area

PPV: Positive predictive value

Q1: 25^{th} percentile

Q3: 75^{th} percentile

REG: Stride regularity

ROC: Receiver operating characteristics

SD: Standard deviation

SD MTC: Standard deviation of MTC values

SF: Stride frequency

SL: Stride length

SM: skeletal muscle

SMI: skeletal muscle index

SMM: skeletal muscle mass

SPPB: Short physical performance battery

SYM: Stride symmetry

TO: Toe off

TUG: Timed up and go test

UPDRS: Unified Parkinson's disease rating scale

Var. MTC: Variance of MTC values

WHO: World Health Organisation

12. Appendices

Appendices following are namely:

- Appendix 1: Correlation between the parameters obtained from the two different instrumental methods used in this study.
- Appendix 2: Approval of the Ethical Committee
- Appendix 3: Comparison according the symmetry DTW cost quartiles
- Appendix 4: Comparison according the FW normalized stride length quartiles
- Appendix 5: Logistic regression analyses with or without considering IADL
- Appendix 7: Comparison according the FW stride length quartiles
- Appendix 8: Comparison of anamnestic, clinical and functional data including gait parameters according to the regularity FW improvement

Appendix 1: Correlation between the parameters obtained from the two different instrumental methods used in this study.

Validation des paramètres de marche fournis par le Locométrix ®

S. Gillain[1], C. Schwartz[2] , M. Dramé[3], M. Demonceau[4] , O. Bruyère[5], JL. Croisier [4], O. Brüls[2] , G. Garraux[6] , JY. Reginster[5], J. Petermans[1]

CHU de Liège Belgique: Service de gériatrie[1], LAMH faculté d'ingénierie civile[2], Département des sciences de la motricité[4], Ecole de santé publique[5], Service de neurologie[6] – CHU de Reims, France: Ecole de santé publique[3]

Introduction: Le Locométrix® est un accéléromètre triaxial portable permettant l'analyse de la marche. Bien que reproductible (1, 2), sa validation reste non documentée.

But du travail: Valider les paramètres de vitesse, cadence et longueur des pas obtenus via l'accéléromètre, à l'aide d'un système d'analyse optoélectronique tridimensionnelle du mouvement, le Coda Motion® précédemment considéré comme méthode de référence (3).

Matériel et méthode: 19 volontaires (de 21 à 77 ans), équipés simultanément de l'accéléromètre et de 4 marqueurs de mouvements placés aux pieds ont réalisé à 5 reprises une marche à vitesse confortable, dans un couloir au sein duquel, la vitesse de marche « stabilisée » est chronométrée. Le Locométrix® analyse les courbes d'accélérations cranio-caudales obtenues lors de 20, 48 secondes de marche « stabilisée ». Le Coda Motion® placé à mi-chemin du parcours de marche, étudie le mouvement des 4 marqueurs placés au pieds durant 3-4 cycles de marche.

Au total, 95 marches (5X19) sont analysées. La reproductibilité intra-sujet des mesures obtenues par les deux techniques est appréciée en considérant les coefficients de corrélation intra-classe (ICC).

Résultats : Les ICC calculés, tous âges confondus, sont tous supérieurs à 0.95 (excellents).

Variables motrices	Locométrix®	Coda Motion®	ICC (IC 95%)
Vitesse moyenne (m/s)(écart-type)	1,457 (0,514)	1,513 (0,533)	0,979 (0,969-0,986)
Longueur moyenne (m)(écart-type)	1,452 (0,112)	1,506 (0,118)	0,977 (0,965- 0,984)
Cadence moyenne (cycle/s)(écart-type)	1,003 (0,071)	1,003 (0,072)	0, 976 (0,964- 0,984)

Discussion : Les ICC confirment l'importante corrélation entre les valeurs obtenues par les deux techniques de mesure. La littérature ne permettant pas d'appréhender le calcul des paramètres « symétrie » et « régularité » obtenus avec le Locométrix®, ces derniers n'ont pu être validés.

Conclusion : Ces résultats attestent de la validité des paramètres locomoteurs obtenus via le Locométrix®. De futurs travaux devraient valider les paramètres « symétrie » et « régularité » obtenus par le même instrument.

1.Analyse de la marche humaine dans la pratique hospitalière par une méthode accélérométrique, Auvinet B, Rev Rhum, 1999..
2. Reference data for normal subjects obtained with an accelerometric device, Auvinet B, Gait and Posture, 2002..
3. Accuracy, reliability and validity of a spatiotemporal gait analysis system, S Barker, Medical engineering and physics, 2006.

Pour contacter les auteurs: sgillain@chu.ulg.ac.be

Comité d'Ethique Hospitalo-Facultaire Universitaire de Liège (707)

Université de Liège

CHU de Liège

Sart Tilman, le 5 mars 2014

Monsieur le **Prof. J. PETERMANS**
Madame le **Dr. S. GILLAIN**
Service de **GERIATRIE**
CHU B35

Concerne: Votre demande d'avis au Comité d'Ethique
Nr belge : B707201319289; Notre réf: 2013/291

Cher Collègue,

J'ai le plaisir de vous informer que le Comité d'Ethique a donné une réponse favorable à votre demande d'avis intitulée :

"Intérêt prédictif de l'analyse instrumentale des paramètres de marche de sujets âgés : évaluation du risque de déclin fonctionnel. "

Vous trouverez, sous ce pli, le formulaire de réponse reprenant, en français et en anglais, les différents éléments examinés et approuvés et la composition du Comité d'Ethique.

Je vous prie d'agréer, Cher Collègue, l'expression de mes sentiments les meilleurs,

Prof. V. SEUTIN
Président du Comité d'Ethique

Appendix 3: Comparison according the symmetry DTW cost quartiles

Variable	Groups	Mean ± SD	Q1	Median	Q3	p-value
CW Symmetry (dimensionless)	All (n=48)	221.71 ± 76.74	178.50	218.00	254.00	
	Symmetry DTW gain ≥ 28 % (n=24)	175.42 ± 44.09	142.00	178.50	211.00	<.0001
	Symmetry DTW cost ≥ 22 % (n=24)	268.00 ± 74.95	228.50	252.50	272.50	
DTW Symmetry (dimensionless)	All (n=48)	221.48 ± 78.72	161.50	198.50	277.50	
	Symmetry DTW gain ≥ 28 %	273.04 ± 77.15	211.00	277.50	322.50	<.0001
	Symmetry DTW cost ≥ 22 % (n=24)	169.92 ± 34.10	142.50	168.00	189.50	
DTW Symmetry Cost (%)	All (n=48)	-10.37 ± 49.41	-49.15	-2.90	33.69	
	Symmetry DTW gain ≥ 28 % (n=24)	-55.79 ± 24.15	-79.95	-49.15	-34.39	<.0001
	Symmetry DTW cost ≥ 22 % (n=24)	35.06 ± 9.96	27.66	33.69	40.28	

Appendix 4: Comparison according the FW normalized stride length quartiles

Variable	Groups	Mean ± SD	Q1	Median	Q3	p-value
Edmonton (/ 17)	All (n=49)	2.02 ± 1.59	1.00	2.00	3.00	
	FW norm. Stride length ≤ 1.64 (n=24)	2.75 ± 1.54	2.00	2.50	4.00	0.0006
	FW norm. Stride length ≥ 2 (n=25)	1.32 ± 1.31	1.00	1.00	2.00	
Body Height (cm)	All (n=49)	165.65 ± 9.55	158.00	165.00	173.00	
	FW norm. Stride length ≤ 1.64 (n=24)	160.33 ± 6.27	156.50	160.00	165.00	<.0001
	FW norm. Stride length ≥ 2 (n=25)	170.76 ± 9.45	169.00	173.00	177.00	
Grip Strength (kPa)	All (n=49)	60.76 ± 17.91	50.00	52.00	72.00	
	FW norm. Stride length ≤ 1.64 (n=24)	49.50 ± 8.05	45.00	50.00	52.00	<.0001
	FW norm. Stride length ≥ 2 (n=25)	71.56 ± 18.19	52.00	72.00	86.00	
Grip Work (kPa x s)	All (n=49)	2202.2 ± 1710.8	940.28	1706.3	3208.1	
	FW norm. Stride length ≤ 1.64 (n=24)	1332.6 ± 901.15	658.41	1041.2	1785.0	0.0004
	FW norm. Stride length ≥ 2 (n=25)	3037.0 ± 1895.5	1706.3	3049.7	4404.1	
Grip work/body weight (kPa x s / kg)	All (n=49)	29.14 ± 21.79	13.43	24.39	39.33	
	FW norm. Stride length ≤ 1.64 (n=24)	18.59 ± 11.58	9.05	14.71	26.08	0.0009
	FW norm. Stride length ≥ 2 (n=25)	39.26 ± 24.56	23.70	38.37	47.59	
FGA (/30)	All (n=49)	26.31 ± 3.12	24.00	27.00	29.00	
	FW norm. Stride length ≤ 1.64 (n=24)	24.67 ± 3.16	23.00	25.00	27.50	0.0002
	FW norm. Stride length ≥ 2 (n=25)	27.88 ± 2.15	27.00	28.00	29.00	
SPPB (/12)	All (n=49)	10.22 ± 1.79	9.00	11.00	12.00	
	FW norm. Stride length ≤ 1.64 (n=24)	9.17 ± 1.90	8.00	9.50	10.50	<.0001
	FW norm. Stride length ≥ 2 (n=25)	11.24 ± 0.88	11.00	11.00	12.00	
CW gait speed (m/s)	All (n=49)	1.26 ± 0.19	1.12	1.22	1.42	
	FW norm. Stride length ≤ 1.64 (n=24)	1.13 ± 0.12	1.04	1.13	1.20	<.0001

	FW norm. Stride length ≥ 2 (n=25)	1.39 ± 0.15	1.28	1.41	1.46	
CW norm. gait speed (/s)	All (n=49)	1.53 ± 0.22	1.37	1.47	1.68	
	FW norm. Stride length ≤ 1.64 (n=24)	1.36 ± 0.12	1.31	1.37	1.43	<.0001
	FW norm. Stride length ≥ 2 (n=25)	1.69 ± 0.18	1.59	1.64	1.81	
FW gait speed (m/s)	All (n=49)	1.67 ± 0.28	1.44	1.65	1.88	
	FW norm. Stride length ≤ 1.64 (n=24)	1.45 ± 0.17	1.32	1.44	1.55	<.0001
	FW norm. Stride length ≥ 2 (n=25)	1.89 ± 0.17	1.82	1.88	1.92	
FW norm. gait speed (/s)	All (n=49)	2.02 ± 0.32	1.73	2.06	2.23	
	FW norm. Stride length ≤ 1.64 (n=24)	1.74 ± 0.16	1.64	1.73	1.80	<.0001
	FW norm. Stride length ≥ 2 (n=25)	2.28 ± 0.19	2.18	2.22	2.37	
DTW gait speed (m/s)	All (n=48)	1.13 ± 0.22	1.00	1.11	1.30	
	FW norm. Stride length ≤ 1.64 (n=24)	1.01 ± 0.15	0.94	1.02	1.11	<.0001
	FW norm. Stride length ≥ 2 (n=24)	1.25 ± 0.21	1.11	1.25	1.34	
DTW norm. gait speed (/s)	All (n=48)	1.37 ± 0.26	1.20	1.32	1.51	
	FW norm. Stride length ≤ 1.64 (n=24)	1.22 ± 0.18	1.14	1.24	1.32	<.0001
	FW norm. Stride length ≥ 2 (n=24)	1.51 ± 0.24	1.32	1.49	1.60	
CW stride length (m)	All (n=49)	1.31 ± 0.18	1.18	1.27	1.45	
	FW norm. Stride length ≤ 1.64 (n=24)	1.16 ± 0.09	1.12	1.17	1.23	<.0001
	FW norm. Stride length ≥ 2 (n=25)	1.44 ± 0.12	1.36	1.45	1.53	
CW norm. stride length	All (n=49)	1.58 ± 0.20	1.40	1.56	1.76	
	FW norm. Stride length ≤ 1.64 (n=24)	1.40 ± 0.09	1.35	1.40	1.46	<.0001
	FW norm. Stride length ≥ 2 (n=25)	1.74 ± 0.10	1.66	1.76	1.80	
FW stride length (m)	All (n=49)	1.54 ± 0.31	1.29	1.47	1.76	
	FW norm. Stride length ≤ 1.64 (n=24)	1.29 ± 0.11	1.24	1.29	1.35	<.0001

	FW norm. Stride length ≥ 2 (n=25)	1.78 ± 0.24	1.67	1.76	1.84	
FW norm. stride length	All (n=49)	1.85 ± 0.37	1.57	2.00	2.07	
	FW norm. Stride length ≤ 1.64 (n=24)	1.55 ± 0.09	1.51	1.55	1.61	<.0001
	FW norm. Stride length ≥ 2 (n=25)	2.15 ± 0.28	2.04	2.07	2.12	
DTW stride length (m)	All (n=48)	1.28 ± 0.31	1.11	1.26	1.35	
	FW norm. Stride length ≤ 1.64 (n=24)	1.12 ± 0.14	1.04	1.13	1.16	0.0002
	FW norm. Stride length ≥ 2 (n=25)	1.43 ± 0.35	1.28	1.31	1.53	
DTW norm. stride length	All (n=48)	1.54 ± 0.35	1.35	1.49	1.68	
	FW norm. Stride length ≤ 1.64 (n=24)	1.35 ± 0.16	1.24	1.36	1.43	<.0001
	FW norm. Stride length ≥ 2 (n=24)	1.73 ± 0.39	1.56	1.67	1.74	

The model obtained including the categorized IADL score (0 or 1)

Number of Observations Used	93

Response Profile		
Ordered Value	**chuteT2**	**Total Frequency**
1	1	34
2	0	59

Probability modeled is chuteT2=1.

Class Level Information		
Class	**Value**	**Design Variables**
iadl_cat	**0**	1
	1	-1

Model Convergence Status
Convergence criterion (GCONV=1E-8) satisfied.

Model Fit Statistics		
Criterion	**Intercept Only**	**Intercept and Covariates**
AIC	124.122	116.382
SC	126.654	131.577
-2 Log L	122.122	104.382

Testing Global Null Hypothesis: BETA=0			
Test	**Chi-Square**	**DF**	**Pr > ChiSq**
Likelihood Ratio	17.7400	5	0.0033
Score	15.8485	5	0.0073
Wald	11.9379	5	0.0356

Type 3 Analysis of Effects			
Effect	**DF**	**Wald Chi-Square**	**Pr > ChiSq**
iadl_cat	1	2.8339	0.0923
UPDRS	1	0.3041	0.5813
SPPB	1	2.0537	0.1518
Symdualcost	1	4.9250	0.0265
LongRAPLOCO1Norm	1	1.1618	0.2811

Analysis of Maximum Likelihood Estimates						
Parameter		**DF**	**Estimate**	**Standard Error**	**Wald Chi-Square**	**Pr > ChiSq**
Intercept		1	5.2852	2.4600	4.6160	0.0317
iadl_cat	**0**	1	-1.0365	0.6157	2.8339	0.0923
UPDRS		1	0.1050	0.1905	0.3041	0.5813
SPPB		1	-0.2476	0.1728	2.0537	0.1518
Symdualcost		1	0.0174	0.00784	4.9250	0.0265
LongRAPLOCO1Norm		1	-1.2997	1.2058	1.1618	0.2811

Odds Ratio Estimates			
Effect	**Point Estimate**	**95% Wald Confidence Limits**	
iadl_cat 0 vs 1	0.126	0.011	1.406
UPDRS	1.111	0.765	1.613
SPPB	0.781	0.556	1.095
Symdualcost	1.018	1.002	1.033
LongRAPLOCO1Norm	0.273	0.026	2.897

Association of Predicted Probabilities and Observed Responses			
Percent Concordant	72.2	**Somers' D**	0.448
Percent Discordant	27.4	**Gamma**	0.450
Percent Tied	0.3	**Tau-a**	0.210
Pairs	2006	**c**	0.724

The model obtained without considering IADL

Number of Observations Used	93

Response Profile		
Ordered Value	**chuteT2**	**Total Frequency**
1	1	34
2	0	59

Probability modeled is chuteT2=1.

Model Convergence Status
Convergence criterion (GCONV=1E-8) satisfied.

Model Fit Statistics		
Criterion	**Intercept Only**	**Intercept and Covariates**
AIC	124.122	117.831
SC	126.654	130.494
-2 Log L	122.122	107.831

Testing Global Null Hypothesis: BETA=0			
Test	**Chi-Square**	**DF**	**Pr > ChiSq**
Likelihood Ratio	14.2906	4	0.0064
Score	13.0658	4	0.0110
Wald	11.1717	4	0.0247

Analysis of Maximum Likelihood Estimates					
Parameter	**DF**	**Estimate**	**Standard Error**	**Wald Chi-Square**	**Pr > ChiSq**
Intercept	1	3.9730	2.1933	3.2812	0.0701
UPDRS	1	0.2145	0.1737	1.5247	0.2169
SPPB	1	-0.2663	0.1708	2.4315	0.1189
Symdualcost	1	0.0177	0.00765	5.3387	0.0209
LongRAPLOCO1Norm	1	-1.0161	1.1464	0.7857	0.3754

Odds Ratio Estimates			
Effect	**Point Estimate**	**95% Wald Confidence Limits**	
UPDRS	1.239	0.882	1.742
SPPB	0.766	0.548	1.071
Symdualcost	1.018	1.003	1.033
LongRAPLOCO1Norm	0.362	0.038	3.424

Association of Predicted Probabilities and Observed Responses			
Percent Concordant	69.9	**Somers' D**	0.399
Percent Discordant	30.0	**Gamma**	0.400
Percent Tied	0.1	**Tau-a**	0.187
Pairs	2006	**c**	0.700

Appendix 6: Correlations between variables involved in logistic regression

Variables	Stiffness	IADL	SPPB	FW Normalized. Stride length	Stride Symmetry DTW cost
Stiffness		-0.36890 0.0003 94	0.22659 0.0281 94	-0.10580 0.3101 94	-0.02733 0.7948 93
IADL	-0.36890 0.0003 94		0.13079 0.2040 96	0.01104 0.9150 96	-0.02033 0.8450 95
SPPB	-0.22659 0.0281 94	0.13079 0.2040 96		0.40676 <.0001 96	-0.00693 0.9469 95
FW Normalized. Stride length	-0.10580 0.3101 94	0.01104 0.9150 96	0.40676 <.0001 96		-0.11540 0.2654 95
Stride Symmetry DTW cost	-0.02733 0.7948 93	-0.02033 0.8450 95	-0.00693 0.9469 95	-0.11540 0.2654 95	

Appendix 7: Comparison according the FW stride length quartiles

Variables	Groups	Mean ± SD	Q1	Median	Q3	p-value
Edmonton (score / 17)	All (n=49)	2.20 ± 1.66	1.00	2.00	3.00	
	FW stride length ≤ 1.38 m. (n=25)	2.96 ± 1.72	2.00	3.00	4.00	0.0009
	FW stride length ≥ 1.71 m. (n=24)	1.42 ± 1.18	1.00	1.00	2.00	
Body Height (cm)	All (n=49)	168.02 ±10.33	160.00	169.00	177.00	
	FW stride length ≤ 1.38 m. (n=25)	160.56 ± 7.00	156.00	160.00	165.00	<.0001
	FW stride length ≥ 1.71 m. (n=24)	175.79 ± 6.93	171.00	177.00	180.00	
Right leg length (cm)	All (n=49)	84.27 ± 5.25	81.00	84.00	87.00	
	FW stride length ≤ 1.38 m. (n=25)	81.14 ± 4.30	78.00	81.00	83.00	<.0001
	FW stride length ≥ 1.71 m. (n=24)	87.52 ± 4.06	85.00	87.00	90.00	
Grip strength (kPa)	All (n=49)	62.18 ± 16.18	50.00	60.00	78.00	
	FW stride length ≤ 1.38 m. (n=25)	50.20 ± 8.23	46.00	50.00	54.00	<.0001
	FW stride length ≥ 1.71 m. (n=24)	74.67 ± 12.52	68.00	78.00	84.00	
Grip work (kPa*s)	All (n=49)	2435.5 ± 1697.2	1152.9	1887.0	3208.1	
	FW stride length ≤ 1.38 m. (n=25)	1483.8 ± 938.26	720.90	1380.8	1938.8	<.0001
	FW stride length ≥ 1.71 m. (n=24)	3426.9 ± 1755.7	1795.0	3128.9	4530.1	
Grip work / Body weight ($kPa*s*Kg^{-1}$)	All (n=49)	32.89 ± 22.80	15.34	27.21	42.35	
	FW stride length ≤ 1.38 m. (n=25)	21.22± 13.86	9.82	15.67	29.03	0.0003
	FW stride length ≥ 1.71 m. (n=24)	45.04 ± 24.13	25.01	39.35	63.01	
Skeletal Muscle Mass SMM (kg)	All (n=46)	26.82 ± 5.27	22.55	26.03	30.03	
	FW stride length ≤ 1.38 m. (n=22)	23.90 ± 4.49	21.51	23.15	25.75	0.0001
	FW stride length ≥ 1.71 m. (n=24)	29.51 ± 4.51	26.49	29.49	32.26	
FGA (score / 30)	All (n=49)	26.14 ± 3.40	24.00	27.00	29.00	
	FW stride length ≤ 1.38 m. (n=25)	23.96 ± 3.30	22.00	24.00	26.00	<.0001

Variables	Groups	Mean ± SD	Q1	Median	Q3	p-value
	FW stride length ≥ 1.71 m. (n=24)	28.42 ± 1.47	27.00	29.00	30.00	
SPPB (score / 12)	All (n=49)	10.10 ± 1.84	9.00	10.00	12.00	
	FW stride length ≤ 1.38 m. (n=25)	9.04 ± 1.88	8.00	9.00	10.00	<.0001
	FW stride length ≥ 1.71 m. (n=24)	11.21 ± 0.93	10.50	11.50	12.00	
CW Gait speed (m/s)	All (n=49)	1.26 ± 0.19	1.12	1.23	1.42	
	FW stride length ≤ 1.38 m. (n=25)	1.13 ± 0.13	1.04	1.12	1.20	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.41 ± 0.14	1.29	1.42	1.49	
FW Gait speed (m/s)	All (n=49)	1.68 ± 0.29	1.43	1.70	1.91	
	FW stride length ≤ 1.38 m. (n=25)	1.45 ± 0.18	1.33	1.43	1.52	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.92 ± 0.16	1.83	1.91	2.02	
DTW Gait speed (m/s)	All (n=49)	1.14 ± 0.23	0.98	1.11	1.31	
	FW stride length ≤ 1.38 m. (n=25)	1.02 ± 0.17	0.91	1.02	1.11	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.27 ± 0.21	1.12	1.26	1.35	
CW Stride length (m)	All (n=49)	1.32 ± 0.19	1.16	1.29	1.51	
	FW stride length ≤ 1.38 m. (n=25)	1.16 ± 0.10	1.12	1.16	1.23	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.49 ± 0.09	1.41	1.52	1.55	
FW Stride length (m)	All (n=49)	1.55 ± 0.32	1.29	1.38	1.76	
	FW stride length ≤ 1.38 m. (n=25)	1.27 ± 0.09	1.24	1.29	1.33	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.83 ± 0.21	1.74	1.77	1.85	
DTW Stride length (m)	All (n=49)	1.30 ± 0.32	1.10	1.26	1.46	
	FW stride length ≤ 1.38 m. (n=25)	1.12 ± 0.15	1.03	1.10	1.17	<.0001
	FW stride length ≥ 1.71 m. (n=24)	1.48 ± 0.35	1.29	1.45	1.55	
FW Stride length improvement (m)	All (n=49)	16.86 ± 16.39	9.32	15.32	20.91	
	FW stride length ≤ 1.38 m. (n=25)	9.88 ± 6.37	5.38	9.73	14.29	<.0001
	FW stride length ≥ 1.71 m. (n=24)	24.12 ± 20.26	16.50	20.53	24.03	

Variables	Groups	Mean ± SD	Q1	Median	Q3	p-value
CW Regularity (dimensionless)	All (n=49)	295.76 ± 51.23	260.00	288.00	337.00	
	FW stride length ≤ 1.38 m. (n=25)	271.40 ± 45.33	249.00	271.00	288.00	0.0003
	FW stride length ≥ 1.71 m. (n=24)	321.13 ± 44.91	289.00	321.00	353.00	
DTW MIN MTC (mm)	All (n=49)	13.81 ± 6.93	8.47	13.11	18.21	
	FW stride length ≤ 1.38 m. (n=23)	10.47 ± 4.75	7.81	10.23	13.02	0.0005
	FW stride length ≥ 1.71 m. (n=22)	17.49 ± 7.19	15.12	18.20	22.55	

Appendix 8: Comparison of anamnestic, clinical and functional data including gait parameters according to the regularity FW improvement

Variables	Groups	Mean ± SD	Q1	Median	Q3	P-value
FW Regularity (dimensionless)	All, n=96	307.15 ± 53.62	275.00	310.00	346.50	
	Regularity FW improvement +, n=57	326.93 ± 42.21	294.00	325.00	362.00	<.0001
	Regularity FW improvement -, n=39	278.23 ± 55.82	246.00	276.00	320.00	
DTW Min MTC (mm)	All, n=84	13.97 ± 7.12	8.69	13.50	18.80	
	Regularity FW improvement +, n=49	12.54 ± 7.21	6.51	11.32	18.18	0.028
	Regularity FW improvement -, n=35	15.98 ± 6.59	9.96	16.59	20.29	
Min MTC FW improvement (%)	All, n=87	-4.14 ± 77.69	-39.98	-17.47	6.44	
	Regularity FW improvement +, n=51	8.96 ± 96.18	-38.31	-10.19	17.33	0.044
	Regularity FW improvement -, n=36	-22.69 ± 31.82	-42.72	-24.02	-4.88	

13. Publications related to this topic

1. *As first author*

GILLAIN S., Boutaayamou M., Beaudart C., Demonceau M., Bruyère O., Reginster J.-Y., Garraux G., Petermans J., ASSESSING GAIT PARAMETERS WITH ACCELEROMETER-BASED METHODS TO IDENTIFY OLDER ADULTS AT RISK OF FALLS: A SYSTEMATIC REVIEW, European Geriatric Medicine, 2018.

GILLAIN S., Boutaayamou M., Dardenne N., Schwartz C., Demonceau M., Gerontitis C., Depierreux F., Salmon E., Garraux G., Bruyère O., Brüls O., Croisier J.-L., Petermans J., DATA SET OF HEALTHY OLD PEOPLE ASSESSED FOR THREE WALKING CONDITIONS USING ACCELEROMETRIC AND OPTO-ELECTRONIC METHODS. Aging Clinical and Experimental Research, 2017.

GILLAIN S., Dramé M., Lekeu F., Wojtasik V., Ricour C., Croisier J-L, Salmon E., Petermans J, GAIT SPEED OR GAIT VARIABILITY, WHICH ONE TO USE AS A MARKER OF RISK TO DEVELOP ALZHEIMER DISEASE?, Aging Clinical and Experimental Research, 2015.

GILLAIN S., Elbouz L., Beaudart C., Bruyère O., Reginster J.-Y., Petermans J. LES CHUTES DE LA PERSONNE ÂGÉE, Revue Médicale de Liège, 2014.

GILLAIN S. and Petermans J. CONTRIBUTION OF NEW TECHNIQUES TO STUDY GAIT IN THE OLDERS. Annals of Physical and Rehabilitation Medicine, 2013.

GILLAIN S., Warzee E., Lekeu F., Wojtasik V., Maquet D., Croisier J.-L., Salmon E., Petermans J. THE VALUE OF INSTRUMENTAL GAIT ANALYSIS IN ELDERLY HEALTHY, MCI OR ALZHEIMER'S DISEASE SUBJECTS AND A COMPARISON WITH OTHER CLINICAL TESTS USED IN SINGLE AND DUAL-TASK CONDITIONS. Annales de Réadaptation et de Médecine Physique, 2009.

2. Contributions

Boutaayamou M., GILLAIN S., Schwartz C. et al. VALIDATED ASSESSMENT OF GAIT SUB-PHASE DURATIONS IN OLDER ADULTS USING AN ACCELEROMETER-BASED AMBULATORY SYSTEM in Proceedings of the 11th International Joint Conference on Biomedical Engineering Systems and Technologies (BIOSTEC 2018) (in press).

Beauchet O., Allali G., Sekhon H., Verghese J., GILLAIN S., Steinmetz J., Kressig R., Barden J., Szturm, T., Launay, C., Grenier S., Bhere L., Liu-ambrose T., Chester V., Callisaya M., Srikanth V., Leonaerd G., Cock A., Sawa R., Duque G., Camicioli R., Helbostad J., Petermans J., GUIDELINES FOR ASSESSMENT OF GAIT AND REFERENCE VALUES FOR SPATIOTEMPORAL GAIT PARAMETERS IN OLDER ADULTS; THE BIOMATHICS AND CANADIAN GAIT CONSORITUMS INITIATIVE. Frontiers in Human Neuroscience, 2017.

Maquet D., Lekeu F., Warzee E., GILLAIN S., Wojtasik V., Salmon E., Petermans J., Croisier J-L. GAIT ANALYSIS IN ELDERLY ADULT PATIENTS WITH MILD COGNITIVE IMPAIRMENT AND PATIENTS WITH MILD ALZHEIMER'S DISEASE: SIMPLE VERSUS DUAL TASK: A PRELIMINARY REPORT. Clinical Physiology & Functional Imaging, 2010.

Maquet D., Warzee E., GILLAIN S., Lekeu F., Wojtasik V., Salmon E., Petermans J., Croisier J-L. Dual task and gait analysis: comparison between patients with mild cognitive impairment, patients with Alzheimer's disease and control subjects. Parkinsonism & Related Disorders, 2008.

Warzee E., GILLAIN S., Lekeu F., Maquet D., Wojtasik V., Croisier J.-L., Salmon E., Petermans J. USEFULNESS OF THREE DIFFERENT TESTS TO ASSESS MOBILITY IN COGNITIVELY FRAIL ELDERLY: INTEREST OF DUAL TASK. Parkinsonism & Related Disorders, 2008.

European Geriatric Medicine
https://doi.org/10.1007/s41999-018-0061-3

REVIEW

Assessing gait parameters with accelerometer-based methods to identify older adults at risk of falls: a systematic review

S. Gillain[1] · M. Boutaayamou[2] · C. Beaudart[3,4] · M. Demonceau[5] · O. Bruyère[3,4] · J. Y. Reginster[3,4] · G. Garraux[6] · J. Petermans[1]

Received: 4 February 2018 / Accepted: 30 April 2018

Abstract

Purpose The purpose of this study was to perform a systematic review to assess the utility of accelerometric methods to identify older adults at risk of falls.

Methods The Preferred Reporting Item for Systematic review and Meta-Analysis (PRISMA) guidelines were followed during all steps of this systematic review. Cross sectional and longitudinal studies assessing gait parameters in older adults using accelerometric devices, and comparing groups based on the risk of falls or fall history were identified from studies published in the MEDLINE, SCOPUS and Cochrane Database of Systematic Reviews databases between January 1996 and January 2017. Study selection and data extraction were performed independently by two reviewers. The quality of the methodology used in the studies included was assessed using the Newcastle–Ottawa Scale.

Results In total, 354 references were identified through the database search. After selection, ten studies were included in this systematic review. According to the cross sectional studies, people who fall or are at risk of fall are slower, and walk with shorter steps, lower step frequency, worse stride and step regularity in terms of time, position and acceleration profiles. One longitudinal study suggests considering harmonic ratio of upper trunk acceleration in the vertical plane. Two other longitudinal studies highlight the importance of considering more than one gait parameter, and sophisticated statistical tools to discern older adults at risk for future fall(s).

Conclusion This systematic review essentially highlights the lack of available literature providing strong evidence that gait parameters obtained using acceleration-based methods could be useful to discern older people at risk of fall. Available literature is encouraging, but further high quality studies are needed to highlight the cross-sectional and longitudinal relationships between gait parameters and falls in older adults.

Keywords Review · Acceleration · Gait · Fall · Older adults

Abbreviations

AUC	Area under the curve
IMF	Intrinsic mode function
HR	Harmonic ratios
MLHR	Medio-lateral harmonic ratios
8-step MLHR	Medio-lateral harmonic ratios based on the 8-step method
NOS	Newcastle–Ottawa Scale
PCA	Principal component analysis
RMS	Root mean square
SD	Standard deviation
SSI	Step Stability Index

✉ S. Gillain
s.gillain@gmail.com

1 Geriatrics Department, University Hospital of Liège, Route de Gaillarmont, 600, Chênée, 4032 Liège, Belgium

2 INTELSIG Laboratory, Department of Electrical Engineering and Computer Science, University of Liège, Liège, Belgium

3 Department of Public Health, Epidemiology and Health Economics, University of Liège, Liège, Belgium

4 WHO Collaborating Centre for Public Health Aspects of Musculoskeletal Health and Ageing, Liège, Belgium

5 Rehabilitation and Movement Sciences Department, University of Liège, Liège, Belgium

6 Neurology Department, University Hospital of Liège, Liège, Belgium

Published online: 24 May 2018

Introduction

Rationale

Falls among community-dwelling older people are known to be a cause of injury, disability, functional decline, decreased quality of life and ultimately, death [1, 2]. Additionally and even in the absence of injury, falls lead to a fear of falling, which is in turn associated with an increased fall risk [3], reduced social or physical activities [4, 5] leading to the functional decline. In this context, one of the major issues for the clinician is to detect people at risk of falling. Several risk factors are established, including gait and balance disorders [6]. Indeed, a decrease in gait speed [7], decreased stride length [8] and increased step variability [9, 10] have been reported to be linked to a higher risk of fall or to a history of previous falls. As they have been shown to be reliable tools for human gait analysis [11, 12], also in older adults [13, 14], accelerometric methods could be useful to obtain relevant gait parameters among older adults. In light of the numerous advantages of using acceleration-based method [15], and thanks to technical progress, several authors have developed data processing methods that make it possible to obtain more sophisticated gait parameters [11, 16]. Moreover, while the accelerometric reference method published by Moe Nilssen uses accelerometers placed over the L3 spinous process [17], other studies published gait parameters obtained using accelerometers placed on different parts of the body. In this context, it could be interesting to know how these gait parameters obtained using acceleration-based methods could be useful to discern, among older community-living people, those who are at risk of fall.

Objective

We performed a systematic review of the literature in order to highlight the usefulness of gait parameters, obtained using accelerometer-based methods, to identify older adults at risk of falls.

Methods

The Preferred Reporting Item for Systematic review and Meta-Analysis (PRISMA) guidelines were followed during all steps of this systematic review [18].

Search strategy

The electronic databases MEDLINE (via Ovid), SCOPUS and Cochrane Database of Systematic Reviews were searched for studies published between January 1996 and January 2017. The authors choose MEDLINE (via Ovid) considering her scoop dedicated to several health related aspects. SCOPUS is chose considering her scoop and because of her main contributor is European while the contributor of MEDLINE is the American library medicine. Finally the Cochrane Database of Systematic Review was also explored in order to be sure non previous similar systematic review is published. The search strategy and the search terms used are detailed in Table 1.

Table 1 Search strategy for the systematic review

1. Acceleration (MeSH) or acceleration.mp (as keyword)
2. Accelerometry (MeSH) or accelerometry.mp (as keyword)
3. Accelerometer.mp (as keyword)
4. Acceleration (MeSH) or acceleration.mp (as keyword)
5. Old.mp (as keyword)
6. Aged (MESH) or middle aged (MeSH) or "Aged 80 and over" (MeSH) or older.mp (as keyword)
7. Aging (MeSH) or Aging.mp (as keyword)
8. Ageing.mp (as keyword)
9. Elderly.mp (as keyword)
10. Accidental falls (MeSH) or Fall.mp (as keyword) or Falls.mp (as keyword)
11. Falling.mp (as keyword)
12. Missteps.mp (as keyword)
13. Gait (MeSH) or gait disorder, neurologic (MeSH) or Gait.mp (as keyword)
14. Walk test (MeSH) or walking (MeSH) or walk.mp (as keyword)
15. Or/1–4
16. Or/5–9
17. Or/10–12
18. Or/13–14
19. And/15–18

Study selection

Study selection criteria

This systematic review focused on studies including community-dwelling people older than 65 years and using accelerometer-based methods to analyse gait parameters during walking tests. Studies that involved people with a health condition impairing usual walking (e.g. Parkinson disease or recent hip prosthesis) were excluded. Studies where gait parameters were measured during static or dynamic balance tests (e.g. the timed up and go test) or during walking on a treadmill or long-term measurements during daily life, or during physical activities were not considered. Only data obtained among older adults and from the accelerometric method were considered. Only studies published in English language were considered. The group size of the studies did not play a role at time of study selection but was considered during assessment of methodological quality.

Study selection process

First, and after removal of duplicates, SG and MD independently read the titles and, if necessary, the abstract of all references found through the search of all three databases to select relevant references for further detailed analysis. If there was disagreement regarding the selection of a paper, MB read the full text in order to take the final decision about the inclusion. In a second step, SG and MB independently read the full text of all articles, not excluded in the first stage, in order to select the studies to include in this review. Additional studies were identified through a manual search of the reference lists of all the selected articles.

Data extraction

Data were extracted independently by each of the two main reviewers (SG and MB) using a standardized data extraction form. Each reviewer systematically recorded the data extracted from each included paper. Then the data obtained were cross-checked by the other reviewer to ensure correct collection data. The following data were extracted: first author's name, year of publication, sociodemographic data (type of population, mean age, sex ratio), sample size, design (duration of follow-up and drop-outs in case of longitudinal studies, number of groups, description of groups including the number of subjects/group), material used to obtain gait parameters, gait parameters measured, walking condition(s). Finally, we contacted authors when relevant information concerning design was missing in the full text of the paper.

Assessment of methodological quality

The quality of the methodology used in the studies included was assessed using the Newcastle-Ottawa Scale (NOS) (for cross-sectional studies [19] or for cohort studies [20]). The quality score was calculated based on three categories: group selection, comparability between groups and outcome assessment. After initial agreement concerning the screening parameters, two reviewers (SG and CB) independently assessed each included study. Differences in scores between the two reviewers were resolved by consensus.

Results

Search strategy

A total of 354 studies were identified from the literature databases. Five duplicates were removed, and the remaining 349 references were screened for title and abstract by two independent reviewers (SG and MD). Ten studies met the inclusion criteria. Among these, three described results from the same study, of which only one was included. A manual search of the reference lists of relevant papers brought two additional studies to light. Thus, a final total of ten studies were included. The flowchart of the study selection is shown in Fig. 1. Concerning information requests sent to the authors of selected studies, five teams were contacted regarding inclusion or exclusion criteria in their study, and/or criteria used to define groups and/or data extraction.

Studies included in the review

This systematic review included seven cross-sectional studies [21–27] and three longitudinal studies [28–30]. The main characteristics of the studies included are presented in Table 2, including a brief summary of the major finding(s) for each study. The assessment of the methodological quality of the studies included is detailed in Table 3.

Cross-sectional studies

Study characteristics The cross-sectional studies included in this review were published between 1998 and 2015. The number of subjects recruited ranged from 16 to 134. The age of subjects included in the cross-sectional studies was from 65 to 93 years old. Three studies classified people according to their fall risk [23–25], while four studies classified par-

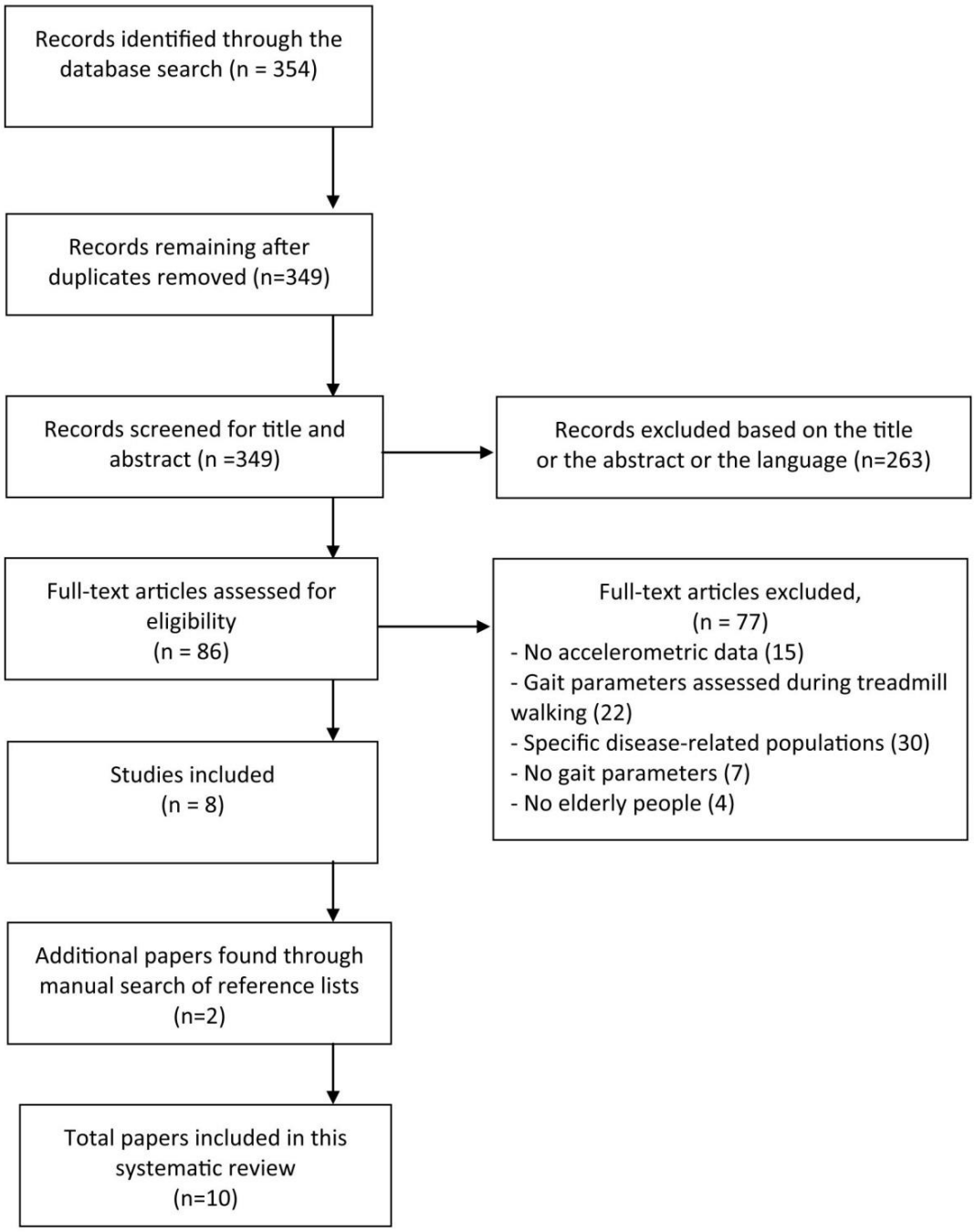


Fig. 1 Flowchart of studies considered for this systematic literature review

ticipants based on self-reported fall history [21, 22, 26, 27]. Two studies placed the accelerometers on the forehead and on the front of the sacrum [21, 23]. Five studies placed the accelerometer on the lower back [22, 24–27].

Usefulness of assessed gait parameters to detect fall risk Briefly, the gait parameters assessed in the included studies were gait speed [21, 23–27], step frequency [23, 25], stride frequency [22], step length [23, 25], stride length [22],

Table 2 Main characteristics of cross-sectional and longitudinal studies include in this systematic literature review

Author and year of publication	Design and inclusion or exclusion criteria	Sample size and sociodemographic data	Device and walking condition(s)	Gait parameter(s)	Summary of results
Cross-Sectional Studies					
Cho 1998 [27]	Two age-matched groups based on fall history Exclusion criteria walking aid needs, documented cardiopulmonary or neurological disease or orthopaedic injuries.	16 subjects included: 8 community-dwellings older. Age range 65–87 years, mean age = 72.6 years. Sex ratio (NA) 8 subjects referred by clinicians as frequent fallers. Age range 67–87, mean age = 76.3. Sex ratio (NA)	Two linear- accelerometers; one taped on the forehead and the other in front of S2 A 10-m walk at comfortable pace, following a straight line on the floor	Gait speed (m/s)	Non-fallers were faster than for fallers
Auvinet 2003 [28]	Two groups according to fall history Exclusion criteria: Musculoskeletal or neurological disorders. Taking no more than 3 therapeutic classes of medicine or neuroleptics	55 subjects included: 20 community dwelling people hospitalized for fall(s) with a mean of 3.6 ± 4.0 falls in the previous year. Mean age = 80.8 ± 5 years. 90% women 33 community-dwelling people. Mean age = 77.2 ± 6.5 years. 45% women	A tri-axial accelerometer placed on the L3–L4 intervertebral space A 20 s comfortable walk on a flat track of 40 m	Gait speed (m/s), Stride length (m), Stride frequency (stride/s), step symmetry (absolute value), Stride regularity (absolute value)	Fallers walked more slowly, with shorter stride length, slower stride frequency, lower symmetry and lower regularity
Menz 2003 [29]	Three groups according to fall risk based on vision, peripheral sensation, strength, reaction time and balance Falls retrospectively recorded as multiple fallers, once-only fallers and non-fallers Exclusion criteria: Parkinson's disease or short portable mental status questionnaire score less than 7	100 community-dwelling older adults included. Age range 75–93 years, mean age = 79.9 ± 4.0. 68% women 34 subjects included in the low risk group. Mean age = 78.94 ± 3.44 33 subjects included in the moderate risk group Mean age = 79.15 ± 3.83. 33 subjects included in the high risk group. Mean age = 81.48 ± 4.45 Sex ratio NA, but results did not show a difference in the proportion of men and women in the three groups	Two 3-axial accelerometers (one taped on a helmet and the other fixed at the sacrum) A 20-m walk in a 1.5 m wide flat corridor and an unpredictably irregular walkway.	Walking speed (m/s), cadence (steps/min), step length (cm), step-timing variability (absolute value), V, AP and ML Harmonic Ratio (absolute value)	High-risk fallers walked more slowly with shorter steps, lower cadence, greater step-time variability and smaller V-, AP- and ML-Harmonic Ratio than low-risk fallers on the level surface

Table 2 (continued)

Author and year of publication	Design and inclusion or exclusion criteria	Sample size and sociodemographic data	Device and walking condition(s)	Gait parameter(s)	Summary of results
Bautmans 2011 [23]	Two groups according to fall risk based on fall history and/or a timed up and go < 15 s and/or a score ≤ 24/28 on the Tinetti scale Exclusion criteria: MMSE score less than 23/30, unable to understand/perform instructions/procedures, unable to walk 20 m. without assistance, Parkinson's disease or cerebro-vascular accident with locomotor disability	81 community-dwelling olders included 40 increased-fall risk old people. Mean age 80.6 ± 5.4 years. 50% women 41 healthy old adults. Mean age = 79.1 ± 4.9 years. 49% women	Accelerometer placed on the sacrum Two 18 m walks at self-selected walking speed.	Gait speed (m/s), Step time asymmetry (%), Vertical and medio-lateral step and stride regularity	Older adults at risk of fall walked more slowly and with worse step and stride vertical and medio-lateral regularity than older adults without fall risk
Senden 2012 [31]	Three groups according to fall risk based on the Tinetti score: low-fall-risk (Tinetti > 24/28); moderate fall-risk (Tinetti 19–24) and high fall-risk group (Tinetti < 19) Inclusion criteria: aged 65 years or older, able to walk without assistive devices and no severe cognitive impairment	100 community-dwelling at least 65 years old 50 subjects included in the low fall risk group (Tinetti > 24). Mean age = 74 ± 5 years. 46% women. 48% with fall history 50 subjects included in the moderate and high fall risk group (Tinetti ≤ 24). Mean age = 79 ± 6 years. 66% women. 24% with fall history. Subjects in this group were also separated in two additional groups: Those with a Tinetti score between 19–24/28, mean age = 76 ± 5 years, 55% with fall history. And those with a Tinetti score < 19/28, mean age = 84 ± 5 years, 37% with fall history	A 3-axial accelerometer taped to the level of the sacrum A 20-m walk in a hospital corridor at preferred gait speed	Walking speed (m/s) Step frequency (step/s) Step length (m), Step time asymmetry (%) Based on vertical accelerations: root mean square, inter-stride amplitude variability and harmonic ratio	Walking speed, step length, step frequency, RMS, inter-stride amplitude variability and harmonic ratio were positively correlated with the Tinetti scale. A negative correlation was found with step time asymmetry. Walking speed, step length and RMS were more useful to discern people with higher Tinetti rather than inter-stride amplitude variability and harmonic ratios

Table 2 (continued)

Author and year of publication	Design and inclusion or exclusion criteria	Sample size and sociodemographic data	Device and walking condition(s)	Gait parameter(s)	Summary of results
Cui 2014 [32]	Two groups according to fall history. Subjects were defined as fallers if they experienced 2 or more falls in the previous year ($n=39$) Inclusion criteria: aged between 65 and 90 years, no clinically diagnosed balance or gait disorders, and a score > 24 on the MMSE	81 community dwelling older included 42 subjects included as controls. Mean age 79 ± 5 years. 62% women 39 subjects included as fallers. Mean age 78 ± 5 years. 59% women	A 3-axial accelerometer taped on the lower back Baseline walk (with and without a harness) and an obstacle course with a harness	Step stability index (SSI) based on the vertical acceleration signal from a 1 min sample	The SSI was significantly higher among non-fallers than among fallers during baseline walking and during obstacle walking
Brodie 2015 [33]	Two groups according to history of fall(s) Two groups according to fall-risk based on 4 factors. A high fall risk was based on age > 65 years, walking speed < 1.08 m/s, fall history, and Physiological Profile Assessment score ≥ 0.97 Exclusion criteria: Parkinson's disease or similar neurological condition, a Short Portable Mental Status Questionnaire score < 7	96 Community-dwelling older subjects 61 subjects included as non-fallers. Mean age = 80 ± 4 years. 64% women 35 subjects included as fallers. Mean age = 79 ± 4 years. 71% women	An accelerometer attached to the sacrum between the posterior superior iliac spines Two repeat walks at self-selected pace over 20 m excluding the first and last 2.5 m of the accelerations analysis.	Walking speed (m/s) Medio-lateral harmonic ratio (MLHR) (absolute value) The 8-Step MLHR (absolute value)	The lateral harmonic stability (8-step MLHR) and walking speed discerned fallers from non-fallers

Table 2 (continued)

Author and year of publication	Design and inclusion or exclusion criteria	Sample size and sociodemographic data and drop outs	Device and walking condition(s)	Gait parameter(s)	Summary of results
Longitudinal studies					
Marschollek 2011 [34]	1-year follow-up study Clinical outcome: fall incidence during the follow-up period Exclusion criteria: musculoskeletal disease, pain during the performance of daily living, fall event within the last month, vestibular problems, MMSE < 23/30.	119 community-dwelling people living at home, recruited at the end of a hospitalization in a geriatric ward. Mean age = 81 years. 74% women among those followed-up Drop outs: 69 subjects left the study during the follow-up period and 4 samples were unusable for technical reasons Among the 46 people for whom data were available, 19 people experienced a fall during follow up (41%)	A 3-axial accelerometer attached to the patient's waist with a custom belt A 20-m walk	Kinetic energy, pelvis sway in ML axis, standard deviation of gait periodicity, mean step duration, step length, number of peaks, frequency of the first peak Unity measures non accurate in the paper.	A classification tree combining information obtained from accelerometric signals, a long-term physical activity scale and history of fall(s), was able to discern fallers from non-fallers with a PPV of 92% (AUC = 0.83)
Doi 2013 [35}	1-year follow-up study Clinical outcome: fall incidence during the follow-up period Exclusion criteria: all neurological disorders. affecting gait	73 community-dwelling older subjects Among those, 16 people experienced a fall during follow-up (22%). Mean age of fallers = 85 ± 6 years, 94% women. Mean age of non-fallers = 79 ± 8 years, 74% women Drop outs: NA	Two 3-axial accelerometers (once taped in front of C7 and the other taped in front of L3 Normal pace on a 15-m smooth horizontal walkway with a 2.5-m space before each end of the walkway for acceleration and deceleration	Walking speed (m/s) HR-VT, HR-ML, HR-AP from upper trunk (absolute value) HR-VT, HR-ML, HR-AP from lower trunk (absolute value)	Fallers walked more slowly than non-fallers. HR-VT and HR-AP were lower in fallers than in non-fallers. The ROC curves of the HR of upper trunk accelerations in the vertical plane allow calculation of a cut-off value for HR-VT
Mignardot 2014 [36}	2-year follow-up study Clinical outcome: fall incidence during the follow-up period Inclusion criteria: age between 66 and 75 years, living at home, never fallen, able to walk without assistance for at least 30 s Fallers were classified according the occurrence of the first fall event	259 community-dwelling people. Age range 66–75 years. Mean age 69.5 ± 2.6, 59% women Among those, 72 people experienced a fall (28%). Subjects were classified into 4 groups according to fall period: non-fallers (n = 187); fallers from +0 to +6 months (n = 20); fallers from +6 to + 12 months (n = 26) and fallers from +12 to +24 months (n = 26)	One tri-axial accelerometer placed in the lumbar position A 30-m walk at self-selected walking speed in hospital corridor	22 gait variables were assessed including gait speed, stride length, stride duration, stride frequency, symmetry (numerous indices), regularity (numerous indices), and the power in the three directions. Features combining gait parameters were examined, where PC1 reflected global kinetics gait pattern, PC2 reflected global gait regularity and PC3 reflected stride time	Raw gait parameters were not shown. ROC marker PC1 showed an AUC = 0.7 (0.64–0.75) and the ROC marker PC2 showed an AUC = 0.67 (0.62–0.72)

NA not available; *V* vertical; *AP* antero-posterior; *ML* medio-lateral; *MMSE* Mini Mental State Examination; *RMS* root mean square; *MLHR* medio-lateral harmonic ratio; *PPV* positive predictive value; *AUC* area under the curve; *ROC* receiver operating characteristic; *HR-VT* vertical harmonic ratio; *HR-ML* medio-lateral harmonic ratio; *HR-AP* antero-posterior harmonic ratio; PC, principal component

Table 3 NOS scores for studies include in this systematic literature review

Study	Selection (5 stars)	Comparability (2 stars)	Outcomes (3 stars)	Total score (10 stars)
NOS scores for cross sectional studies				
Cho [21]	***	*	**	6 stars
Auvinet [22]	**	**	**	6 stars
Menz [23, 32]	***	*	**	6 stars
Bautmans [24]	**	**	***	7 stars
Senden [25]	***	*	**	6 stars
Cui [26]	***	**	**	7 stars
Brodie [27]	***	**	**	7 stars
Study	Selection (4 stars)	Comparability (2 stars)	Outcomes (3 stars)	Total score (9 stars)
NOS scores for longitudinal studies				
Marschollek [28]	***	–	**	5 stars
Doi [29]	***	**	**	7 stars
Mignardot [30]	***	**	**	7 stars

step-time variability [23], stride-to-stride variability [25], stride regularity [22], stride symmetry [22], harmonic ratio [23, 25, 27], step time asymmetry [24, 25], cranio-caudal step and stride regularity [24], medio-lateral step and stride regularity [24], root mean square [25], step stability index [26], medio-lateral harmonic ratio and 8-step medio-lateral harmonic ratio [27].

It should be noted that in the study reported by Senden et al. [25], the order of the columns in the table presenting the clinical characteristics of the subjects should be inversed. Indeed, the author has confirmed that in fact, the first column represents subjects with a Tinetti score > 24/28 (and not ≤ 24/28), while the second column actually corresponds to subjects with a Tinetti score ≤ 24/28 (and not > 24/28) (personal communication).

Gait speed Briefly, all studies included in this review showed that fallers or people at risk of falls were slower, irrespective of how gait speed was measured. Cho et al. [21], Auvinet et al. [22] and Cui et al. [26] reported that fallers walked significantly slower than non-fallers. Similarly, after defining three groups according to the fall risk, Menz et al. [23] showed a significant effect of the fall risk category on gait speed (after correction for subject's height). Finally, Senden et al. [25] and Bautmans et al. [24] also confirmed this relationship between a higher fall risk and a lower gait speed.

Step or stride frequency Auvinet et al. [22] showed that fallers had significantly lower stride frequency. Menz et al. [23] found no difference between groups in term of step frequency assessed in steps/min, whereas Senden et al. [25] showed that people with a Tinetti score ≤ 24/28 had slower step frequency than those with a Tinetti score > 24, while the people with Tinetti score between 19 and 24 show no difference in stride frequency compared with those who have a Tinetti score < 19.

Step or stride length Auvinet et al. [22] reported that fallers had shorter stride length than non-fallers. Senden et al. [25] showed that people with higher Tinetti score had longer step length than those with lower Tinetti score and Menz et al. [23] confirmed this relationship lower stride length and higher fall risk, even after correction for the subject's height.

Step-time variability Menz et al. [23] showed no difference between the three fall risk groups.

Step-time asymmetry Senden et al. [31] defined step-time asymmetry as:

$$\text{Step - time asymmetry} = \frac{(\text{Mean step time left leg}) - (\text{mean step time right leg})}{\text{Mean step time of both legs}}.$$

In [25], people with Tinetti score ≤ 24/28 had higher step-time asymmetry than those with a Tinetti score > 24/28, while subjects with a Tinetti score between 19 and 24 had no step-time asymmetry stride frequency difference compared to those with a Tinetti score < 19/28. Bautmans et al. [24], using the same definition of step-time asymmetry, confirmed the tendency of older adults with fall risk to have a higher step-time asymmetry (but the difference remained non-significant). These different results could be explained by possible right/leg length differences not considered in these studies. Thus, the relationship between step-time asymmetry and fall(s) risk or history remains poorly understood.

Cranio-caudal (or vertical) and medio-lateral step and stride regularities: These two parameters were computed by Bautmans et al. [24] as an unbiased autocorrelation coefficient according to Moe-Nilssen and Helbostad [11]. According to published box plots, older adults with fall risk showed worse vertical and medio-lateral step and stride regularities than older adults without fall risk.

Stride-to-stride variability Senden et al. [25] calculated the inter-stride amplitude variability based on vertical accelerations. In this study, people with a Tinetti score ≤ 24/28 had higher stride-to-stride variability than those with a Tinetti score > 24/28, while those with a Tinetti score between 19 and 24 had no difference in stride-to-stride variability compared to those with a Tinetti score < 19/28.

Step stability index (SSI) Cui et al. [26] defined the SSI as:

$$\mathrm{SSI} = \mathrm{SD(IMF4)}/(\mathrm{SD\ of\ IMF1} + \mathrm{SD\ of\ IMF2} + \mathrm{SD\ of\ IMF3}),$$

where SD of IMF1, 2, 3 and 4 are the standard deviations of the intrinsic mode functions 1, 2, 3 and 4 derived from acceleration signals in the vertical direction [33]. Cui et al. computed this SSI from the vertical signal of an accelerometer worn on the lower back. Preliminary results showed that this SSI was significantly higher ($p < 0.05$) in the controls than in those with a history of falls.

Stride regularity According to Auvinet et al. [22], stride regularity was automatically derived from two coefficients (C1 and C2) computed from an autocorrelation function on the cranial-caudal signal and quantifying the peak values of the first and second dominant periods of the autocorrelation function, respectively corresponding to the step regularity and to the stride regularity [11]. These coefficients were automatically transformed according to equations detailed in the paper of Auvinet et al. [12]. Stride regularity was calculated as $(C1 + C2) \times 100$ and describes the similarity of vertical movements over successive strides (dimensionless). In their study [22], Auvinet et al. showed that fallers had significantly lower stride regularity (191.3 ± 56.0) than non-fallers (291.9 ± 51.9) ($p < 0.001$).

Stride symmetry According to Auvinet et al. [12], stride symmetry is calculated as $(C1/C2)100$. This parameter describes the similarity of left and right cranial-caudal movements, and is independent of fluctuations in the successive movement of each limb. In [22], Auvinet et al. reported that fallers had significantly lower stride symmetry (173.4 ± 56.0) than non-fallers (210.9 ± 39.4) ($p < 0.01$).

V-, A-P- or M-L- Harmonic ratio Menz et al. defined harmonic ratios and detailed the methods for their calculation [32]. Briefly, the technique involves decomposing the acceleration signal into individual harmonics by use of a finite Fourier transform. The summed amplitudes of the even-numbered harmonics are then divided by the summed amplitudes of the odd-numbered harmonics, providing a harmonic ratio. Higher ratios represent a more stable walking pattern [23]. In [23], Menz et al. calculated harmonic ratio at the pelvis and at the head in vertical, antero-posterior and medio-lateral axes on a level surface for three groups based on the fall risk (low, moderate and high fall risk). A significant effect of fall risk group was shown on the harmonic ratio at the pelvis and in the three axes. The harmonic ratio at the head showed also a fall risk group effect in vertical and antero-posterior axes. In Senden et al. [25] showed than people with a higher Tinetti score had a significantly higher harmonic ratio in the vertical plane than those with a lower Tinetti score ($p < 0.01$). Moreover, Brodie et al. [25] focused on medio-lateral harmonic ratios (MLHR) calculated by two different methods (namely traditional MLHR and 8-step MLHR). In this study, a significant interaction between walking speed and lateral harmonic stability was observed using the 8-step MLHR, with the fast and stable subgroup 5.3 times less likely to be fallers than the other participants (relative risk 0.19, 95% confidence interval: 0.06–0.57). However, the medio-lateral harmonic ratio obtained using the traditional method was not useful to differentiate fallers and non-fallers.

Root mean square Root mean square, defined and detailed in [32], is a measure of dispersion of the data relative to zero and this value was used to provide an indication of the average magnitude of accelerations in each direction during a complete walking trial. Senden et al. [25] used the definition of Menz et al. and showed that root mean squares based on vertical accelerations were significantly higher in people with higher Tinetti scores than in those with lower Tinetti scores ($p < 0.01$ for Tinetti score ≤ 24/28 vs > 24/28, and $p = 0.03$ for Tinetti score 19–24 vs < 19/28).

NOS scores of cross sectional studies Regarding the assessment of methodological quality (see Table 3), all cross sectional studies included used satisfactory methodology and their results may be considered as robust. The lower scores obtained for the items under the heading "Selection" of the NOS are mainly linked to the representativeness of the sample, and the sample size of the studies.

Longitudinal studies

Study characteristics Three longitudinal studies were included, published between 2011 and 2014 and including from 73 to 259 subjects. Two studies had one-year follow-up [28, 29] and one study had a two-year follow-up period [30].

Usefulness of the gait parameters assessed to detect fall risk In the German study by Marschollek et al. [28], more than half the subjects dropped out (57%) and 4 gait assessment parameters were not available for analyses. After 1 year of follow-up, 46 sets of data were available for analyses. In

this study, the authors considered [1] parameters computed using spectral analysis of accelerometer data and [2] physical activity scores. The method used to calculate the spectral density distribution parameters of the accelerometric signal was developed by the authors and made it possible to obtain: the number of peaks, the frequency, width and relative prominence of the first peak, the frequency, width and relative prominence of the dominant peak and the number of peaks above a relative threshold. The authors derived a classification tree and logistic regression models based on these parameters and scores. The results showed the utility of considering several gait parameters at once or in classification tree where long-term physical activity is added. As shown in the results section, classification tree involving only gait parameter is useful to detect the fallers while best predictive model needed long term activity scores to be considered in addition to the gait parameters.

In the Japanese study [29], the authors assessed physical performance and gait parameters using one accelerometer placed on the upper trunk and another placed on the lower trunk to assess the vertical, medio-lateral and antero-posterior harmonic ratios at inclusion. After 1 year, the study population was classified into two groups based on their fall history during the follow-up period. The results showed that receiver operator characteristic curves of the harmonic ratio of upper trunk acceleration in the vertical plane were associated with fall(s) during follow-up, independently of physical performance.

During the follow-up of the French study [30], 72 people experienced a fall (28%). The authors used principal component analysis (PCA) of gait parameters (extracted from an accelerometer worn on the lower back) to discriminate non-fallers from those who fell [1] during the first 6 months, [2] between 6 and 12 months, and [3] during the second year. These gait parameters included: gait speed, stride length, stride duration and stride frequency, symmetry, regularity and mechanical power in the three axes. The authors considered the first three principal components (PC 1-3) to study the association between gait parameters and the fall risk. PC1 represents the "global kinetics of gait pattern", PC2 represents the "global gait regularity" and the PC3 is labelled "stride time". The original gait parameters used to calculate PC 1-3 were not shown. The results from a regression analysis showed that (1) PC1 discriminated the first fall onset during the first 6 months, (2) PC2 discriminated the fall status between the 6th and 12th months and (3) PC3 did not discriminate any fall onset during the whole follow-up period.

NOS scores of longitudinal studies Regarding the assessment of methodological quality, all the longitudinal studies included used satisfactory methodology and their results may be considered as robust (see Table 3).

Discussion

The purpose of this literature review was to examine the utility of gait parameters, obtained by accelerometric methods during walking, in the assessment of the risk of fall in community-dwelling older adults. In summary, according to the cross-sectional studies analysed here, fallers or people at risk of falls are slower [21, 23–25] and walk with shorter steps [23, 25], most often lower step frequency [23, 25], higher step-time asymmetry [24], less vertical and medio-lateral stride and step regularity [24], less SSI [26], less stride regularity and symmetry [22] with lower harmonic ratios [23, 25, 27] and lower root mean squares [25]. According to the longitudinal studies included in this review, the harmonic ratio of upper trunk acceleration in the vertical plane is associated with the risk of falling within the next 12 months, although the two other longitudinal studies suggest the interest of combining several gait parameters and considering sophisticated statistical tools to discern older people at risk of falls.

The studies included in our review used a range of different devices, placed on different parts of the body, and they did not all assess the same parameters. Nevertheless, the findings of this review are consistent across the studies included, and also across studies using different instrumental methods to obtain gait parameters. Indeed, the studies included showed that fall history or fall risk is associated with slower gait speed. These results are consistent with a recent systematic review focusing on gait speed (including studies assessing gait speed with an electronic walkway or a stopwatch or kinematic analysis) as a marker of fall-risk in older adults [7]. The relationship between reduced step length and fall(s) or fall-risk found in two of the studies included [23, 25] was confirmed by the KORA-age study, a population-based sample of 890 community-dwelling people aged 65–90 years, which showed that stride length (measured using an electronic walkway) was associated with fall risk in older men [33].

Auvinet et al. [22], Menz et al. [23] and Senden et al. [25] reported that non-fallers have a faster cadence than fallers or people at risk of fall. A recent systematic review investigating gait parameters assessed using different instrumental methods, showed that the link between cadence and falls is not consensual. Actually, in Mortaza review [34], and among the four studies assessing cadence, one showed a positive relationship between a history of fall(s) and cadence, while a second study found an inverse relationship, and two studies failed to find any difference between fallers and non-fallers regarding cadence. Moreover, in the longitudinal study of Verghese et al. (2009), the prospective relationship between cadence and future fall(s) was not significant in the full adjusted model (RR: 1.066,

CI 0.984–1.155). Considering that the underlying relationship between cadence or stride frequency and fall(s) risk and/or fall history remains unclear in cross sectional and longitudinal studies, different hypotheses could be formulated: either stride frequency has a non-linear relationship with fall(s), or their relationship could be mediated by a third component that is unrecognized at this time (e.g. gait speed, or stride regularity); or stride frequency has no relationship with fall(s) risk or/and fall(s) history. Moreover, previous studies did not take into account the influence of gender and/or leg length or total body height in their comparison groups introducing bias in the interpretation of their results. Sex ratio and mean leg length could explain these different results.

The link between step time variability and the fall risk observed by Menz et al. [23] is similar to that observed by Hausdorff et al. [9] in a one-year follow-up study including 52 community-dwelling adults older than 70 years.

Unfortunately, some of the parameters reported in this review are available only using accelerometric devices (e.g. harmonic ratio, regularity, symmetry, asymmetry, root mean square, step stability index) restricting the comparison. Nevertheless, the results of studies assessing either step/stride regularity or step/stride variability, are consistent and showed a strong link between gait variability and fall risk, as in the study of Herman et al. [35] in which 262 community-living healthy older adults wearing force-sensitive insoles (for gait speed and gait variability quantification) confirmed the strong link between fall(s) and gait variability.

Concerning the utility of considering gait parameters obtained with accelerometric methods and the risk of future fall(s), one longitudinal study tends to confirm that lower gait speed and lower harmonic ratio in the vertical plane could be considered as markers of fall risk in community-living older people, while two longitudinal studies underlined the utility of considering several gait parameters, and of implementing these in the form of classification tools. In this context of predictive risk quantification, classification trees or algorithms seem to be powerful and useful in clinical applications. In our view, the main advantage of classification tools is that they make it possible to accurately predict the individual fall risk taking into account of the heterogeneity of older adults.

In our opinion, the main strengths of this systematic review are—the step by step following PRISMA guidelines—the study selection and data extraction independently made by two reviewers (and the mediation by a third reviewer in case of disagreement), and the systematic contact to the corresponding authors of the studies considered for inclusion in case of missing data related to inclusion criteria or results. Unfortunately, this literature review has some limitations. First, the search was limited to published studies, meaning that some research published in journals or publications not cited in official databases may have been omitted. Second, the quality and reliability of the studies included in this review could be questionable. Indeed, despite satisfactory scores on the NOS, the methodological quality of the studies included is not equivalent in terms of sample size, clinical characteristics assessed or dropout rates. Furthermore, regarding the study by Marschollek et al. met the inclusion criteria for eligibility in this review and considering the innovative data mining involved in this longitudinal study, we chose to include it in this review. However, given the possible confounding factors linked to fall incidence and related to health problems and immobilisation, the incidence of falls described in Marschollek's study and its relationship with gait performance should be interpreted with caution. In addition, the relationship between some gait parameters and the fall risk or fall history is supported by one to three studies, which does not provide sufficiently strong validation for their use in identifying fall risk. A third limitation is that we authors chose not to include data obtained from long term recordings, or from treadmill walk assessments. This choice reduced the range of studies included, but facilitated comparison across those studies that were included. Indeed, it seems logical not to amalgamate in a single analysis gait parameters obtained during long term recordings (often performed in very different environments and conditions), and gait parameters obtained in a laboratory setting; firstly, because the environment could be very different (between and within subjects) during daily-life acquisition, as compared to laboratories, where the environmental conditions were constant; and second, because supervision can influence gait performance when gait tests are organised in a gait laboratory, as previously shown by Foucher et al. [36] and explained by the Hawthorne effect [37] (i.e., a change in behaviour in response to observation and assessment). Gait parameters obtained during a walk on a treadmill would likely be significantly different from the same parameters obtained in the same subject at self-selected walking speed on the ground as shown by Rispens et al. [38]. Nevertheless, considering the recent integration of accelerometer-based methods into the insoles of shoes, smart clothing or smart phones, accelerometric data obtained in daily-life conditions could represent a more physiological approach to assessing spontaneous mobility and walking in older adults, whatever the setting. Undoubtedly, there is an opening for a literature review considering the utility of accelerometric data obtained in daily-life conditions.

In view of the search strategy applied and the consistency of the results shown in the studies included, the relevance of accelerometric methods applied in gait assessment in order to discern older adults at risk of falls could be considered. However, considering the few available studies supporting strong evidence, further studies including larger sample and using high quality methodology are necessary to confirm the

prospective relationship between gait parameters obtained using acceleration-based methods and falls, and allowing their use as marker of risk in clinical setting. In our opinion, based on the studies included, this systematic review does not allow to edit complete recommendations. Actually, this literature review highlights the potential interest to consider gait speed, stride length, the parameter assessing the stride-to stride variability and the harmonic ratio in the vertical plan in order to detect people at risk for fall. However, studies included show gait parameters obtained using accelerometer worn on forehead or on trunk or on lower back or front of the sacrum, while none study showing gait parameter obtained using accelerometer worn on shoes is included. So then, in our opinion, the method of further studies should consider the need to accurate the best segment of the body to worn accelerometers (one place and whish place or a combination of different places) and to specify the parameters to assess considering the place they are obtained.

Conclusion

In summary, this systematic review suggests the usefulness of gait assessment using accelerometer-based data to identify older people at risk of falls. However, prospective studies with high methodological quality are needed to confirm whether relevant gait parameters could be used as markers of fall risk in older adults and how to integrate them to better assess the individual risk of fall. Considering the public health burden related to fall incidence among older adults, this research area deserves to be encouraged and supported.

Acknowledgements S.G. is supported by a fellowship from the FNRS (Fonds National de la Recherche Scientifique de Belgique—FRS-FNRS—www.frsfnrs.be). The authors would like to thank Mrs. Françoise Pasleau for advice on search strategies.

Compliance with ethical standards

Conflict of interest All authors declared that they have no conflict of interest.

Ethical approval This article does not contain any studies with human-participants or animals performed by any of the authors.

Informed consent For this type of study formal consent is not required.

References

1. Rubenstein LZ (2006) Falls in older people: Epidemiology, risk factors and strategies for prevention. Age Ageing 35:ii37–ii41
2. Thiem U, Klaaßen-Mielke R, Trampisch U, Moschny A, Pientka L, Hinrichs T (2014) Falls and EQ-5D rated quality of life in community-dwelling seniors with concurrent chronic diseases: a cross-sectional study. Health Qual Life Outcomes 12:2
3. Scheffer AC, Schuurmans MJ, van Dijk N, van der Hooft T, de Rooij SE (2008) Fear of falling: measurement strategy, prevalence, risk factors and consequences among older persons. Age Ageing 37(1):19–24
4. Delbaere K, Crombez G, Vanderstraeten G, Willems T, Cambier D (2004) Fear-related avoidance of activities, falls and physical frailty. A prospective community-based cohort study. Age Ageing 33(4):368–373
5. Choi K, Ko Y (2015) Characteristics associated with fear of falling and activity restriction in South Korean Older adults. J Aging Health 27(6):1066–1083
6. Ambrose AF, Paul G, Hausdorff JM (2013) Risk factors for falls among older adults: a review of the literature. Maturitas 75(1):51–61
7. Menant JC, Schoene D, Sarofim M, Lord SR (2014) Single and dual task tests of gait speed are equivalent in the prediction of falls in older people: a systematic review and meta-analysis. Ageing Res Rev 16:83–104
8. MacAulay RK, Allaire TD, Brouillette RM, Foil HC, Bruce-Keller AJ, Han H et al (2015) Longitudinal assessment of neuropsychological and temporal/spatial gait characteristics of elderly fallers: taking it all in stride. Front Aging Neurosci 7:34
9. Hausdorff JM, Rios DA, Edelberg HK (1050) Gait variability and fall risk in community-living older adults: A 1-year prospective study. Arch Phys Med Rehabil 2001(82):1056
10. Allali G, Launay CP, Blumen HM, Callisaya ML, De Cock A-M, Kressig RW et al (2017) Falls, Cognitive Impairment, and Gait Performance: results From the GOOD Initiative. J Am Med Dir Assoc 18(4):335–340
11. Moe-Nilssen R, Helbostad JL (2004) Estimation of gait cycle characteristics by trunk accelerometry. J Biomech 37(1):121–126
12. Auvinet B, Berrut G, Touzard C, Moutel L, Collet N, Chaleil D et al (2002) Reference data for normal subjects obtained with an accelerometric device. Gait Posture 16(2):124–134
13. Hartmann A, Luzi S, Murer K, de Bie RA, de Bruin ED (2009) Concurrent validity of a trunk tri-axial accelerometer system for gait analysis in older adults. Gait Posture 29(3):444–448
14. Hartmann A, Murer K, de Bie RA, de Bruin ED (2009) Reproducibility of spatio-temporal gait parameters under different conditions in older adults using a trunk tri-axial accelerometer system. Gait Posture 30(3):351–355
15. Kavanagh JJ, Menz HB (2008) Accelerometry: a technique for quantifying movement patterns during walking. Gait Posture 28(1):1–15
16. Moe-Nilssen R, Helbostad JL (2005) Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. Gait Posture 21(2):164–170
17. Moe-Nilssen R (1998) A new method for evaluating motor control in gait under real-life environmental conditions. Part 1: the instrument. Clin Biomech 13(4):320–327
18. Moher D, Liberati A, Tetzlaff J, Altman DG (2010) Preferred reporting items for systematic reviews and meta-analyses: the PRISMA statement. Int J Surg 8(5):336–341
19. (2018) The modified Newcastle Ottawa scale for cross sectional studies. http://journals.plos.org/plosone/article/file?type=supplementary&id=info:doi/10.1371/journal.pone.0136065.s004
20. Newcastle (2017) Ottawa quality assessment scale cohort studies [cited 2017 07-10]. http://www.ohri.ca/programs/clinical_epidemiology/nosgen.pdf
21. Cho CY, Kamen G (1998) Detecting balance deficits in frequent fallers using clinical and quantitative evaluation tools. J Am Geriatr Soc 46(4):426–430

22. Auvinet B, Berrut G, Touzard C, Moutel L, Collet N, Chaleil D et al (2003) Gait abnormalities in elderly fallers. J Aging Phys Act 11(1):40–52
23. Menz HB, Lord SR, Fitzpatrick RC (2003) Acceleration patterns of the head and pelvis when walking are associated with risk of falling in community-dwelling older people. J Gerontol Ser A Biol Sci Med Sci 58(5):M446–M452
24. Bautmans I, Jansen B, Van Keymolen B, Mets T (2011) Reliability and clinical correlates of 3D-accelerometry based gait analysis outcomes according to age and fall-risk. Gait Posture 33(3):366–372
25. Senden R, Savelberg HH, Grimm B, Heyligers IC, Meijer K (2012) Accelerometry-based gait analysis, an additional objective approach to screen subjects at risk for falling. Gait Posture 36(2):296–300
26. Cui X, Peng CK, Costa MD, Weiss A, Goldberger AL, Hausdorff JM (2014) Development of a new approach to quantifying stepping stability using ensemble empirical mode decomposition. Gait Posture 39(1):495–500
27. Brodie MAD, Menz HB, Smith ST, Delbaere K, Lord SR (2015) Good lateral harmonic stability combined with adequate gait speed is required for low fall risk in older people. Gerontology 61(1):69–78
28. Marschollek M, Rehwald A, Wolf KH, Gietzelt M, Nemitz G, Zu Schwabedissen HM et al (2011) Sensor-based Fall risk assessment – an expert 'to go'. Methods Inf Med 50(5):420–426
29. Doi T, Hirata S, Ono R, Tsutsumimoto K, Misu S, Ando H (2013) The harmonic ratio of trunk acceleration predicts falling among older people: results of a 1-year prospective study. J Neuroeng Rehabil 10:7
30. Mignardot JB, Deschamps T, Barrey E, Auvinet B, Berrut G, Cornu C et al (2014) Gait disturbances as specific predictive markers of the first fall onset in elderly people: a two-year prospective observational study. Front Aging Neurosci 6:22
31. Senden R, Grimm B, Heyligers IC, Savelberg HHCM, Meijer K (2009) Acceleration-based gait test for healthy subjects: reliability and reference data. Gait Posture 30(2):192–196
32. Menz HB, Lord SR, Fitzpatrick RC (2003) Acceleration patterns of the head and pelvis when walking on level and irregular surfaces. Gait Posture 18(1):35–46
33. Thaler-Kall K, Peters A, Thorand B, Grill E, Autenrieth CS, Horsch A et al (2015) Description of spatio-temporal gait parameters in elderly people and their association with history of falls: results of the population-based cross-sectional KORA-Age study. BMC Geriatr 15:32
34. Mortaza NAON (2014) Mehdikhani N Are gait parameters capable of distinguishing a faller than a non-faller elderly? Eur J Phys Rehabil Med 50(6):677–691
35. Herman T, Mirelman A, Giladi N, Schweiger A, Hausdorff JM (2010) Executive control deficits as a prodrome to falls in healthy older adults: a prospective study linking thinking, walking, and falling. J Gerontol Ser A Biol Sci Med Sci 65A(10):1086–1092
36. Foucher KC, Thorp LE, Orozco D, Hildebrand M, Wimmer MA (2010) Differences in preferred walking speeds in a gait laboratory compared with the real world after total hip replacement. Arch Phys Med Rehabil 91(9):1390–1395
37. Sedgwick P, Greenwood N (2015) Understanding the Hawthorne effect. BMJ 351:h4672. https://doi.org/10.1136/bmj.h4672
38. Rispens SM, Van Dieën JH, Van Schooten KS, Cofré Lizama LE, Daffertshofer A, Beek PJ et al (2016) Fall-related gait characteristics on the treadmill and in daily life. J Neuroeng Rehabil 13:12

Aging Clin Exp Res
DOI 10.1007/s40520-017-0730-y

ORIGINAL ARTICLE

Data set of healthy old people assessed for three walking conditions using accelerometric and opto-electronic methods

S. Gillain[1] · M. Boutaayamou[2] · N. Dardenne[3] · C. Schwartz[4] · M. Demonceau[5] · C. Gerontitis[1] · F. Depierreux[6] · E. Salmon[6] · G. Garraux[6] · O. Bruyère[7] · O. Brüls[4] · J.-L. Croisier[4] · J. Petermans[1]

Received: 16 November 2016 / Accepted: 20 January 2017

Abstract

Background Gait patterns of healthy aging are needed to allow a comparison with pathological situations. However, little data is available.

Objective To present gait pattern of healthy older specially selected to be "healthy walkers".

Method Fifty-seven older people benefited from a geriatric assessment including clinical and functional evaluations to include only those without gait disorders. Gait data were simultaneously recorded using a tri-axial accelerometer placed on the waist and four 3D position markers placed on the feet at the level of the heel and the toe. Volunteers walked at comfortable self-selected speed (CW), fast self-selected speed (FW), and finally in dual task walking condition (DTW). The extracted gait parameters were: gait speed, stride length, stride frequency, regularity and symmetry, swing, stance and double support time and ratio and minimum toe clearance. Gait speed and stride length were normalized to the right leg length.

Results Fifty-seven older people with a mean age of 69.7 ± 4.2 years old (range from 65 to 82 years) were included. Data were analyzed according to the gender and according to the age (<70 or ≥ 70 years old). After normalization to leg length, the main significant differences were shown for stride length and minimum toe clearance in CW, FW and in DTW that were shorter in women. The regularity in FW was significantly lower among older volunteers.

Conclusions This work provides a data set considering 14 gait parameters obtained from 57 healthy old people strictly selected and assessed for three walking conditions and shows that GS, SL and MTC have to be related to the gender. The age-related impact on gait performances appears reduced in this cohort.

Keywords Gait · Kinematics · Accelerometers · Reference values · Comfortable · Fast · Dual task

Abbreviations

CW	Comfortable walking condition
FW	Fast walking condition
GS	Gait speed
DSR	Double support ratio
DST	Double support time
DTW	Dual task walking condition
MTC	Minimum toe clearance
NGS	Normalised GS
NSL	Normalised SL
REG	Regularity
StR	Stance ratio
SwR	Swing ratio
StT	Stance time

✉ S. Gillain
sgillain@chu.ulg.ac.be

1 Geriatric Department, University Hospital of Liège, C.H.U. site NDB, Route de Gaillarmont, 600, 4032 Chênée, Belgium

2 INTELSIG Laboratory, Department of Electrical Engineering and Computer Science, University of Liège, Liege, Belgium

3 Public Health Department, University of Liège, Liege, Belgium

4 Laboratory of Human Motion Analysis, LAMH, University of Liège, Liege, Belgium

5 Science of Motricity Department, University of Liège, Liege, Belgium

6 Neurology Department, University Hospital of Liège, Chênée, Belgium

7 Research Unit in Public Health, Epidemiology and Health Economics, University of Liège, Liege, Belgium

Published online: 28 February 2017

SwT Swing time
SF Stride frequency
SL Stride length
SYM Symmetry

Introduction

Trunk accelerometers have been described as reliable tools for human gait analysis outside the laboratory (i.e. in clinical environment) [1–4]. They offer many advantages such as an easy handling and a low-cost technology. Furthermore, pilot studies have previously shown their accuracy to differentiate the gait profiles from cognitively healthy and cognitively impaired older people [5, 6]. Unfortunately, reference data obtained with this method for healthy older people and comparison with gold standard laboratories methods like optoelectronic systems are still lacking. Nevertheless and considering the growing interest using the gait pattern as a marker of risk of negative clinical outcomes or as a marker of robustness and to use this to suggest specific drugs, intervention or rehabilitation, the development of such database seems essential as a reference to pathological conditions analysis. Several impediments explain this lack of available literature.

A few papers show reference data but some important clinical information is missing (e.g. functional status, cognitive performance, mood disorders, medications, and previous falls) to check the global and complete health status of the patients. Additionally, in several papers about specific gait conditions (e.g. Parkinson's disease), control volunteers are selected based on the absence of the pathology of interest but mostly without considering health-related criteria. In 2011, Hollman et al. published a notable reference data set showing 23 gait parameters from healthy older adults assessed in comfortable walking condition (CW) and using a 5.6 m electronic walkway [7]. However, some motion and gait analysis laboratories are equipped with other instrumental systems, then showing gait data obtained with other instrumental methods could be useful and could allow comparing the results from different data acquisitions and data process. Another issue is the lack of mention or adjustment to the height, known to influence gait speed and stride length [8, 9]. Finally, some papers highlight the importance to consider not only the CW but also the dual task walking condition (DTW) and the fast walking condition (FW). Considering all these aspects, the aim of this study is to assess the gait pattern of healthy old people during these three clinically relevant walking conditions (CW,FW and DTW) using simultaneously an accelerometer and an opto-electronic method and to show gait parameters normalized to the leg length. This study would provide accurate gait pattern of healthy old people and help other researchers to compare these data to their own. Finally, the authors will compare their results to those of Hollman et al.

Methodology

Population

Volunteers were selected from a prospective study aiming to clarify the interest of gait analysis associated to brain imaging to earlier discern old people at risk of cognitive decline and falls, the Gait Analysis and Brain Imagery Study (GABI Study).

Inclusion criteria

To be at least 65 years old, living independently at home, to understand French, to be able to reach the motion analysis laboratory, to approve inform consent.

Exclusion criteria

Exclusion criteria included fall in the previous year, history of stroke, physiotherapeutic cares, recent hip or knee prosthesis (≤1 year), musculoskeletal pain during walking, Timed Up and Go (TUG) longer than 10 s, an acute respiratory or cardiac illness (<6 month), cognitive disorders according to the MoCA score (<26/30), mood disorder according GDS-4 (people with a score of 2/ were included were the score was linked to the status of proxy and the fear to become ill because of this responsibility of caregiving), frailty according to the Edmonton score (>5/17) and neoplasm (people with a cancer in remission since more than 5 years were included).

Clinical and functional assessment

Subjects were assessed for socio-demographic and anthropometric data and benefited from a complete functional assessment (Table 1). The ADL and IADL scale scores were calculated to confirm the independency of included volunteers. Considering that some housework are usually and preferentially done by the same member of the family (e.g. the gardening or the meals), the IADL score was calculated by considering the sum of the results obtained in the items applicable to each subject, divided by the sum of the maximum possible scores in the applicable items (to not consider the volunteer disable for a housework he or she never did).

Table 1 Main clinical and functional characteristics of the population

Variables	Mean ± SD	P25	Median	P75	Min	Max
Age (years)	69.7 ± 4.2	67.0	68.0	71.0	65.0	82.0
Katz (/24)	6.2 ± 0.4	6.0	6.0	6.0	6.0	7.0
Lawton	1.0 ± 0.05	1.0	1.0	1.0	0.6	1.0
GDS (/4)	0.5 ± 0.7	0.0	0.0	1.0	0.0	2.0
COVI (/15)	3.1 ± 0.6	3.0	3.0	3.0	3.0	6.0
MNA (/14)	13.1 ± 1.3	12.0	14.0	14.0	10.0	14.0
BMI (Kg/m^2)	25.8 ± 3.6	23.3	25.5	28.4	19.0	34.4
Edmonton (/17)	1.7 ± 1.2	1.0	2.0	2.0	0.0	5.0
MOCA (/30)	28 ± 1.2	27.0	28.0	29.0	26.0	30.0
FGA (/30)	27.7 ± 2.2	27.0	28.0	29.0	22.0	30.0
SPPB (/12)	10.9 ± 1.1	10.0	11.0	12.0	9.0	12.0

SD standard deviation, *P25* percentile 25, *P75* percentile 75, *Min* minimal value, *Max* maximal value

Materials

During walking test all subjects were assessed using simultaneously two validated methods to collect data; a triaxial accelerometer (Locometrix®) and an optoelectronic device (CodaMotion®). Combined together, these two methods are used to assess 14 parameters frequently shown in the literature and clinically relevant regarding the adverse clinical age-related outcomes (e.g. the fall risk). To our knowledge, this is the first published work using simultaneously these two methods.

The accelerometric system (Locometrix®, Centaure Metrix, Evry, France) is a validated method [2] comprising an acceleration sensor, a recording device and a computer program for processing the acceleration signal. The acceleration sensor is composed of three accelerometers placed perpendicularly to each other in a plastic box with the recording device. The sensor's box is incorporated in an elastic abdominal belt, behind the back over the L3–L4 intervertebral lumbar space (the 3rd lumbar vertebra level). The acquisition frequency of the signal is 100 Hz.

The Codamotion® system (Charnwood Dynamics, Rothley, UK) is a 3-dimensional kinematic tool validated for its use in laboratory [10–12]. The 3-dimensional position and orientation of the feet were tracked using four Codamotion CX1 units at a sampling rate of 200 Hz. Two position markers were attached on each shoe of the volunteers; one on the heel and one on the toe. These four position markers give accurate measurements of the position of the feet in 3D in the Codamotion® system covers 4 to 5 stride cycles (depending on the subject and the walking condition). See Fig. 1.

Gait tests

Clinical and functional assessments were realized by the same examiner. Subjects were asked to wear comfortable shoes. After familiarization and once the subject equipped, gait was recorded under three different experimental conditions: self-selected comfortable walking speed (CW) (to assess spontaneous gait pattern), self-selected fast walking speed (FW) (because its relevance considering the relationship to the risk of cognitive decline [13, 14] and disability [15]) and during dual-task walking condition (DTW) (because of the spread brain areas and neuronal networks implicated in gait disturbances and associated to normal aging [16] and considering his relationship to the cognitive decline [5, 6] and to the disability [17]). The instruction

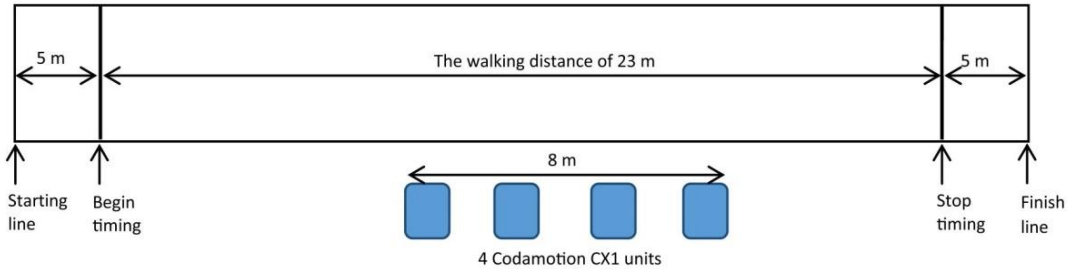


Fig. 1 Plan and organization of the motion laboratory and walking distances the volunteers has to walk for each walking condition (CW, FW and DTW) in one way

for FW was "walk quickly without running in order not to miss your train" and for DTW "walk at a comfortable walking speed while simultaneously making arithmetic (seven counting down since the number "100" to increase gait pattern modifications [18]). Because explicit instruction concerning the prioritization could affect dual tasking [19], the single instruction gave to the volunteers was "make both tasks simultaneously and as well as you can". The CW was always the first walking condition done. The order of fast and dual-task walking was randomized to avoid a systematic measurement error. The starting point was on the left side of the Fig. 1. The volunteer has to walk 5 m to increase their walking speed and reach a walking steady state. The examiner starts the stopwatch when the first foot of the volunteer cross the time line. Time needed to walk 23 m is assessed. The last 5 m are used to decrease the walking gait and stop. In the middle of the 23 m, four position cameras allow computing the position of heels and toes on 8 m.

Data processing

Concerning Locometrix®, the mean gait speed (GS) was computed from the time needed to cover 23 m. Time was recorded by means of a stopwatch. The following gait variables were inferred from the walking speed and 20.48 s of stationary cranial-caudal acceleration signal:

- the stride frequency (SF) or number of cycles per second (Hz) was calculated from the cranio-caudal acceleration following the application of the Fourier transform,
- the stride length (SL) was deduced from the equation [speed (m/s) = frequency (Hz) × stride length (m)] and expressed in meters.
- the stride regularity and symmetry were automatically derived from two coefficients (C1 and C2) computed from an autocorrelation function on the cranial-caudal signal. The autocorrelation coefficients C1 and C2 quantify the peak values of the first and second dominant periods of the autocorrelation function, respectively, corresponding to the step regularity and to the stride regularity [1]. These coefficients are automatically transformed according to equations detailed in the paper of Auvinet et al. [2]. Regularity and symmetry indexes are dimensionless. The regularity index describes the similarity of vertical movements over successive strides. The symmetry index describes the similarity of left and right cranial-caudal movements and it is independent of fluctuations in the successive movement of each limb.

The Codamotion® system is used to measure reference kinematic signals: the vertical (horizontal) heel position Z_h (X_h) and the vertical (horizontal) toe position Z_t (X_t). A signal-processing algorithm is then applied to these signals to extract—during consecutive strides—gait events of importance: the heel strike (HS) and the toe off (TO). This algorithm is based on a piecewise linear fitting method [20] that identifies accurately HS and TO in local position signals. Temporal gait parameters are extracted based on these gait events as follows:

- Stance Time (StT) (time between HS and TO during stride i) = $\mathrm{TO}(i) - \mathrm{HS}(i)$.
- Swing Time (SwT) (time between TO of stride i and HS of stride $i+1$) = $\mathrm{HS}(i+1) - \mathrm{TO}(i)$.
- Right double support duration [time between left HS ($\mathrm{HS_{left}}$) and right TO ($\mathrm{TO_{right}}$) during stride i] = $\mathrm{TO_{right}}(i) - \mathrm{HS_{left}}(i)$.
- Left double support duration [time between right HS ($\mathrm{HS_{right}}$) and left TO ($\mathrm{TO_{left}}$) during stride i] = $\mathrm{TO_{left}}(i) - \mathrm{HS_{right}}(i)$.

The time of minimum toe clearance (MTC) is extracted as being the moment at which the toe reaches a local minimal vertical position during the swing phase.

In fact the Codamotion® system also allows extracting GS, SL and SF. However, considering the longer walking distance covered by the accelerometric method (23 m compare to 8 m using the Codamotion® system), the authors will consider and present GS, SL and SF measured using the accelerometric method.

Considering the potential confounder effect of the height on the gait speed and the stride length and to decrease the gender effect (considering that men are often higher than women), these gait parameters were normalized to the length of the right lower limb (measured to the trochanter until the heel) as the following formula: "normalized gait parameter = (gait parameter/right leg length) × 100".

Statistical analysis

Statistical analyses were performed using Statistica 13 (StataCorp. 2013. Stata Statistical Software: Release 13. College Station, TX: StataCorp LP.) and MATLAB R2013a (Math Works, Natick, MA). Clinical parameters were expressed as mean ± standard deviation (SD), median, percentile 25 (P25), percentile 75 (P75), minimal and maximal values. Gait parameters obtained considering the left and the right side (double support, stance and swing time) were expressed as the mean values of both sides. Left and right measures were previously checked to be not significantly different side by side using the Student's paired t test. Normality was tested using the Shapiro–Wilk test and investigating mean and median values, histograms and Quantile–Quantile plots. Homoscedasticity was tested using the Levenne test. Paired off groups were compared using the Welch–Satterthwaite t test. Results

were considered significant at the 5% critical level (p value <0.05). 95% confidence intervals (CI 95%) for means were also calculated for men and women. Intra-class coefficient correlation (ICC) (2, k) were additionally calculated for gait parameters both extracted by the Locometrix® and the Codamotion® system.

Results

Characteristics of the population

After applying severe exclusion criteria (see previously), 57 healthy old people were included. Table 1 shows the main clinical and functional characteristics of the 57 selected subjects, including 25 men (43.86%) and 32 women (56.14%). Mean age was 69.7 ± 4.2 years old. The group of volunteers <70 years old included 44 persons (with 25 women) and the group of volunteers ≥70 years old included 23 persons (with seven women). All volunteers realized the TUG faster than 10 s. Median values highlighted the high physical and functional performances of the group.

Gait parameters

The 14 gait parameters obtained from CW including "normalized gait parameters" are presented considering the gender in the Table 2. During CW, the GS ($p=0.002$), the SL ($p<0.001$), the NSL ($p=0.002$) and the MTC ($p<0.001$) showed a significant difference according to the gender. Gait parameters obtained in FW are presented in the Table 3. During FW, the GS ($p<0.001$), the SL ($p<0.001$), the NSL ($p<0.001$), the SF ($p=0.019$), the SwT ($p=0.002$), the MTC ($p<0.001$) and the StT ($p=0.018$) showed a significant difference according to the gender.

Gait parameters obtained in DTW are presented in the Table 4. During DTW, the GS ($p=0.028$), the SL ($p=0.007$) and the MTC ($p<0.001$) showed a significant difference between groups according to the gender.

The REG in FW showed a significant difference according to the age ($p=0.049$) with a mean = 327 and a SD = 47 among volunteers <70 years old and with a mean = 301 and a SD = 48 among oldest group. Considering all walking conditions, no other parameter showed a significant clinical or statistical difference according to the age.

Concerning the consistency between gait parameters both extracted by the two methods. ICC (2, k) considering GS were 0.94 in CW, 0.91 in FW and 0.96 in DTW. ICC considering SL were 0.96 in CW, 0.94 in FW and 0.80 in DTW. ICC considering SF were 0.96 in CW, 0.87 in FW and 0.82 in DTW.

Discussion

Briefly, this work provides gait patterns of healthy old people selected with great care and assessed for three different walking conditions widely used in research and clinical settings. For each walking condition, 14 gait

Table 2 Gait parameters from healthy old people in CW according to the gender

Gait parameters (unit)	Women $N=32$ Mean ± SD	Men $N=25$ Mean ± SD	p value	CI 95% Women	CI 95% Men
Gait speed (m/s)	**1.26 ± 0.21**	**1.40 ± 0.13**	**0.002**	**1.20–1.31**	**1.37–1.44**
Normalized gait speed (1/s)	1.53 ± 0.25	1.61 ± 0.18	0.135	1.46–1.59	1.57–1.66
Stride length (m)	**1.29 ± 0.13**	**1.47 ± 0.11**	**<0.001**	**1.25–1.32**	**1.44–1.50**
Normalized stride length (dimensionless)	**1.56 ± 0.15**	**1.68 ± 0.12**	**0.002**	**1.52–1.60**	**1.65–1.72**
Stride frequency (stride/s)	0.97 ± 0.08	0.95 ± 0.06	0.471	0.95–0.99	0.94–0.97
Stride regularity (dimensionless)	310 ± 42	315 ± 41	0.698	300–321	304–325
Symmetry (dimensionless)	223 ± 61	214 ± 57	0.564	207–239	199–229
Swing time (s)	0.37 ± 0.02	0.38 ± 0.03	0.095	0.36–0.37	0.376–0.39
Swing ratio (% of the stride time)	35.78 ± 1.93	36.30 ± 1.36	0.251	35.28–36.29	35.95–36.66
Double support time (s)	0.15 ± 0.03	0.14 ± 0.02	0.387	0.14–0.15	0.14–0.15
Double support ratio (% of the stride time)	14.16 ± 1.95	13.56 ± 1.40	0.196	13.66–14.67	13.20–13.93
Min toe clearance (mm)	**15.47 ± 4.64**	**20.17 ± 3.63**	**<0.001**	**14.27–16.68**	**19.23–21.11**
Stance time (s)	0.66 ± 0.07	0.66 ± 0.04	0.903	0.65–0.68	0.65–0.68
Stance ratio (% of the stride time)	64.22 ± 1.93	63.70 ± 1.35	0.251	63.71–64.71	63.34–64 0.08

Data in bold indicates significant difference between groups

N number of volunteers, *SD* standard deviation, *CI 95%* 95% confidence interval

Table 3 Gait parameters from healthy old people in FW according to the gender

Gait parameters (unit)	Women $N=32$ Mean ± SD	Men $N=25$ Mean ± SD	p value	CI 95% Women	CI 95% Men
Gait speed (m/s)	**1.64 ± 0.17**	**1.84 ± 0.18**	**<0.001**	**1.60–1.69**	**1.79–1.89**
Normalized gait speed (1/s)	1.99 ± 0.24	2.11 ± 0.24	0.068	1.93–2.06	2.05–2.18
Stride length (m)	**1.47 ± 0.12**	**1.70 ± 0.16**	**<0.001**	**1.44–1.51**	**1.66–1.75**
Normalized stride length (dimensionless)	**1.78 ± 0.18**	**1.95 ± 0.18**	**<0.001**	**1.74–1.83**	**1.91–2.00**
Stride frequency (stride/s)	**1.12 ± 0.07**	**1.08 ± 0.04**	**0.019**	**1.10–1.14**	**1.07–1.09**
Stride regularity (dimensionless)	310 ± 49	326 ± 47	0.221	297–322	313–338
Symmetry (dimensionless)	226 ± 61	219 ± 42	0.628	210–242	208–230
Swing time (s)	**0.33 ± 0.02**	**0.35 ± 0.02**	**0.002**	**0.32–0.33**	**0.34–0.35**
Swing ratio (% of the stride time)	37.83 ± 1.70	37.76 ± 1.26	0.881	37.38–38.27	37.44–38.10
Double support time (s)	0.10 ± 0.02	0.11 ± 0.01	0.213	0.10–0.11	0.11–0.12
Double support ratio (% of the stride time)	12.01 ± 1.85	12.15 ± 1.27	0.745	11.53–12.50	11.82–12.48
Min toe clearance (mm)	**16.19 ± 5.66**	**23.69 ± 6.28**	**<0.001**	**14.72–17.66**	**22.06–25.32**
Stance time (s)	**0.54 ± 0.05**	**0.57 ± 0.04**	**0.018**	**0.53–0.56**	**0.56–0.58**
Stance ratio (% of the stride time)	62.17 ± 1.70	62.23 ± 1.26	0.881	61.73–62.62	61.90–62.56

Data in bold indicates significant difference between groups

N number of volunteers, *SD* standard deviation, *CI 95%* 95% confidence interval

Table 4 Gait parameters from healthy old people in DTW according to the gender

Gait parameters (unit)	Women $N=32$ Mean ± SD	Men $N=25$ Mean ± SD	p value	CI 95% Women	CI 95% Men
Gait speed (m/s)	**1.15 ± 0.23**	**1.29 ± 0.22**	**0.028**	**1.09–1.2**	**1.23–1.35**
Normalized gait speed (1/s)	1.40 ± 0.28	1.48 ± 0.26	0.280	1.33–1.47	1.41–1.55
Stride length (m)	**1.22 ± 0.14**	**1.44 ± 0.36**	**0.007**	**1.19–1.26**	**1.35–1.54**
Normalized stride length (dimensionless)	1.48 ± 0.17	1.66 ± 0.42	0.063	1.44–1.53	1.55–1.77
Stride frequency (stride/s)	0.93 ± 0.11	0.90 ± 0.08	0.275	0.90–0.96	0.88–0.93
Stride Regularity (dimensionless)	260 ± 76	267 ± 57	0.672	240–279	252–282
Symmetry (dimensionless)	234 ± 73	213 ± 58	0.228	216–253	198–228
Swing time (s)	0.38 ± 0.03	0.38 ± 0.03	0.573	0.37–0.39	0.38–0.39
Swing ratio (% of the stride time)	35.16 ± 2.65	35.57 ± 1.42	0.474	34.48–35.86	35.20–35.94
Double support time (s)	0.16 ± 0.04	0.16 ± 0.02	0.440	0.15–0.17	0.15–0.16
Double support ratio (% of the stride time)	14.74 ± 2.61	14.35 ± 1.37	0.484	14.06–15.42	13.99–14.70
Min toe clearance (mm)	**13.15 ± 4.44**	**17.80 ± 4.69**	**<0.001**	**12.00–14.31**	**16.59–19.02**
Stance time (s)	0.71 ± 0.11	0.70 ± 0.05	0.655	0.63–0.74	0.68–0.71
Stance ratio (% of the stride time)	64.83 ± 2.65	64.43 ± 1.42	0.474	64.15–65.52	64.06–64.80

Data in bold indicates significant difference between groups

N number of volunteers, *SD* standard deviation, *CI 95%* 95% confidence interval

parameters, chosen because of their clinical relevance, are presented considering gender. To not underestimate age-related gait modifications and according to the median age reached by this cohort, the authors divided the cohort in two groups; one including volunteers <70 years old and another group including those ≥70 years old. Furthermore, to decrease the gender effect due to the height, the most height-dependent-gait parameters like GS and SL were showed as "raw data" but also as "normalized data". Then researchers could easily compare this data set to the gait pattern obtained in their own cohort.

After a first look, an important observation to do is that the same gait parameters (GS, the SL, the normalized SL and the MTC) show a difference according to the gender in

the three walking conditions (except for the NSL in DTW). This observation underlies the relevance of these parameters and the reliability of the data acquisitions. In FW the SwT and the StT presented statistical differences between groups (p value <0.05), but these numerical differences remain skinny and clinically irrelevant. Of course, some other clinical variables could partially explain the observed difference between the groups but the normalization to right leg length and the strict selection of the volunteers, applying several exclusion criteria, has limited potentials confounders.

In CW, the GS, the SL, the SF obtained from trunk accelerations are comparable to those published by Auvinet et al. [2], Senden et al. [8, 9] and Ko et al. [21], where men are faster than women and they show longer stride length. However, applying the normalization to the leg length, this study showed a difference according to the gender only for SL (and normalized SL).

The REG and SYM are similar in both gender as previously showed by Auvinet et al., Senden et al. and Patterson et al. [22].

The MTC shown in this study are consistent with the MTC obtained by Dadashi et al. [23]. Nevertheless, the MTC showed in the Swiss study did not show a difference according to the gender. In our opinion, the overall precision of 9 mm of the inertial extraction method used (not allowing to measure a difference <9 mm as recorded in our sample) could explain these differences in the results shown.

After comparison with the results published by Hollman and even if he used another instrumental method, our results are similar. Indeed, in the Hollman study, men were faster than women and their stride length were longer. In the same line, the swing ratio and the stance ratio are similar in the two studies. The double support time and ratio expressed in the Hollman study express the sum of the double support ratio of the left and the right step. In our study, the double support time and ratio were assessed stride by stride, that could explain the difference (almost twice more) shown in the two studies. Actually Hollman and al. showed an influence of age on gait performances. Conversely this work did not find the same results. Two components could explain this inconsistency; first the small size of our cohort and the small number of very old volunteers (only two people older than 80) furthermore the lower proportion of women (slower than men and with shorter stride length) included in the ≥ 70 years old group could hide the age-related impact on gait speed and stride length. Nevertheless, the consistency of the two studies agrees with the reliability and the validity of the gait parameters assessed and the data acquisition protocol used.

In FW, our results are similar to those founded by Laufer using the GaitRite walkway system [24]. The DSR is lower in our cohort (around 14 and 12% of the stride time during CW and FW, respectively) than in the Laufer's cohort (30 and 25%, respectively). However, the reductions of DSR secondary to the increase of GS were similar in the two populations (reduction of 17% in Laufer's cohort and reduction of 15% in our cohort).

In DTW, and as found by van Iersel et al. [25], Hausdorff et al. [26] and Yogev-Seligmann et al. [19], GS decreases during dual tasking even for old people free of cognitive disorder. Interestingly in DTW, the normalized SL did not remain significantly different in both gender (like in CW and FW). Even if the mean values between the two groups are high, the higher SD in men than in women could explain this statistical result (related to the sample size effect). Unfortunately, this work has some limitations to take into account. The sample size was modest because of several exclusion criteria and the cost of the cognitive task was not accounted for. Finally, due to the limited number of available CX1 measurement units, only 8 meters of stabilized walk are available for gait analysis based on the opto-electronic method (meaning 4 or 5 gait cycles/each volunteer). Nevertheless, the excellent ICC concerning gait parameters measured by the two instrumental methods and the consistency between our results and those previously published (even those obtained using other instrumental methods) confirm the reliability of data acquisition and data analysis used in this work.

Finally, even considering to the small size of the cohort, the limited age-related gait modifications showed in this work is, in our opinion, an important fact to underline. Indeed, these results support the idea that people with gait disorders have to benefit at least of a clinical examination to detect some pathological condition linked to the gait performances and so even in case of advanced age.

Conclusion

This work provides a data set of gait parameters obtained from 57 healthy old people assessed for three walking conditions using a tri-axial accelerometer and an optoelectronic method allowing researchers to compare with their own data. In healthy old people, the gait speed, the stride length and the minimal toe clearance have to be related to the gender. The age-related impact on gait performances appears reduced in this cohort. The strict selection of the old people included and the consistency with previous published data allow considering the present data set as reference values.

Acknowledgements The authors would like to thank the Dr. Sophie Christelbach and Ms Vinciane Wojtasik for helping with the volunteers recruitment and the Drs. Claire Geurten and Xavier Schmitz for

helping with data acquisition. This study was supported by a Grant from the Belgian fund for scientific research (F.N.R.S.).

Author contribution statement SG: concept and design of the study, data acquisition, presentation and discussion of the results, writing. MB: data extraction, statistical analysis, discussion of the results, writing. ND: statistical analysis, writing. CS: data acquisition, data extraction, discussion of the results, writing. MD: data acquisition, data extraction, discussion of the results, writing. CG: data acquisition, writing. FD: data acquisition, writing, corrections as native speaker. ES: concept and design of the study, methodological advice, writing. GG: concept and design of the study, methodological advice, writing. OB: concept and design of the study, methodological advice, writing. OB: concept and design of the study, methodological advice, writing. JLC: concept and design of the study, methodological advice, writing. JP: concept and design of the study, methodological advice, writing.

Compliance with ethical standards

Conflict of interest All authors disclose any financial and personal relationships with other people or organization that could inappropriately influence their work.

Statement of human and animal rights The medical ethical committee of the University hospital of Liège (CHU Liège, Belgium) approved the protocol.

Informed consent All participants signed inform consent.

References

1. Moe-Nilssen R, Helbostad JL (2004) Estimation of gait cycle characteristics by trunk accelerometry. J Biomech 37:121–126. doi:10.1016/S0021-9290(03)00233-1
2. Auvinet B, Berrut G, Touzard C et al (2002) Reference data for normal subjects obtained with an accelerometric device. Gait Posture 16:124–134. doi:10.1016/S0966-6362(01)00203-X
3. Bautmans I, Jansen B, Van Keymolen B et al (2011) Reliability and clinical correlates of 3D-accelerometry based gait analysis outcomes according to age and fall-risk. Gait Posture 33:366–372. doi:10.1016/j.gaitpost.2010.12.003
4. Hartmann A, Luzi S, Murer K et al (2009) Concurrent validity of a trunk tri-axial accelerometer system for gait analysis in older adults. Gait Posture 29:444–448. doi:10.1016/j.gaitpost.2008.11.003
5. Gillain S, Warzee E, Lekeu F et al (2009) The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects and a comparison with other clinical tests used in single and dual-task conditions. Ann Phys Rehab Med 52:453–474. doi:10.1016/j.rehab.2008.10.004
6. Gillain S, Dramé M, Lekeu F et al (2015) Gait speed or gait variability, which one to use as a marker of risk to develop Alzheimer disease? A pilot study. Aging Clin Exp Res 28:249–255. doi:10.1007/s40520-015-0392-6
7. Hollman JH, McDade EM, Petersen RC (2011) Normative spatiotemporal gait parameters in older adults. Gait Posture 34 (1):111–118. doi:10.1016/j.gaitpost.2011.03.024
8. Senden R, Grimm B, Heyligers IC et al (2009) Acceleration-based gait test for healthy subjects: Reliability and reference data. Gait Posture 30:192–196. doi:10.1016/j.gaitpost.2009.04.008
9. Senden R, Meijer K, Heyligers IC et al (2012) Importance of correcting for individual differences in the clinical diagnosis of gait disorders. Physiotherapy 98:320–324. doi:10.1016/j.physio.2011.06.002
10. Gorton Iii GE, Hebert DA, Gannotti ME (2009) Assessment of the kinematic variability among 12 motion analysis laboratories. Gait Posture 29:398–402. doi:10.1016/j.gaitpost.2008.10.060
11. O'Sullivan K, O'Sullivan L, Campbell A et al (2012) Towards monitoring lumbo-pelvic posture in real-life situations: concurrent validity of a novel posture monitor and a traditional laboratory-based motion analysis system. Man Ther 17:77–83. doi:10.1016/j.math.2011.09.006
12. Schwartz C, Denoël V, Forthomme B et al (2015) Merging multi-camera data to reduce motion analysis instrumental errors using Kalman filters. Comput Methods Biomech Biomed Eng 18:952–960. doi:10.1080/10255842.2013.864640
13. Deshpande N, Metter EJ, Bandinelli S et al (2009) Gait speed under varied challenges and cognitive decline in older persons: a prospective study. Age Ageing 38:509–514. doi:10.1093/ageing/afp093
14. Fitzpatrick AL, Buchanan CK, Nahin RL et al (2007) Associations of gait speed and other measures of physical function with cognition in a healthy cohort of elderly persons. J Gerontol Ser A Biol Sci Med Sci 62:1244–1251
15. Artaud F, Singh-Manoux A, Dugravot A et al (2015) Decline in fast gait speed as a predictor of disability in older adults. J Am Geriatr Soc 63:1129–1136. doi:10.1111/jgs.13442
16. Scherder E, Eggermont L, Swaab D et al (2007) Gait in ageing and associated dementias; its relationship with cognition. Neurosci Biobehav Rev 31:485–497. doi:10.1016/j.neubiorev.2006.11.007
17. Verghese J, Holtzer R, Lipton RB et al (2012) Mobility stress test approach to predicting frailty, disability, and mortality in high-functioning older adults. J Am Geriatr Soc 60:1901–1905. doi:10.1111/j.1532-5415.2012.04145.x
18. Montero-Odasso M, Muir SW, Speechley M (2012) Dual-task complexity affects gait in people with mild cognitive impairment: the interplay between gait variability, dual tasking, and risk of falls. Arch Phys Med Rehabil 93:293–299. doi:10.1016/j.apmr.2011.08.026
19. Yogev-Seligmann G, Rotem-Galili Y, Mirelman A et al (2010) How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. Phys Ther 90:177–186. doi:10.2522/ptj.20090043
20. Boutaayamou M, Schwartz C, Stamatakis J et al (2015) Development and validation of an accelerometer-based method for quantifying gait events. Med Eng Phys 37:226–232. doi:10.1016/j.medengphy.2015.01.001
21. Ko S-u, Tolea MI, Hausdorff JM et al (2011) Sex-specific differences in gait patterns of healthy older adults: results from the baltimore longitudinal study of aging. J Biomech 44:1974–1979. doi:10.1016/j.jbiomech.2011.05.005
22. Patterson KK, Nadkarni NK, Black SE et al (2012) Gait symmetry and velocity differ in their relationship to age. Gait Posture 35:590–594. doi:10.1016/j.gaitpost.2011.11.030
23. Dadashi F, Mariani B, Rochat S et al (2014) Gait and foot clearance parameters obtained using shoe-worn inertial sensors in a large-population sample of older adults. Sens Basel Sens 14:443–457. doi:10.3390/s140100443

24. Laufer Y (2005) Effect of age on characteristics of forward and backward gait at preferred and accelerated walking speed. J Gerontol Ser A Biol Sci Med Sci 60:627–632. doi:10.1093/gerona/60.5.627
25. van Iersel MB, Ribbers H, Munneke M et al (2007) The effect of cognitive dual tasks on balance during walking in physically fit elderly people. Arch Phys Med Rehabil 88:187–191. doi:10.1016/j.apmr.2006.10.031
26. Hausdorff JM, Schweiger A, Herman T et al (2008) Dual-task decrements in gait: contributing factors among healthy older adults. J Gerontol Ser A Biol Sci Med Sci 63:1335–1343

Aging Clin Exp Res
DOI 10.1007/s40520-015-0392-6

ORIGINAL ARTICLE

Gait speed or gait variability, which one to use as a marker of risk to develop Alzheimer disease? A pilot study

S. Gillain[1] · M. Dramé[2] · F. Lekeu[3] · V. Wojtasik[3] · C. Ricour[4] · J.-L. Croisier[5] · E. Salmon[6] · J. Petermans[4]

Received: 4 February 2015 / Accepted: 22 May 2015

Abstract

Background Previous literature demonstrates the interest of gait analysis to predict cognitive decline in old people.
Aims This pilot study aims to determine if gait speed or gait variability is a marker able to early identify, among mild cognitive impairment (MCI) subjects, those at risk to develop Alzheimer's disease (AD) in the future.
Methods 13 MCI subjects were included in 2007. Their gait parameters (walking speed, stride length and gait frequency, regularity and symmetry) were measured in 2007 and 2008 in simple task (ST) and in dual task (DT) using a triaxial accelerometer (Locometrix®). Among the 13 MCI subjects included in 2007, 10 were assessed in 2008. So, 23 (13 in 2007 + 10 in 2008) gait tests were collected. In 2011, MCI people were considered as "MCI+" when they developed AD (between baseline and 2011) and as "MCI−" if they did not. Among the 23 gait tests, 15 were from MCI+ (9 gait tests in 2007 and 6 in 2008) and 8 from MCI− (4 gait tests in 2007 and 4 gait tests in 2008). Mann–Whitney non-parametric *U* test was used to compare gait parameters of MCI+ and MCI−.
Results Gait speed, symmetry and regularity were lower in MCI+ than in MCI−.
Discussion Despite the small sample size, the results presented in this original pilot study are in line as the infrequent previous literature related to this topic. The authors discuss lacks and strengths of this work.
Conclusions These results suggest that both gait speed and gait variability could be markers to early identify MCI at risk to develop AD.

Keywords Variability · Regularity · Gait speed · MCI · Alzheimer disease

Abbreviations

AD	Alzheimer disease
CIRS	Cumulative Illness Rating Scale
DT	Dual task
GDS-15	Geriatric Depression Scale—15 items
MNA	Mini Nutritional Assessment
MCI	Mild Cognitive Impairment
MCI+	MCI who will develop AD
MCI−	MCI who will not develop AD
MCR	Motor cognitive risk
MMSE	Mini Mental State Evaluation
ST	Simple task

Introduction

Since the last 20 years, the number of studies including instrumental gait analysis are growing, especially those concerning the relationships between gait performances

✉ S. Gillain
s.gillain@gmail.com

[1] Geriatric Department, CHU Site Notre Dame des Bruyères, Rue de Gaillarmont, 600, 4032 Chenee, Belgium
[2] Faculty of Medicine, University of Reims Champagne-Ardenne, 3797 Reims, France
[3] Geriatric Day Hospital and Memory Clinic, University Hospital of Liège, Liège, Belgium
[4] Geriatric Department, University Hospital of Liège, Liège, Belgium
[5] Motricity Sciences Department, University Hospital of Liège, Liège, Belgium
[6] Memory Clinic, University Hospital of Liège, Liège, Belgium

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | LE | TYPESET
MS Code : ACER-D-15-00050 | CP | DISK

and cognitive functions [1, 2], the relationships between the gait performance and the brain modifications related to neurodegenerative process [3, 4], and the relationship between gait performance and vascular burden [5, 6].

In this context, gait speed [7, 8] and variability of the gait seem to be potential parameters predicting cognitive decline and dementia in seniors [9, 10]. Performances in these two parameters could be influenced by several confounders as the age [11, 12], the gait speed [12], history of falls [13], cognitive functioning [10, 14], frail status [15] and the walking conditions [16]. A recent study tries to identify between these two parameters which one is the most associated with specific cognitive functions among MCI people [17]. But actually it remains unclear which parameters (gait speed, gait variability or both) are most useful to predict cognitive decline.

The goals of this prospective and exploratory study were first, to analyze gait performance of a group of MCI presenting at least a possible confounder; second, after a 3-year follow-up, to identify into this group, the MCI patients who will develop AD and those who will not; third, to compare gait performance obtained at baseline. The authors hypothesize that the gait speed and the parameters showing the variability of the gait could help the clinician to discern earlier MCI at risk to develop AD.

Population, materials and method

Population

The MCI patients were recruited among those attending Liège University Hospital's Memory Centre. Memory disorders were diagnosed by standard medical imaging and neuropsychological evaluation methods. According to Petersen criteria [18], the diagnostic of MCI was established when patients present a confirmed, isolated cognitive disorder without important impact on their activities of daily living and undergo neurological, neuropsychological and neuro-imaging diagnostic evaluations with a clinical dementia rating score (CDR) below 0.5 [19]. Other exclusion criteria included mental retardation, less than four regular years of education, cranial trauma, epilepsy, cancer, depression, drugs abuse or any other acute organic disease. At inclusion, none of the patients was taking any medication likely to influence their cognitive performance. Their score in the Mini Mental State Examination (MMSE) [20] had to be 24/30 or more. In 2011, after a 4-year follow-up, MCI considered as "MCI+" were diagnosed as having probable AD according to the criteria defined by the National Institute of Neurological and Communicative Disorders and Stroke/Alzheimer Disease and Related Disorders Association (NINCDS-ADRDA) [21]. All subjects attending this study were assessed by a complete neurological and neuropsychological evaluation, and with a FDG-PET scan to provide the diagnosis.

The following medical conditions were exclusion criteria: vascular stroke with motor or sensory disorder; Parkinson's disease; non-compensated diabetes; non-compensated arterial hyper- or hypotension; any cardiac or respiratory disease which could cause gait-limiting weakness or dyspnea; a hip or knee prosthesis; arthritis or another invalidating bone/joint disease.

The use of benzodiazepine, antidepressant or small doses of neuroleptics (without motor repercussions) was accepted. Patients needing glasses and/or hearing aid were eligible but the subjects had to be completely satisfied with the performance of these sensory aids. A medical evaluation including an interview (to establish the subject's full personal medical history), and a comprehensive clinical and functional examination was performed for all patients to check for the absence of exclusion criteria and to ensure that the gait test results and the neuropsychological assessment would not be influenced by any organic, affective or functional factors. Then, medical and functional assessment included sex, age, body mass index (BMI in kg/m^2), Mini Nutritional Assessment (MNA) [22], co-morbidities according to the Cumulative Illness Rating Scale (CIRS) [23], pain evaluation using pain horizontal analogue visual scale [24, 25], mood evaluation using the 15-items Geriatric Depression Scale (GDS-15) [26] and an evaluation of autonomy for basic and instrumental daily living activities using Katz scale (ADL) [27] and Lawton scale (IADL) [28], respectively. The scores considered for the GDS and the Lawton scale were the sum of the score obtained divided by the number of items applied (an item were not applied if the activity never has been done by the subject; e.g. men never doing housework).

After this assessment, 13 MCI persons are eligible and accept the follow-up. They were informed about the experimental procedure and provided written inform consent. The study was approved by the local ethics committee of the University Medical Centrum of Liège (Belgium).

Material

Gait analysis system

The gait analysis system used (Locometrix®) is an accelerometric method comprising an acceleration sensor, a recording device and a computer program for processing the acceleration signal. The sensor is composed of two accelerometers placed perpendicularly to each other in a plastic box as previously explained [29, 30]. The sensor's box is incorporated in an elastic abdominal belt, behind the back over the L3–L4 intervertebral lumbar space (the third

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | LE | TYPESET
MS Code : ACER-D-15-00050 | CP | DISK

lumbar vertebra level) using an elastic, abdominal belt. The first accelerometer is aligned to the mediolateral axis of the body; the second is aligned to the cranio-caudal axis. Acquisition frequency of the signal was of 50 Hz. The system can record continuously for 10 min. The recorded signals are transferred to a laptop computer using a transfer program operated under windows 98, formatted in files and analyzed by software developed in the MATLAB 5 environment. The data are transferred to a computer for statistical spreadsheet analysis.

Gait analysis

As explained previously [31, 32], during the test, the subject walks up and back along a straight 40 m corridor, free of obstacles or visual/auditory distractions, at a freely chosen pace and cadence, and using their usual walking shoes avoiding high heels. Two timing lines are located 5 m after the starting line and 5 m before the 40 m line, respectively, allowing the time measurement on 30 m walk. First, subjects were asked to walk in simple task at preferred walking speed. The same day, subjects were asked to walk in dual task condition, again at preferred walking speed and without prioritization instructions. According to Professor O. Beauchet, we choose a countdown from 50 as cognitive task during dual task because this is the additional cognitive task that perturbs most of the gait parameters in a dual-task paradigm [33].

Data processing

As explain before [29, 30], two periods of steady state walking of 20.48 s was selected from each subject. The first one was concerning simple task (ST) conditions and the other one concerned dual task (DT) conditions. Each period (of 20.48 s) contained about 1024 acceleration measurements and provided an optimal calculation time. This period correspond to 19–21 gait cycles. Using the walking time and according to the software (using fast Fourier transformation), the following gait variables are available:

- Gait speed, measured using a timing line and expressed in meters/second.
- Stride frequency or number of cycles per second (Hz), calculated from the cranio-caudal acceleration following application of a Fourier transform.
- Stride length, deduced from the equation [speed = frequency × stride length] and expressed in meters.
- Regularity, measured by the similarity (in terms of duration and amplitude) of the shape of cranio-caudal acceleration curves from steps and strides. This parameter is expressed in absolute value.
- Symmetry, defined as the similarity (in terms of duration and amplitude) of the shape of cranio-caudal acceleration curves when focusing on the right and left steps. This parameter is expressed in absolute value.

As explain before [29, 30], symmetry and regularity were calculated based on two different coefficients, C1 and C2. These coefficients are calculated based on the auto correlation of the vertical accelerating signal. C1 represents the correlation between the vertical accelerating signals considering one step to the following step (a step is a part of a stride and a stride includes a left step and a right step). In fact, each step is correlated to the following step (autocorrelation) and C1 shows the mean value of all these autocorrelations. C2 represents the correlation between the vertical accelerating signals considering all successive strides. The symmetry is calculated as C1/C2 × 100. The regularity is calculated as (C1 + C2) × 100.

All subjects walked first time in ST and after in DT.

From 2007 until 2011, all subjects were yearly assessed by neuropsychological testing as used in the memory clinic. Their cognitive status was classified according to the neurological and neuropsychological criteria previously detailed. In 2011, according to the neurological diagnosis, subjects were considered as "MCI+" when they developed AD between 2007 and 201,1 and as "MCI−" if they did not. According to this distribution, 15 gait tests were coming from MCI+ patients and 8 gait tests were coming from MCI− patients. Among the 23 gait tests, 15 were from MCI+ (9 gait tests in 2007 and 6 in 2008) and 8 from MCI− (4 gait tests in 2007 and 4 gait tests in 2008). In the pilot study, we considered each gait testing as an individual gait test and not as a serial test on the same person. Then, we performed statistical analysis concerning 23 walking tests. We use the Mann–Whitney *U* test, a non-parametric statistical test to do the comparison between the mean gait performance of "MCI+" and "MCI−" patients. A p value <0.05 was considered significant throughout and data normality was confirmed using the Lilliefors test.

Results

Main medical characteristics, functional and neuropsychological performances from MCI subjects at inclusion are presented in the Table 1.

As shown in Table 2, MCI+ patients have a significant statistical difference with MCI− patients concerning the gait speed (in ST and DT) and concerning the symmetry in DT. Gait speed and symmetry are higher in MCI− patients than in MCI+ patients.

All MCI people show worse gait performances in DT compared to ST.

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | LE | TYPESET
MS Code : ACER-D-15-00050 | CP | DISK

AQ6 **Table 1** Main characteristics of MCI at inclusion

Medical and functional variables	MCI+, $N = 9$ (mean ± SD)	MCI−, $N = 4$ (mean ± SD)
Age	74.44 ± 4.16	70.00 ± 2.16
Sex	4 women	2 women
MNA	19.87 ± 7.00	23.25 ± 6.86
CIRS	5.00 ± 2.60	5.50 ± 3.42
Visual analog scale pain	0.89 ± 2.67	0.50 ± 1.00
GDS	0.13 ± 0.14	0.19 ± 0.02
ADL	6 ± 0.00	6 ± 0.00
IADL	0.26 ± 0.03	0.26 ± 0.00
MMSE	26.11 ± 1.45	27.25 ± 1.70
Mattis, total score	133.11 ± 5.67	137.50 ± 5.20
Mattis, attentional score	35.78 ± 1.09	36.00 ± 1.15
Mattis, initiation score	32.89 ± 3.56	36.00 ± 1.41
Mattis, construction score	5.78 ± 0.44	6.00 ± 00
Mattis, conception score	37.22 ± 1.48	37.25 ± 1.71
Mattis, memory score	21.44 ± 3.28	22.25 ± 2.28
GrB, free recall total score	15.77 ± 8.13	18.50 ± 8.70
GrB, cued free recall	35.11 ± 8.82	38.00 ± 9.83
GrB, delay free recall	4.33 ± 2.45	5.50 ± 4.36
GrB, delay cued free recall	12.00 ± 2.78	13.00 ± 3.46

AQ7 **Table 2** Gait performances in simple task (ST) and dual task (DT) of MCI+ patients and MCI− patients and respective p value obtained using the U test of Mann–Whitney

Gait's parameters	Gait of MCI+ (N = 15), mean ± SD	Gait of MCI− ($N = 8$), mean ± SD	p value
Speed in ST (m/s)	1.15 ± 0.13	1.29 ± 0.10	**0.017**
Regularity in ST	286.20 ± 37.45	298.0 ± 22.46	0.196
Symmetry in ST	219.47 ± 36.92	269.13 ± 69.86	0.061
Speed in DT (m/s)	1.01 ± 0.18	1.18 ± 0.12	**0.036**
Regularity in DT	220.67 ± 254.88	254.88 ± 32.86	0.129
Symmetry in DT	220.60 ± 66.78	201.13 ± 28.03	**0.039**

Moreover, the regularity is lower in MCI+ than in MCI−, but the difference is not statistically significant.

Discussion

The results of this pilot study highlight the interest of accelerometric measurements of gait to help the early detection of MCI at risk of developing AD, especially according to the gait speed and its symmetry. According to the recent literature, we would discuss the interest to study the gait speed, the use of a DT and the interest to consider the variability of gait.

Concerning the gait speed, in this study, MCI patients who develop AD have a lower gait speed (in ST and in DT) than those who do not. Our results are convenient with those obtained in different cohort included in prospective study with dementia as clinical outcome [2, 7, 8]. Actually, these previous studies highlight the interest of gait speed essentially concerning the risk of vascular dementia. According to JM Hausdorff [34], imaging studies and pathology studies highlight the presence of vascular burden and AD lesions even in older people without clinical signs of dementia. According to Verghese [35], studying gait speed of an old person already presenting a mild cognitive decline could help to detect people at risk to develop dementia. The main idea of the concept of MCR is that the motor dysfunction and the cognitive decline are both sign of the same pathological process including diffuse lesions in the brain leading to dementia. This MCR concept has already shown its association with cognitive decline in a wider cohort [36]. Unfortunately and according to our knowledge, no published data shows a strong relation between slow gait speed and specific risk to specifically

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | LE | TYPESET
MS Code : ACER-D-15-00050 | CP | DISK

develop AD. Then, a strict comparison of our results to previous literature is still limited.

Concerning the use of a DT, in this study, all MCI people show worse gait performances in DT comparing in ST. These results are similar to those found by Montero-Odasso and Muir using a GaitRite system, and showing the importance of executive function and working memory considering gait performances in DT [37, 38].Concerning the variability of the gait, and as explained previously, the parameters available with the Locometrix® are the regularity showing the shape of cranio-caudal acceleration curves from steps and strides, and the symmetry showing the similarity (in terms of duration and amplitude) of the shape of cranio-caudal acceleration curves when comparing right and left steps. In the actual literature [11, 39], the terms used to translate the variability (or the less-regularity) of the gait are more often the "variability of the stride length" or the "variability of the stride time" or the "variability of the step width" as expressed in terms of the coefficients of variation for each term [CV or CoV calculated as (SD/mean) × 100]. According to Moe-Nilssen [40], step time variability seems to be correlated with vertical (cranio-caudal) interstep trunk variability. Considering that Locometrix, regularity is calculated on cranio-caudal accelerations curves, and considering findings of Moe-Nilssen, the authors allow themselves that the regularity and the symmetry obtained by the Locometrix® could represent a translation of the "step time variability".

In this study, MCI+ group has a lower regularity than MCI− group. And overall, MCI+ group shows a decreasing regularity when walking in DT. Unfortunately, and even if this decreasing regularity in DT seems to be important, the difference between the two groups remains statistically non-significant, probably because of the sample size.

The second parameter showing the variability of the gait using this accelerometer is symmetry. In our study, MCI+ group presents in DT a symmetry significantly lower than MCI− group. This observation can be explained by the way used by the software to obtain the symmetry. Indeed, symmetry is calculated as C1/C2 × 100. So the symmetry can increase in case of an increase of C1 or in case of a decreasing C2. This second possibility is probably the best explanation of this decrease of symmetry in DT.

Moreover, the fact that this decrease in C2 is more "numerically important" in terms of symmetry than in terms of regularity, is probably linked to the mean to obtain the regularity [(C1 + C2) × 100], decreasing the relative importance of a decrease of C2.

These results considering the regularity and the symmetry are in the same line that other studies showing an increasing variability of the gait in MCI people at risk to develop AD [2, 17].

This study has a number of limitations and our results have to be considered with caution. First, the size of the sample is reduced because of the size of the only memory clinic attending, the number of exclusion criteria and the long time of the follow-up. The results presented have to be considered with caution. Second, a comparison would be interesting between the four sub-types of MCI (anamnestic single domain MCI, anamnestic multiple domain MCI, executive single domain MCI and executive multiple domain MCI) but unfortunately this cohort included mainly anamnestic MCI whose usually does not present a high level of gait modification. However, considering the fact that FDG-PET scanner realized at inclusion confirmed the neurodegenerative process occurring in the brain and the high level of AD development in this cohort (9 MCI/13 in 3 years), we could consider these MCI particularly "at risk" to develop AD and probably presenting widespread brain lesions. This "at risk" status could probably explain the early gait modifications.

Third, we do not know the time of conversion from MCI to AD, because we do not consider when they develop AD. Finally, only MCI developing AD was considered. MCI developing other dementia, for example, vascular dementia or frontotemporal dementia were excluded.

The strengths of this pilot study include a population strictly selected. Exclusion criteria included a lot of potential confounders as orthopedic prosthesis, previous falls, depression, sedative medication, previous neurologic pathology, clinical neurologic disorders or abnormalities. All subjects attending this study were assessed by a complete neurological, neuropsychological and with a brain FDG-PET scan to confirm the MCI neurodegenerative syndrome. MCI patients included by this way were free of confounders usually met in older people and they were more prompt to develop AD.

Moreover, the length of the straight long corridor used (40 m) allows guaranteeing strong conditions to reliably assess the gait parameters. Indeed, we exclude more than the first 2.5 m of walking to be sure to achieve the steady state walking as recommended by Lindeman [41] and with the guidelines for clinical applications of spatio-temporal gait analysis in older adults from the European Gait Rite Network [42]. Regarding the assessment of gait variability, the same author recommends walking at least 20 gait cycles. In our study, the mean cadence was 0.89 ± 0.05 for the MCI who will develop AD and the cadence of those who will not develop AD was 0.95 ± 0.04. So, by considering and analyzing period of 20, 48 s., all subjects walked at least 20 gait cycles as recommended [41].

Furthermore these results are really interesting regarding their statistical significance using a non-parametric statistical test.

Further research are needed to confirm these results in bigger sample, including not only amnestic MCI but also

Journal : Large 40520 Dispatch : 26-5-2015 Pages : 7
Article No. : 392 LE TYPESET
MS Code : ACER-D-15-00050 CP DISK

executive MCI, and not only considering AD as the only outcome but rather all the cognitive evolutions.

Conclusion

In this prospective and exploratory study, MCI who will develop AD have lower gait speed, lower symmetry and lower regularity in DT than those who will not develop AD. According with previous literature, and even if this results have to be considered with caution, the gait speed and the parameters showing the variability of gait seem to be important, considering the risk of developing dementia among MCI people.

Acknowledgments The authors wish to acknowledge the Professor Olivier Beauchet, from the CHU of Angers in France, for his lecture and his precious advices in the preparation of this paper.

Conflict of interest None of the authors have a financial or personal relationship with people or organizations that could inappropriately influence this work.

References

1. Scherder EJ, Eggermont LH, Swaab DF et al (2007) Gait in aging and associated dementias; its relationship with cognition. Neurosci Biobehav Rev 31:485–497
2. Verghese J, Wang C, Lipton RB et al (2007) Quantitative gait dysfunction and risk cognitive decline and dementia. J Neurol Neurosurg Psychiatry 78:929–935
3. Zimmerman M, Lipton R, Pan J et al (2009) MRI− and MRS−derived hippocampal correlated on quantitative locomotor function in older adults. Brain Res 1291:73–81
4. Annweiler C, Beauchet O, Bartha R et al (2013) Slow gait in MCI is associated with ventricular enlargement: results from the Gait and Brain Study. J Neural Transmit 120:1083–1092
5. Annweiler C, Montero-Odasso M (2012) Vascular burden as a substrate for higher-level gait disorders in older adults. A review of brain mapping literature. Panminerva Med 54:189–204
6. Bruijn SM, Van Impe A, Duysens J et al (2014) White matter microstructural organization and gait stability in older adults. Frontiers Aging Neurosci 6:1–11
7. Waite LM, Grayson DA, Piguet O et al (2005) Gait slowing as a predictor of incident dementia: 6-year longitudinal data from the Sydney Older Persons Study. J Neurol Sci 229–230:89–93
8. Mielke MM, Roberts RO, Savica R et al (2013) Assessing the temporal relationship between cognition and gait: slow gait predicts cognitive decline in the Mayo Clinic Study of Aging. J Gerontol 68:929–937
9. Webster KE, Merory JR, Wittwer JE (2006) Gait variability in community dwelling adults with Alzheimer disease. Alzheimer Dis Assoc Disord 20:37–40
10. Ijmker T, Lamoth CJ (2012) Gait and cognition: the relationship between gait stability and variability with executive function in people with and without dementia. Gait Posture 35:126–130
11. Beauchet O, Allali G, Anneweiler C (2009) Gait variability among healthy adults: low and high stride to stride variability are both a reflection of gait stability. Gerontology 200955:702–706
12. Helbostad JL, Moe-Nilssen R (2003) The effects of gait speed on lateral balance control during walking in healthy elderly. Gait Posture 18:27–36
13. Ayoubi F, Launay C, Anneweiler C et al (2013) Fear of falling, falls and gait variability in older community-dwelling individuals: is there an association? JAGS 61:1236–1238
14. Beauchet O, Anneweiler C, Montero-Odasso M (2012) Gait control: a specific subdomain of executive function? J Neuro Eng Rehabil 9:12
15. Moe-Nilssen R, Helbostad JL (2005) Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. Gait Posture 21:164–170
16. Yogev-Seligman G, Rotem-Galili Y, Mirelman A et al (2010) How does explicit prioritization alter walking during dual-task performance? Effects of age and sex on gait speed and variability. Phys Ther 90:177–186
17. Montero-Odasso M, Casas A, Hansen KT et al (2009) Quantitative gait analysis under dual-task in older people with mild cognitive impairment: a reliability study. J Neuro Eng Rehabil 6:35
18. Petersen RC, Doody R, Kurz A et al (2001) Current concepts in mild cognitive impairment. Arch Neurol 58:1985–1992
19. Morris JC (1993) The clinical dementia rating (CDR): current version and scoring rules. Neurology 43:2412–2414
20. Folstein MF, Folstein SE, McHugh PR (1975) Mini-mental state: a practical method for grading the state of patients for the clinician. J Psychiatry Res 12:189–198
21. Mc Kahn G, Drachman D, Folstein M et al (1984) Clinical diagnosis of Alzheimer's disease: Report of the NINCDS-ADRDA Work Group under the auspices of Department of Health and Human Services Task Force on Alzheimer's disease. Neurology 34:939–944
22. Rubenstein LZ, Harker JO, Salva A et al (2001) Screening for undernutrition in geriatric practice: developing the short-form mini nutritional assessment (MNA-SF). J. Geront 56:366–377
23. Linn BS, Linn MW, Gurel L (1968) Cumulative illness rating scale. J Am Geriatr Soc 16:622–626
24. Price DD, McGrath PA, Raffi A et al (1983) The validation of visual analogue scales as ratio scale measures of chronic and experimental pain. Pain 17:45–56
25. Wewers ME, Lowe NK (1990) A critical review of visual analogue scales in the measurement of clinical phenomena. Res Nurs Health 13:227–236
26. Sheikh JA, Yeasavage JA (1986) Geriatric depression scale (GDS): recent findings and development of a shorter version. In: Tl Brink (ed) Clinical gerontology: a guide to assessment and intervention. Howarth Press, New York, pp 165–172
27. Katz S, Ford AB, Moskovitz RW et al (1963) Studies of illness of in the aged: the index of MAL: a standardized measure of biological and psychosocial function. JAMA 185:914–919
28. Lawton MP, Brody EM (1999) Assessment of older people: self-maintaining and instrumental activities of daily living. Gerontologist 9:179–186
29. Auvinet B, Chaleil D, Barrey E (1999) Accelerometric gait analysis for use in hospital outpatients. Rev Rhum 66:389–397
30. Auvinet B, Berrut G, Touzard C et al (2002) Reference data for normal subjects obtained with an accelerometric device. Gait Posture 16:124–134
31. Gillain S, Warzée E, Lekeu F et al (2009) The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects and a comparison with other clinical tests used in single and dual-task conditions. Ann Phys Rehabil Med 52:453–474
32. Maquet D, Lekeu F, Warzee E et al (2010) Gait analysis in elderly adult patients with mild cognitive impairment and patients

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | LE | TYPESET
MS Code : ACER-D-15-00050 | CP | DISK

with mild Alzheimer's disease: simple versus dual task: a preliminary report. Clin Physiol Funct Imaging 30:51–56
33. Beauchet O, Dubost V, Aminian K et al (2005) Dual-task-related gait changes in the elderly: does the type of cognitive task matter. J Mot Behav 37:459–464
34. Hausdorff JM, Buchman AS (2013) What links gait speed and MCI with dementia? A fresh look at the association between motor and cognitive function. J Gerontol A Biol Sci Med Sci 68:409–411
35. Verghese J, Wang C, Lipton RB et al (2013) Motoric cognitive risk syndrome and risk of dementia. J Gerontol A Biol Sci Med Sci 68:412–418
36. Verghese J, Anneweiler C, Ayers E (2014) Motoric cognitive risk syndrome. Neurology 83:718–726
37. Muir SW, Speechley M, Wells J et al (2012) Gait assessment in mild cognitive impairment and Alzheimer's disease: the effect of dual-task challenges across the cognitive spectrum. Gait Posture 35:96–100
38. Montero-Odasso M, Muir SW, Speechley M (2012) Dual-Task complexity affects gait in people with mild cognitive impairment: the interplay between gait variability, dual tasking, and risk of falls. Arch Phys Med Rehabil 93:293–299
39. Beauchet O, Allali G, Anneweiler C et al (2009) Gait variability among healthy adults: low and high stride to stride variability are both a reflection of gait stability. Gerontology 55:702–706
40. Moe-Nilssen R, Aaslund MK, Hodt-Billington C et al (2010) Gait variability measures may represent different constructs. Gait Posture 32:98–101
41. Lindemann U, Najafi B, Zijlstra W et al (2008) Distance to achieve steady state walking speed in frail elderly persons. Gait Posture 27:91–96
42. Kressig R, Beauchet O (2006) Guidelines for clinical applications of spatio-temporal gait analysis in older adults. Aging clinical and experimental research 18:174–176

Journal : Large 40520 | Dispatch : 26-5-2015 | Pages : 7
Article No. : 392 | ☐ LE | ☐ TYPESET
MS Code : ACER-D-15-00050 | ☑ CP | ☑ DISK

LES CHUTES DE LA PERSONNE ÂGÉE

S. Gillain (1), L. Elbouz (2, 3), C. Beaudart (4, 5), O. Bruyère (6, 7), J-Y. Reginster (8, 9), J. Petermans (10)

Résumé : La chute de la personne âgée est un évènement fréquent, aux conséquences potentiellement graves, parfois révélatrice de pathologies associées. Il s'agit d'un réel problème de santé publique mobilisant de nombreuses équipes attelées à la recherche de solutions de prévention, de soins adaptés, et de moyens de réduction des coûts secondaires. Ce texte se propose de revoir les étiologies les plus fréquemment rencontrées en gériatrie et présente le bilan et la prise en charge multidisciplinaires réalisés à l'hôpital de jour gériatrique du CHU de Liège chez les personnes âgées chuteuses ou à risque de chute.
Mots-clés : *Personne âgée - Chute - Etiologies*

Falls in the elderly

Summary : Falls are frequent in the elderly; they can have severe consequences, and sometimes reveal some underlying pathology. They represent a real public health problem which prompts numerous teams to search for adequate preventative measures, adapted therapeutic options, and means to reduce the various resulting costs. This paper describes the multidisciplinary hospital day service organized, by the CHU of Liege, for older patients who fall, or are at risk of falling.
Keywords : *Fall - Elderly - Etiology of falls*

Introduction et épidémiologie

Dans de nombreux pays industrialisés, le vieillissement de la population est une réalité liée, d'une part, à l'allongement de l'espérance de vie et, d'autre part, à une baisse du nombre de naissances. En Belgique, on observe un accroissement de la population des personnes âgées de plus de 60 ans. En 2030, on estime que la population des 60 ans et plus sera de plus de 3 millions, soit environ 30% de la population générale (1).

Chez la personne âgée, la chute, qui correspond à l'action de tomber au sol indépendamment de sa volonté, représente un problème majeur de santé publique. Lors de la dernière enquête de santé par interview réalisée en Belgique, 24% des personnes âgées de 65 ans et plus déclaraient avoir été victimes d'au moins une chute accidentelle au cours des 12 derniers mois. Sans surprise, le pourcentage de personnes âgées de 65 ans et plus qui déclarent en avoir été victimes augmente avec l'âge : environ 15% entre 65 et 74 ans, pour atteindre 37% chez les personnes âgées de 85 ans et plus. Le pourcentage de personnes âgées de 65 ans et plus qui déclarent avoir été victimes d'au moins une chute accidentelle est deux fois plus élevé chez les femmes (30%) que chez les hommes (15%) (2).

Les conséquences d'une chute sont extrêmement variables, mais on rapporte généralement qu'entre 5 et 20% des chutes entraînent une blessure aux implications cliniques parfois importantes. Les chutes sont responsables de plus de 50% des hospitalisations pour traumatisme chez les personnes âgées de plus de 65 ans (3). Parmi les répercussions importantes des chutes, les fractures occupent une place considérable. La fracture du col du fémur, par exemple, a peut-être les conséquences cliniques et socio-économiques les plus manifestes. En effet, dans l'année suivant la survenue d'une fracture de hanche, 25% des individus décéderont et près de 80% connaîtront un déclin permanent de leur mobilité (4). Certaines chutes auront aussi pour conséquence d'entraîner une diminution de la qualité de vie de l'individu, une peur de tomber à nouveau et une réduction de l'autonomie nécessitant parfois l'admission en maison de repos. De plus, les données européennes issues d'Eurostat (http://epp.eurostat.ec.europa.eu) rapportent que le nombre de décès causés par des chutes accidentelles dans l'Union Européenne est de plus de 50.000, ce qui représente près de 20% des décès dus aux causes externes. Le taux standardisé (ajusté en fonction d'une distribution standard par âge) de mortalité par chute est d'environ 8 pour 100.000. Enfin, les conséquences économiques des chutes sont très difficiles à évaluer, mais certaines données suggèrent que près de 20% des dépenses en soins de santé engendrées par des traumatismes sont directement liés aux chutes (5).

(1) Gériatre, (2) Kinésithérapeute, (10) Chef de Service, Service de Gériatrie, Site NDB, CHU de Liège.
(3) Département des Sciences de la Motricité, Université de Liège.
(4) Assistante Chargée de Recherches Clinique, (6) Chargé de Cours, (8) Professeur Ordinaire, Ecole de Santé Publique, Département des Sciences de la Santé Publique, Epidémiologie et Economie de la Santé, Université de Liège.
(5) Chargée de Recherches cliniques, (7) Chargé de Cours et Chargé de l'encadrement des projets de Recherches, (9) Professeur Ordinaire, Directeur des Recherches, Unité de Soutien Méthodologique en Epidémiologie et en Biostatistiques, Université de Liège.

Etiologies

La marche est un acte moteur volontaire, non automatique, dirigé vers un but et nécessitant l'intégrité de nombreux systèmes organiques. Le vieillissement physiologique altère ces différents systèmes, mais l'avancée en âge n'explique pas, à elle seule, les chutes. La chute non accidentelle de la personne âgée est la conséquence de l'évolution des comorbidités, d'une pathologie (ou d'une fragilité) atteignant un ou plusieurs des systèmes impliqués et des effets des médications ou de toxiques. Ainsi, plusieurs causes de chute peuvent coexister chez une personne âgée et il sera de l'art du clinicien de percevoir quelle est la (ou quelles sont les) cause(s) principale(s) impliquée(s) afin de tenter d'en prévenir la récidive. Il est important de distinguer la chute accidentelle de la personne âgée des chutes à répétition. Dans le cas de la chute accidentelle, l'anamnèse permettra d'évaluer les circonstances et causes de l'accident et de prévenir sa récidive. Enfin, il est important de considérer certains aspects socio-économiques plus particuliers à la personne âgée comme le déconditionnement, l'inactivité, l'isolement social, la diminution des ressources humaines et financières fragilisant encore la personne âgée, malade ou non.

La première cause de chute est la chute elle-même (2). En effet, la chute du patient âgé peut entraîner une crainte de tomber, induire des modifications de la marche et majorer le risque de récidive de chute (6). Enfin, la chute, même accidentelle et sans complication, peut entraîner un syndrome de désadaptation psycho-motrice. Décrit par les professeurs Pfitzenmeyer et Manckoundia (7, 8), le syndrome de désadaptation psychomotrice, est composé d'une rétropulsion du tronc (en position assise et en position debout), d'une hypertonie oppositionnelle, d'une crainte de la verticalité et de la marche associée à une réduction des déplacements et d'une perte des réactions parachutes. La physiopathologie de ce syndrome repose sur la présence de lésions sous-corticales frontales associées à une perturbation de la programmation motrice de la mise debout et de la marche. Elle entraîne une crainte de tomber et majore le risque de chute. Sans prise en charge rapide et adéquate, le sujet souffrant d' un syndrome de désadaptation psycho-motrice présentera une réduction du périmètre de marche et un déclin fonctionnel secondaire.

La chute de la personne âgée peut parfois révéler une pathologie méconnue ou insuffisamment traitée. Ainsi, toute chute accompagnée d'une syncope, de signes de réponse vagale, ou d'un étourdissement orientera le diagnostic vers une pathologie cardiaque ischémique, valvulaire ou rythmique avec ses possibles complications thromboemboliques. Elle devra entraîner un bilan complémentaire comprenant un ECG, une échographie cardiaque de repos et / d'effort, un Doppler des vaisseaux du cou et un Holter de rythme ou de pression. Les chutes accompagnées de signes neurologiques déficitaires orienteront davantage vers une étiologie neuro-vasculaire ou comitiale.

Les causes médicamenteuses (somnifères, neuroleptiques, médications anticholinergiques) doivent bien entendu être relevées.

Le clinicien sera attentif aux déformations ostéo-articulaires entraînant une projection du centre de gravité du sujet hors de la base de sustentation ou provoquant une douleur qui modifie la marche par une boiterie compensatrice ou antalgique et diminue les ressources attentionnelles allouées à la marche.

L'amyotrophie et la sarcopénie (9), physiologiques avec l'âge, mais très nettement aggravées par l'inactivité, la dénutrition, les comorbidités, l'évolution des pathologies chroniques, et les médications catabolisantes, sont, bien entendu, des causes de perturbation de la marche et de diminution des réactions posturales. Enfin, les troubles visuels, la douleur chronique et aiguë, les pathologies respiratoires chroniques et l'hypoxie, la consommation d'alcool, le chaussage et l'environnement du sujet doivent également être considérés.

Les auteurs choisissent de considérer les causes neurologiques des chutes en termes de symptômes (10, 11).

La faiblesse des membres inférieurs, une sensation de dérobement ou de fatigabilité musculaire importante peuvent être secondaires à une atteinte du premier ou du second motoneurone ou à une myopathie. La spasticité signe une atteinte du premier motoneurone, indépendamment de la présence d'une fatigabilité. Une myélopathie en est la cause. L'examen clinique retrouve une augmentation du tonus de la jambe atteinte.

L'altération de la proprioception est le plus souvent secondaire à une atteinte de la corne postérieure ou des voies proprioceptives périphériques et du ganglion postérieur par une spondylite cervicale, une polyneuropathie sévère (souvent secondaire au diabète), une carence en vit B12, un tabès syphilitique ou une atteinte paranéoplasique. Classiquement,

le patient tape les talons sur le sol et laisse tomber lourdement le pied sur le sol; la base de sustentation est élargie et la marche se dégrade nettement dans l'obscurité. Un Romberg est typiquement présent.

Les signes extrapyramidaux sont retrouvés dans la maladie de Parkinson, les syndromes parkinsoniens dus aux toxiques (pesticides), aux lésions vasculaires, ou à des processus neurodégénératifs comme l'atrophie multisystémique, la paralysie supranucléaire progressive ou la dégénérescence cortico-basale entraînant un dysfonctionnement des noyaux gris centraux. Dans ce contexte, les altérations des mouvements, des réactions d'adaptation posturale, et de la marche sont nombreuses et méritent, à elles seules, un chapitre dédié dans ce numéro spécial.

L'ataxie cérébelleuse entraîne une incoordination des mouvements volontaires et réflexes, faite de dysmétrie, de dyssynergie et de dysrythmie. L'ataxie est multidirectionnelle au Romberg, non augmentée à la fermeture des yeux et sans altération des réflexes posturaux. Les causes les plus fréquentes en gériatrie sont l'alcoolisme, une atteinte vasculaire, une atteinte néoplasique ou paranéoplasique (Ac anti-Yo), une hypothyroïdie sévère, un traumatisme ou un épisode d'anoxie. La démarche est comme «saoule» avec une large base de sustentation et de petits pas irréguliers.

L'atteinte vestibulaire peut être bi- ou unilatérale. L'atteinte unilatérale est marquée par un vertige rotatoire, des vomissements, et un nystagmus dont la phase lente indique le côté lésé. La posture est instable, parfois déviée et le déséquilibre est augmenté à la fermeture des yeux. La marche est déviée dans la direction du vestibule atteint. Le patient écarte les jambes et réduit la longueur de ses pas pour gagner en stabilité.

La marche frontale accompagne les lésions corticales et sous-corticales des régions frontales et préfrontales. Elle est associée au syndrome de désadaptation psychomotrice évoqué précédemment dans ce texte.

Les myoclonies à l'orthostatisme (12) accompagneraient des troubles métaboliques aussi bien que des états neurodégénératifs avancés.

Les troubles cognitifs, attentionnels et la confusion sont associés à des perturbations de la marche et à un risque de chute accru (13-15). En effet, le développement des techniques d'analyse instrumentale de la marche et les progrès de l'IRM et de l'imagerie fonctionnelle ont permis de démontrer que les structures impliquées sont communes à la marche et aux fonctions cognitives, notamment les fonctions exécutives (16). Ce partage des structures a permis l'application du paradigme de la double tâche. Toute lésion (vasculaire, neurodégénérative, tumorale, toxique ou infectieuse) atteignant ces systèmes ou leurs connections peut entraîner des troubles cognitifs et une altération de la marche.

Ainsi, les facteurs de risque de chute les plus fréquemment rencontrés en gériatrie sont présentés dans la revue de littérature de M. Tinetti (2). Sont cités dans l'ordre de fréquence : les antécédents de chute, les troubles de l'équilibre, la sarcopénie, les troubles visuels, les médications (> 4 ou les médications psychotropes), les troubles de la marche objectivés ou les difficultés ressenties à la marche, la dépression, les vertiges et l'hypotension orthostatique, la perte d'autonomie objectivée par une échelle de Katz, l'âge > 80 ans, le sexe féminin, un BMI bas (< 20), les troubles de la continence, les troubles cognitifs, l'arthrite, le diabète et la douleur. Il est important de noter que l'âge n'occupe que la dixième place. Etonnamment aussi, les troubles cognitifs n'apparaissent «seulement» qu'en quatorzième position, probablement parce que ce facteur de risque, unanimement reconnu à l'heure actuelle, n'a pas été systématiquement relevé lors des études cliniques réalisées à l'époque.

Conséquences de la chute

Outre la crainte de tomber, le syndrome post-chute et la récidive, les conséquences des chutes sont très variées (plaies, fractures, hématomes sous-duraux). Les conséquences thymiques et sociales sont davantage fonction des ressources physiques, psychiques, humaines et économiques dont dispose la personne âgée. Enfin, les conséquences socio-économiques, comme rapportées précédemment, sont importantes.

Bilan de chute ou dépistage du risque

Toute personne âgée présentant des facteurs de risque de chute doit subir un bilan étiologique multidisciplinaire permettant la mise en place des stratégies adaptées. Conscient de l'importance de ce bilan, l'INAMI autorise le remboursement partiel de deux bilans fonctionnels complets par an. Ceux-ci sont réalisés en hôpital de jour gériatrique. Le bilan fonctionnel pluridisciplinaire comprend un bilan médical et infirmier ainsi qu'une évaluation ergothérapeutique et neuropsychologique. Au CHU de

Liège, ce bilan est complété par une évaluation kinésithérapeutique détaillée ci-après.

Le bilan médical comprend l'anamnèse et l'examen clinique.

Le bilan infirmier utilise des échelles validées permettant d'évaluer l'autonomie, de dépister la dénutrition, les troubles de la déglutition, la douleur, les troubles du sommeil, la dépression ou les troubles du comportement; il comporte encore la prise des paramètres et la recherche d'une hypotension orthostatique.

Le bilan ergothérapeutique permet une revue de l'environnement, de la gestion médicamenteuse et de l'autonomie dans les gestes quotidiens (toilette, habillage, repas …), un bilan des afférences visuelles, un bilan des capacités d'équilibre et de marche, du chaussage et de l'utilisation d'une éventuelle aide à la marche (canne ou déambulateur).

Le bilan neuropsychologique comprend le bilan cognitif et thymique ainsi que l'éducation du patient et de ses proches aux troubles cognitifs relevés et aux méthodes d'aide à l'apprentissage et à la mémorisation.

Enfin, le bilan de chute comprend une prise de sang incluant les sérologies syphilis et Borrelia, une imagerie cérébrale, et, selon l'anamnèse et l'examen clinique, un EEG, un EMG, une échographie cardiaque, un Doppler des vaisseaux du cou à la recherche d'une pathologie méconnue ou d'une comorbidité non ou insuffisamment traitée. A des fins de clarté, les auteurs proposent un schéma résumant l'orientation diagnostique (fig. 1).

Le bilan kinésithérapeutique relève l'ensemble des troubles fonctionnels secondaires aux comorbidités ostéo-articulaires, musculaires, neurologiques et/ou à la présence d'une douleur aiguë ou chronique. Les tests de marche et d'équilibre sont adaptés à l'environnement propre au cadre de vie du sujet. Une mise en situation de chute est également réalisée. Enfin, au terme du bilan, des conseils sont prodigués au patient et à ses proches.

De nombreux tests cliniques évaluent les capacités d'équilibre statique et/ou dynamique et il est impossible de les évoquer tous dans le cadre de cet article. Certains, fréquemment retrouvés dans la littérature sont devenus «incontournables» comme le Timed Up and Go (17). Très discuté concernant sa valeur prédictive du risque de chute (18), ce test est simple à réaliser et informe le clinicien sur la qualité des mouvements quotidiens nécessaires à l'autonomie du sujet. Le test de Tinetti est également utilisé en hôpital de jour; il évalue l'équilibre statique et dynamique du sujet en dix minutes et son score couplé à celui du TUG ou du Timed chair stand (> 14 secondes) permet, en cas d'antécédent de chute durant l'année écoulée, d'obtenir le remboursement partiel de 60 séances de kinésithérapie de revalidation à la marche. La mesure de la vitesse de marche sur 4 mètres (démarrage à l'arrêt) est devenue incontournable, surtout depuis le remarquable travail de Studensky qui s'est attaché à démontrer le caractère prédictif de cette vitesse de marche (19). Enfin, un examen complet de la marche inclut une inspection de la marche en double tâche. Appliqué à la marche, le paradigme de la double tâche (marcher ET, dans le même temps, réaliser une tâche cognitive de mémoire de travail ou de fluence) permet de révéler au clinicien une fragilité de la marche et un risque de chute passé inaperçu en tâche simple (20, 21). Utilisée lors de l'étude instrumentale de la marche, la marche en double tâche serait prédictive du déclin fonctionnel (22) et cognitif (23), mais son application en pratique quotidienne reste encore limitée.

Actuellement encore réservée au domaine de la recherche, l'analyse instrumentale de la marche permet une étude plus précise de la marche en simple et double tâche, dans

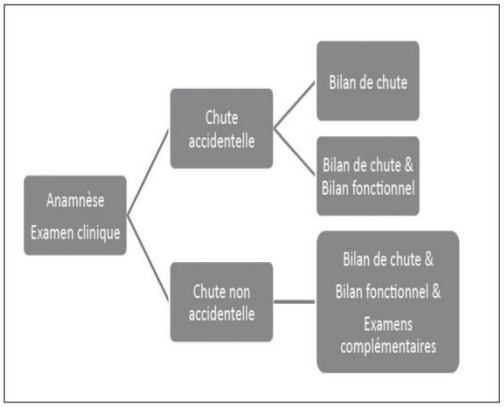


Figure 1. Chute accidentelle : si la chute accidentelle est associée à un ou plusieurs signe(s) de fragilité relevé(s) lors de l'anamnèse ou de l'examen clinique, le bilan de chute sera associé à un bilan fonctionnel complet tel que décrit dans cet article.
Chute non accidentelle : suspicion de pathologie cardiaque ou neuro-vasculaire, endocrinienne, comitiale ou neuro-dégénérative.
Bilan de chute : bilan réalisé dès la première chute, qu'elle soit accidentelle ou non. Ce bilan comprend : - une revue de la marche, du chaussage, de l'aide à la marche et de l'environnement, - une revue des afférences visuelles et si nécessaire un avis ophtalmologique, - un test de marche en double tâche à la recherche d'un trouble attentionnel méconnu, - envisager l'adaptation du domicile et le port d'une télé-biovigilance, - envisager la revalidation à la marche, - envisager le bilan osseux et biologique, prévention des fractures, révision des médications.

différentes conditions de marche et d'équilibre. Différents instruments existent. Citons les méthodes cinématiques, les plateformes de force, les tapis de marche et les méthodes accélérométriques analysant les composants spatio-temporels de la marche. Utilisées dans différents protocoles, ces méthodes ont permis la distinction de profils de marche associés à des situations pathologiques (24-26).

Prévention de la chute/des récidives

La prévention des chutes et de leur récidive découle, pour une grande part, des fragilités mises à jour lors du bilan. Les solutions proposées doivent être adaptées et discutées avec le sujet pour faciliter leur compréhension et leur intégration dans les habitudes de vie au quotidien. Enfin, le clinicien doit veiller à ne pas placer le sujet en situation d'échecs répétés, ce qui est décourageant pour le patient, les soignants, et renforce les stéréotypes négatifs véhiculés à l'égard des personnes âgées.

Les exercices aérobies comme la marche, la natation ou la danse durant 30 à 40 minutes trois fois par semaine ont des effets bénéfiques sur la force musculaire et aident à la prévention de la sarcopénie, du déclin fonctionnel et cognitif et améliorent l'équilibre (27, 28). Le renforcement musculaire ciblé des fessiers, des quadriceps, des jambes, mais aussi des lombaires et des abdominaux, améliore les réactions d'adaptation posturale permettant d'éviter les chutes secondaires à une brusque perte d'équilibre. La revalidation fonctionnelle est composée :

- d'un travail des transferts,
- d'une rééducation de l'initiation de la marche,
- d'une correction de la marche (gain de longueur de pas, restauration du ballant des bras, restauration d'un rythme régulier, posture),
- d'un renforcement musculaire des membres inférieurs,
- d'exercices de proprioception dont la difficulté augmente selon les progrès du patient,
- d'un parcours de motricité composé d'obstacles au sol, et de différents revêtements de sol,
- d'un travail des réactions posturales et, enfin,
- d'une revalidation de la marche réalisée en double tâche.

Si le patient en semble capable, le kinésithérapeute pourra encore proposer une mise en situation de chute et l'apprentissage aux méthodes pour se relever seul ou avec une chaise.

Mobilisant attention et coordination musculaire, le Tai Chi a également démontré des effets positifs y compris dans les populations plus âgées (29). Les méthodes de revalidation à la marche et à l'équilibre en double tâche ont apporté de réels effets bénéfiques de prévention des chutes. C'est le cas notamment de la revalidation rythmique de Jacques Delcroze (30).

La revalidation à la marche des patients déments est réalisable et montre des effets en termes d'amélioration de l'équilibre et de prévention du déclin fonctionnel (31) y compris en double tâche (32).

Enfin, il reste indispensable de prévenir les complications organiques des chutes. Etonnamment, et même s'il paraît évident que les interventions visant à réduire le risque de chute diminuent probablement l'incidence des fractures fémorales, aucune étude n'a démontré un lien direct entre ces deux phénomènes (33). Concernant les fractures, la vitamine D, le calcium, les médicaments du métabolisme osseux, les protecteurs de hanche devront être discutés (34). En effet, de nombreuses études épidémiologiques suggèrent qu'un faible taux sanguin de 25(OH)D (qui est un reflet fiable du statut en vitamine D) est associé à une augmentation du risque de chute et/ou de fracture chez la personne âgée. Les effets d'une supplémentation en vitamine D (35) sur le risque de chute ont été évalués dans plusieurs études randomisées contrôlées en double aveugle, mais avec des résultats contradictoires, probablement dus à une grande variabilité dans les méthodologies utilisées. Les trois méta-analyses les plus récentes suggèrent néanmoins qu'une supplémentation en vitamine D est efficace pour diminuer le risque de chute chez la personne âgée (36-38). Il a été suggéré de prescrire une dose de minimum 700 UI/jour et de l'associer à une supplémentation de calcium afin d'avoir un effet plus important sur la diminution du risque de chute. Cet effet de la vitamine D reste toutefois modeste avec une diminution du nombre de chutes de l'ordre de 15 à 20%.

L'indication de traitements antiagrégants ou anticoagulants sera évaluée à la lumière du risque de chute résiduel, même si le risque de chute ne paraît pas une contre-indication absolue à ces traitements (39). Le clinicien veillera également à réduire les sédatifs, les médications à clairance rénale stricte et les médications anabolisantes. Enfin, une télé-biovigilance est conseillée.

Conclusion

La chute de la personne âgée est un évènement fréquent, aux conséquences potentielle-

ment graves, parfois révélateur de pathologies associées. Il s'agit d'un réel problème de santé publique, mobilisant de nombreuses équipes attelées à la recherche de solutions de prévention, de soins adaptés, et de moyens de réduction des coûts secondaires. Ainsi, toute personne âgée fragile doit bénéficier d'un bilan de dépistage tel que réalisé dans les hôpitaux de jour gériatriques. Enfin, la chute de la personne âgée reste un sujet de constante recherche de la part des cliniciens gériatres, neurologues, radiologues et anatomopathologistes, ce qui en fait un sujet passionnant alimentant l'imaginaire des chercheurs et des soignants.

Bibliographie

1. Bureau fédéral du plan et direction générale statistique: Perspectives de population 2010-2060. Editeur responsable BFP. Henri Bogaert; éditeur responsable DGSIE. Anne Versonnen, Décembre 2011.
2. Tinetti ME, Kumar C.— The patient who falls. *JAMA*, 2010, **303**, 258-266.
3. Tinetti ME.— Clinical practice. Preventing falls in elderly persons. *N Engl J Med*, 2003, **248**, 42-49.
4. Ambrose AF, Paul G, Hausdorff JM.— Risk factors for falls among older adults : a review of the literature. *Maturitas*, 2013, **75**, 51-61.
5. Hartholt KA, van Beeck EF, Polinder S, et al.— Societal consequences of falls in the older population: injuries, healthcare costs, and long-term reduced quality of life. *J Trauma*, 2011, **71**, 748-53.
6. Rochat S, Büla CJ, Martin E, et al.— What is the relationship between fear of falling and gait in well-functioning older persons aged 65 to 70 years? *Arch Phys Med Rehabil*, 2010, **91**, 879-884.
7. Pfitzenmeyer P, Mourey F, Tavernier B.— Psychomotor desadaptation syndrome. *Arch Geront Geriatrics*, 1999, **28**, 217-225.
8. Manckoundia P, Mourey F, Tavernier B, et al.— Psychomotor desadaptation syndrome. *Rev Med Interne*, 2007, **28**, 79-85.
9. Weber J, Gillain S, Petermans J.— La sarcopénie : un marqueur de fragilité. *Rev Med Liège*, 2010, **65**, 514-20.
10. Ronthal M.— Gait disorders. Butterworth-Heinemann, Boston Mass, Oxford, 2002.
11. Les troubles de la marche.— Collège des enseignants en neurologie
12. Glass GA, Ahlskog JE, Matsumoto JY.— Orthostatic myoclonus : a contributor to gait decline in selected elderly. *Neurology*, 2007, **68**, 1826.
13. G Allali, F Assal, R Kressig, et al.— Impact of impaired executive function on gait stability. *Dement Geriatr Cogn Disord*, 2008, **26**, 364-369.
14. G Yogev-Seligmann, J Hausdorff, N Giladi.— The role of executive function and attention in gait. *Mov Disord*, 2008, **23**, 329-342.
15. Scherder E, Eggermont L, Swaab D, et al.— Gait in ageing and associated dementias; its relationship with cognition. *Neurosci and Biobehav Rev*, 2007, **31**, 485-497.
16. Gillain S, Petermans J.— Contribution of new techniques to study gait in old populations. *Ann Phys*
17. Podsiadlo D, Richardson S.— The timed «up and go»: a test of basic functional mobility for frail elderly persons. *J Am Geriatr Soc*, 1991, **39**, 142-148.
18. Beauchet O, Fantino B, Allali G, et al.— Timed up and go test and risk of falls in older adults : a systematic review. *J Nutr Health Aging*, 2011, **15**, 933-938.
19. Studensky S, Perera S, Patel K.— Gait speed and survival in older adults. *JAMA*, 2011, **305**, 50-58.
20. Beauchet O, Dubost V, Allali G, et al.— Faster counting while walking' as a predictor of falls in older adults. *Age Ageing*, 2007, **36**, 418-423.
21. Beauchet O, Annweiler C, Kressig R.— Stop walking when talking : a predictor of falls in older adults. *J Neurol Neurosurg Psychiatry*, 2004, **75**, 994-997.
22. Verghese J.— Mobility stress test approach to predicting frailty, disability and mortality in high-functioning older adults. *JAGS*, 2012, **60**, 1901-1905.
23. Beauchet O, Hélard L, Mateus-Hamoan L, et al.— Hypovitaminosis D in geriatric acute care unit. A biomarker of longer length of stay. *J Nutr Health Aging*, 2013, **35**, 525-529.
24. Maki B.— Gait changes in older adults : predictors of falls or indicators of fear. *J Am Geriatr Soc*, 1997, **45**, 313-320.
25. Gillain S, Warzee E, Petermans J.— The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects and a comparison with other clinical tests used in single and dual-task conditions. *Ann Phys Rehab Med*, 2009, **52**, 453-474.
26. Bridenbaugh S, Kressig R.— Laboratory review : the role of gait analysis in seniors'mobility and fall prevention. *Gerontology*, 2011, **57**, 256-264.
27. Killey B, Watt E.— The effect of extra walking on the mobility, independance and exercise self-efficacy of elderly hospital in-patients a pilot study. *Contemp Nurse*, 2006, **22**, 120-33.
28. Simonsick EM, Guralnik JM, Volpato S.— Just get out the door! Importance of walking outside of home for maintaining mobility: findings from the women health and aging study. *J Am Geriatr Soc*, 2005, **53**, 198-203.
29. Li F, Hamer P, Ficher KJ.— Tai Chi and fall reductions in older adults: a randomized controlled trial. *J Gerontol A Biol Sci Med Sci*, 2005, **60**, 187-94.
30. Trombetti A, Haus M, Hermann Fr.— Effect of music-based multitask training on gait, balance, and fall risk in elderly. *Arch Intern Med*, 2011, **171**, 525-533.
31. Blankevoort C, van Heuvlen M, Boersma F, et al. — Review of effects of physical activity on strength, balance, mobility and ADL performance in elderly subjects with dementia. *Dement Geriatr Cogn Disord*, 2010, **30**, 392-402.
32. Schenk M, Zieschang T, Oster P.— Dual-task performances can be improved in patients with dementia. *Neurology*, 2010, **74**, 1961-1968.

33. Body JJ, P. Bergmann, S. Boonen, et al.— Non-pharmacological management of osteoporosis : a consensus of the Belgian Bone Club. *Osteoporos Int*, 2011, **22**, 2769-2788.

34. Rizzoli R, Boonen S, Brandi ML.— Vitamin D supplementation in elderly or postmenopausal women : a 2013 update of the 2008 recommendations from the European Society for Clinical and Economic and Economic Aspects of Osteoporosis and Osteoarthritis (ESCEO). *Curr Med Res Opin*, 2013, **29**, 1-9.

35. Bischoff-Ferrari HA, Dawson-Hughes B, Staehelin HB.— Fall prevention with supplemental and active forms of vitamin D : a meta-analysis of randomised controlled trials. *BMJ*, 2009, **339**, b3692.

36. Kalyani RR, Stein B, Valiyil R, et al.— Vitamin D treatment for the prevention of falls in older adults: systematic review and meta-analysis. *JAGS*, 2010, **58**, 1299-1310.

37. Murad MH, Elamin KB, Abu Elnour NO, et al.— Clinical review : the effect of vitamin D on falls: a systematic review and meta-analysis. *J Clin Endocrinol Metab*, 2011, **96**, 2997-3006

38. Kanis JA, Mc Closkey EV, Johansson H.— European guidance for the diagnosis and management of osteoporosis in postmenopausal women. *Osteoporos Int*, 2013, **24**, 23-57.

39. Donzé J, Clair C, Hug B, et al.— Risk of falls and major bleeds in patients on oral anticoagulation therapy. *Am J Med*, 2012, **125**, 773-778.

Les demandes de tirés à part sont à adresser au Pr. J. Petermans, Service de Gériatrie, CHU de Liège, Belgique.
Email : jean.petermans@chu.ulg.ac.be

Available online at
SciVerse ScienceDirect
www.sciencedirect.com

Elsevier Masson France
EM|consulte
www.em-consulte.com

Annals of Physical and Rehabilitation Medicine 56 (2013) 384–395

ANNALS OF PHYSICAL AND REHABILITATION MEDICINE

Original article / Article original

Contribution of new techniques to study the gait in old populations

Contribution des nouvelles techniques à l'étude de la marche chez la personne âgée

S. Gillain *, J. Petermans

Geriatric department, CHU Liège, Liège, Belgium

Received 28 June 2012; accepted 25 May 2013

Abstract

Objectives. – For several years, the concept of ''physiological senile gait'' has been strongly contested and seems to be associated with abnormal gait. Indeed, some changes characteristic of senile gait appear early on in subjects with neurodegenerative pathologies. The aim of this article was to determine how recent contributions can improve the study of gait in old populations. This paper is a thematic review of recent contributions from medical imaging techniques as well as instrumental gait analysis techniques in older adults. This article did not focus on Parkinson's disease or other specific diseases bearing certain gait disturbances, since they belong to literature focusing on these particular disorders.
Material and methods. – This work was not intended as a systematic review but only as a thematic one conducted by geriatricians in order review the recent literature in order to better apprehend how new technics could be implemented within their clinical practice. Articles were selected in online Medline and Cochrane Library databases, and some were previously identified by the authors.
Results. – This paper highlights the most recent contributions in magnetic resonance imaging, functional magnetic resonance imagery, positron emission tomography and instrumental gait analyzing devices better understanding the underlying gait mechanisms in elderly populations.
Conclusions. – This thematic review suggests that gait could be considered as a marker of ''successful aging''. Its evaluation associated to longitudinal follow-up could be useful to predict cognitive and functional changes in frail older adults.

Keywords: Gait; Aging; Imagery; Cognitive; Instrumental

Résumé

Objectif. – Depuis quelques années, le concept de « démarche sénile physiologique » est fortement contesté. Certaines modifications de la marche présentes chez le sujet âgé semblent le plus souvent être associées à des processus pathologiques débutants. Cet article propose une revue thématique des apports récents obtenus par les techniques d'imagerie médicale et les techniques instrumentales d'analyse de la marche appliquées aux personnes âgées. La maladie de Parkinson ainsi que d'autres pathologies incluant des modifications de la marche déjà décrites lors de publications spécifiques, ne sont pas spécifiquement évoquées dans cet article.
Matériel et méthodes. – Les références citées dans cet article ont été en partie sélectionnées en recherchant dans la Cochrane Library et dans Medline. Certains articles utilisés étaient déjà connus des auteurs. Cet article ne prétend pas être une revue systématique mais une revue thématique faite récemment par des gériatres cliniciens afin de rassembler les données récentes de la littérature et envisager leur intégration lors de leur pratique quotidienne.
Résultats. – Cet article résume les contributions récentes de l'imagerie par résonance magnétique structurelle et fonctionnelle, de la tomographie par émission de positons et des instruments d'analyse des paramètres de marche permettant une meilleure connaissance des mécanismes sous-jacents de la démarche chez les personnes âgées.
Conclusion. – La qualité de la marche pourrait être considérée comme un marqueur de « vieillissement réussi » et l'analyse de sa détérioration pourrait aider à prévoir les changements cognitifs et fonctionnels y afférant.

Mots clés : Marche ; Vieillissement ; Imagerie ; Cognition ; Instrumentale

* Corresponding author.
E-mail address: sgillain@chu.ulg.ac.be, s.gillain@gmail.com (S. Gillain).

1877-0657/$ – see front matter
http://dx.doi.org/10.1016/j.rehab.2013.05.002

1. English version

1.1. Introduction

Over the last few years, studying gait in older adults has raised a great deal of interest. Indeed, gait changes can lead to falls increasing the risk of institutionalization, functional decline and onset of comorbidities [39]. Therefore, gait changes in older adults can be considered as a frailty marker. Recent studies have underlined that changes in certain gait parameters could be considered as a sign of cognitive frailty [11,37]. Therefore, the concept of a "physiological senile gait" has been strongly argued and the question in fact remains: Is the "physiological senile gait" truly physiologically norm or could it reflect the onset of some pathological disorders, such as cognitive ones, leading to falls? Furthermore, could there be a cut-off discriminating healthy older adults from frail ones?

It is well known that gait results from a motor command stemming from the central nervous system and is executed by the peripheral nervous system. The execution of this motor command involves not only the vestibular and peripheral nervous systems but also good quality osteoarticular and muscle features, without forgetting the cardiovascular system. Clinical features are the first step to evaluate gait, including thorough medical history (surgeries, treatments, and preexisting diseases), physical examination and other clinical tests, these elements are essential to understand the locomotor abilities in elderly populations. Nevertheless, nowadays there are two core components for analyzing in-depth gait features: anatomical and functional analysis of the central nervous system in charge of managing motor commands and measures of gait spatiotemporal parameters. In fact, motor command organization can be studied through imaging techniques such as magnetic resonance imaging (MRI), functional MRI (fMRI) and positron emission tomography (PET). The tests and tools for gait analysis used in daily clinical practice have shown their limits compared to instrumental gait analysis [11,43]. Walkway and accelerometric methods allow analyzing spatiotemporal gait parameters and these techniques have recently been used in old populations. The purpose of this article is to comment new, useful and promising techniques for gait analysis in elderly subjects. This article did not specifically focus on Parkinson's disease or other pathologies where gait changes have already been defined in specific domains within the literature.

1.2. Method

The aim of this article was not to conduct a systematic review of the literature with precise inclusion or exclusion criteria, but instead to propose a thematic review based the most recent articles and those already known by the authors. The literature selection was based on:

- their contribution to the cover a large overview input within a large review of gait analysis;
- experience of the working team in gait studies;
- practical aspect of the knowledge contained within the article. In a second time, the references of each article were studied and some articles were selected by this way.

To select the most recent articles, we carried out a search in the Cochrane Library using the keywords "gait AND older". This resulted in eight articles but in fact, only two were focused on gait (specifically gait rehabilitation) in older adults. In the same database, using the keywords "gait AND elderly", only one article was found and this article only concerned gait rehabilitation. No articles were found on the input of medical imaging and instrumental gait analysis.

We searched for articles in MEDLINE with the same keywords and limited our search to "items with links to full text, Human, English, French and Age > 65 years published within the last 10 years". We found more than 900 references. The most interesting references were selected based on the previously reported criteria. Forty-seven abstract were initially considered to amount to a final of 12 articles.

1.3. Results

1.3.1. Medical imaging contribution

1.3.1.1. Magnetic resonance imaging. Table 1 lists the following commented studies. Zimmerman [46] highlighted a correlation between the anatomical structures of the hippocampus and changes in gait parameters. In a group of 48 cognitively healthy older adults, mean age of 81 years, hippocampus volume was measured using volumetric MRI and its metabolism evaluated by proton nuclear magnetic resonance spectrometry. A correlation between step length and hippocampal volume was unveiled, as well as a relationship between step length variability and hippocampal metabolism. Some studies reported a relationship between vascular lesions and gait changes [24,26]. A recent study evaluated mobility in 331 adults with no history of strokes and not affected by dementia or Parkinson's disease. In this group of healthy older adults, the vascular lesions in the white matter and subclinical strokes were associated with slower gait speed, shorter stride length and longer double support time [30]. Furthermore, the same study indicated a relationship between a greater step length variability and higher prevalence of infarcts, including basal ganglia infarcts [31].

Morphological MRI has also promoted the "vascular" theory to explain gait changes in relations to cognitive disorders linked with aging. Age itself has been related to microangiopathy. In two similar publications, the role of arteriosclerotic brain lesions in the development of dementia in patients with mild cognitive impairment (MCI) has been demonstrated [14,28].

In previous studies, vascular lesions have been identified as a significant risk factor for developing gait changes and/or cognitive decline. The risk of falls depends on how vascular lesions are distributed in the frontal region in charge of executive functions and organization of motor commands. Another study highlights the relevance of subcortical lesions not only in frontal regions but also in seven specific regions: frontal, temporal, parietal, and occipital lobes, basal ganglia,

Table 1
Summary of new knowledge from magnetic resonance imaging (MRI).

First authors and references	Observations
Zimmerman et al., Brain Res, 2009	Correlation between steps length and the hippocampal volume and between the variability of the steps length and the hippocampal metabolism
Onen et al., Neurosci Lett 355, 2004 Onen et al., Brain Res, 2008	Relationship vascular lesions in specific regions and gait disorders in MCI
Onen, J Neuroradiol, 2005 [25]	Role of microangiopathy and circulation of cerebral spinal fluid and gait disorders in MCI
Rosano et al., Neuroepidemiology, 2006	Vascular lesion in white matter and subclinical strokes are associated with slower gait speed, shorter stride length and longer double support time
Rosano et al., Neuroepidemiology, 2007	Infarcts in basal ganglia are associated with variability of step length
Pugh and Lipsitz, Neurobiol Aging, 2002	Age itself is associated with microangiopathy
Henry-Feugeas et al., Clin Interven Aging, 2008	Role of arteriosclerotic brain lesions in the development of dementia in patients with MCI
Guerini et al., Arch Phys Med Rehabil, 2008	Subcortical hypoperfusion interrupting long loops reflexes of deep white-matter motor tracts and descending motor fibers arising from medial cortical areas

MCI: mild cognitive impairment.

internal capsule, external capsule and cerebellum. A multivariate analysis reported that subcortical vascular lesions were the sole significant predictor for the risk of falls in elderly populations. According to the authors, this could be related to the subcortical hypoperfusion interrupting long loop reflexes critical for gait and balance mediated by deep white matter sensory and motor tracts as wells as descending motor fibers arising from medial cortical areas [12].

To sum up, it would seem that morphological MRI has unveiled a correlation between presence of vascular lesions, atrophy and hippocampal metabolism to changes in gait and subsequent relations to the cognitive evolution of older adults with MCI.

1.3.1.2. Functional imaging. Table 2 lists the studies discussed. Structures involved in gait control can be evaluated with techniques such as single photon emission tomography with technetium (HMPAO-SPECT), near-infrared spectroscopy (NRIS), transcranial magnetic stimulation (TMS), electroencephalography (EEG), fMRI and PET. Depending on technical limitations, fMRI and PET are the techniques most often used [4] and this is why we chose to only focus on these techniques. Furthermore, some studies reported in this paragraph concern young subjects because evaluations have yet to be conducted in older adults. Functional MRI does not use a radioactive tracer, but instead it is the concentration of desoxyhemoglobine which reflects a rise in the activation level. This technique distinguishes the cortical or subcortical regions activated during the task requested by the examiner however, fMRI (requiring for the patient to remain still) is not able to study brain activation directly during the real task, but rather the secondary activation obtained when imagining this task. In 2006, Bakker et al. [3] demonstrated that characteristics (time needed) involved in the motor imagery of walking compared to real walking under the same conditions (path length and path width) in young people. They concluded to a high temporal

Table 2
Summary of new knowledge from functional imagery.

First author, technics and references	Observations
Bakker et al., J Neural Transm, 2007	fMRI and PETscan are most often used to study gait
Bakker et al., fMRI, Exp Brain Res, 2007	High temporal correspondence between real and imagined walk in young people
Personnier et al., Neurosci Lett, 2010	Old people are at risk of overestimating the duration of imagined movements
Beauchet et al., J Neurol Sci, 2010	Old people are at risk of underestimating the duration of imagined movements
Malouin et al., PETscan, Hum Brain Mapp, 2003	Role of the pre-supplementary motor area, leg area of the motor cortex and basal ganglia in voluntary locomotor movements
Jahn et al., fMRI, Neuroimage, 2004	Standing: activation in thalami, left putamen, left frontal gyri and vermis Walking: activation in SMA, parahippocampal and fusiform gyri, occipital visual areas, inferior frontal gyri, left putamen and vermis
Jahn K, fMRI, Brain Res, 2008	Importance of activity in parahippocampic and fusiform gyri when precised gait is needed
Deshpande and Patla, Galvanic vestibular stimulations, Brain Res, 2007	Decreasing down regulation capacity of vestibular information with age
Zwergal et al., fMRI, Neurobiol Aging, 2012	Cortical inhibition and interaction between sensory systems during locomotion and stance declines with age
La Fouchère, fMRI and PET, Neuroimage, 2010	Role of basal ganglia in the initiation of gait
Vidoni et al., fMRI, J Neurol Phys Ther, 2012	Decreased activation in accessory motor regions, supplementary motor area and cerebellum in early stages of Alzheimer's disease and increased coactivation in primary motor cortex of bilateral motor and visual regions in AD subjects

AD: Alzheimer's disease; SMA: supplementary motor area.

correspondence between real and imagined walking and unveiled the possibility of studying the central components of gait with this technique. To our knowledge, only two studies comparing young and older people have been conducted. One study demonstrated that older subjects overestimated the duration of imagined movements [27], while the other study showed that the imagined performance was faster than the real physical performances [7]. In these studies, the first step could have been to systematically consider the comparison between time needed to perform the real task and time required for imagining that same task. Furthermore, considering the available scientific literature, the population studied by functional imaging must be completely defined and the conclusions may not be extrapolated to other populations. In 2003, Malouin et al. [21], in a PET study assessing six healthy adults (mean age: 55.9 years) compared brain activation during motor imagery of locomotor-related tasks and reported the involvement of the pre-supplementary motor area (pre-SMA) as well as the leg area of the motor cortex in conditions requiring locomotor movements (walking, initiating gait and walking over an obstacle). This study also suggested that the basal ganglia played a role in locomotor movements, which are automatic by nature. In 2004, Jahn et al. [17], in a study focusing on 13 healthy young subjects (mean age: 27.3 years), compared cerebral activation during standing, walking and running. Standing was associated with activation in both thalami (more to the right), the left putamen, left frontal gyrus and cerebellar vermis and there were no deactivation areas. Walking demonstrated activation in additional motor areas like SMA, parahippocampal gyrus and fusiform gyrus as well as occipital visual areas, inferior frontal gyrus, left putamen and the vermis as well as the anterior lobe hemisphere of the cerebellum. When subjects imagined themselves walking, there was an activation in the vermis, (next to the area activated during walking) and in the pontomedullary area, furthermore a decreased activation of the fusiform gyrus and parahippocampal gyrus was highlighted (just like during walking). In 2008, in a study evaluating nineteen healthy young adults (mean age: 33.5 years), the same authors asked the subjects to represent themselves walking along a curved line, using the same method published in their previous study, and they validated the importance of the activation in the parahippocampal and fusiform gyrus when visual navigation was required to perform a precise walking task. However, deactivations were noted in the superior and medial temporal gyrus. Jahn et al. suggested that this deactivation could decrease the importance of vestibular afferents and avoid a possible conflict between visual and vestibular inputs [41]. This modulation of the information coming from the vestibular system was confirmed in a study by Deshpande and Patla using galvanic vestibular stimulation (GVS) to the mastoid bone, the authors noted the effect of vestibular disruptions in the locomotor performance of young (nine healthy subjects aged 20–35 years) and healthy older adults (age 65–85 years) [9]. This study showed that in young populations, the vestibular stimuli can be down-regulated in favor of better visual stimuli. However, this down regulation does decline with in age and thus older adults are more prone to impaired balance when vestibular and visual stimuli differ. Similarly, it has been demonstrated that the mechanism of cortical inhibitory reciprocal interaction between the sensory systems as observed during locomotion and stance phases does decline with age [47]. These data seem relevant when studying gait in older adults with sensory impairments.

Finally, the author compared the results from two functional imaging techniques: fMRI and, 18F-FDG PET, and validated the role of basal ganglia in gait initiation (GI) in a population that included sixteen healthy adults (age range: 51–73, mean age: 61.3 ± 7.8 years) [18].

Briefly and according to the author's research, the stance phase seems modulated by the brainstem and the cerebellum while specific area have various well-defined roles during gait: the brainstem seems to involved in gat initiation and ending, the cortical areas, basal ganglia and thalami are implicated in GI and changes in direction.

The basal ganglia and cortical areas are responsible for gait changes for following specific direction or avoiding obstacles. The cerebellum seems to command gait rhythm and velocity. Cortical areas like SMA are involved the cognitive control of gait and its importance increases with age. In fact, while older adults are less able to down-regulate vestibular inputs, the cortical component seems more important during directed gait and changes of direction. According to the recent literature, the activity of the cortical network increases with age and thus could explain difficulties in performing dual tasks. A recent study has reported that the network organizing gait evolves along a person's lifetime and according to certain specific disorders [40]. Then, in the early stages of Alzheimer's disease (AD), patients exhibit a decreased activation in the accessory motor areas, the supplementary motor area and the cerebellum in comparison to subjects without dementia. At the same time AD subjects show an increased coactivation of bilateral motor and visual regions with the primary motor cortex.

Considering the actual literature, available studies most often concern younger people. Very few have focus on older adults and especially elderly populations with cognitive disorders. This lack of data could be explained by the difficulties in convincing older adults to try these technics, as they are not easy to reach and the techniques are complicated to implement. Another difficulty seems to lie in the interpretation of the results as these cannot be extrapolated to other populations, furthermore carrying out a comparison between ''healthy'' older adults and adults with specific disorders is quite difficult and most often includes bias.

1.3.2. Contribution of the spatio-temporal parameters of gait

The Table 3 lists the studies discussed below.

Today, to conduct a precise study on spatiotemporal gait's parameters there is a choice between treadmills (i.e., GaitRite®) or accelerometer devices (i.e., Locométrix® or Dynaport®). Many teams have used the GaitRite® system (however other types of treadmills are available). This treadmill comes in different distances and is equipped with pressure sensors to measure spatiotemporal parameters. Collected data

Table 3
Contribution of the spatio-temporal parameters of gait.

Author, references	Observations
Bridenbaugh and Kressig, Gerontology, 2011	Instrumental method more precise than clinical tests
Wang et al., J Neural Transm, 2009	Attentional component of gait increase with age
Verghese et al., J Gerontol A Biol Sci Med Sci, 2009	Gait speed and variability of gait are two marker of risk to fall
Hausdorff et al., Arch Phys Med Rehabil, 2001	An increase of 0.017 m in the length of the step in a cycle of step doubles the risk to fall
Maki, J Am Geriatr Soc, 1997	Variability of gait is a marker of fear of falling not a marker of fall risk
Beauchet et al., JAGS, 2008	Interest of study the speed and the length of the step and the variability of gait during dual task considering the risk of falls
Dubost et al., Hum Mov Sci, 2006	
Allali et al., Dement Geriatr Cogn Disord, 2008	Importance of executive functions considering the variability of the gait
Sheridan et al., JAGS, 2003	Importance of divided attention capacities considering the variability of the gait
Holtzer et al., Neuropsychology, 2007	Importance of cognitive functions considering the variability of gait speed
Allali et al., Mov Disord, 2010	Specific gait's profile considering people with AD or fronto-temporal dementia
Verghese et al., J Neurol Neurosurg Psychiatry, 2007	Modifications of gait's profile could help to predict the cognitive evolution of MCI
Bautmans et al., Gait Posture, 2011	Reliability between the results obtained by Dynaport® and GaitRite® when based on two walks over 18 m and correlation with falls and functional status in older subjects
Gillain et al., Ann Phys Rehabil Med, 2009	An accelerometric device allows to highlight specific gait's profiles according to the cognitive status of older subjects

MCI: mild cognitive impairment; AD: Alzheimer's disease.

consist in speed, number of gait cycles, step length and width as well as duration of the double stance period. Then, by calculating the variables' coefficient of variation (CV) and comparing each gait cycle to the other, it is possible to express gait "variability".

As well-detailed in a recent review regarding the relevance of gait analysis, these technics help to improve the precision of gait profiles, much better than clinical tests alone could [8]. Then, in the last ten years, studies which have used these techniques have reported that with age walking requires more attention [42].

Some authors have shown that gait speed and variability are two predictors for the risk of falls in older adults [38]; a variation of 0.017 m in step length within a gait cycle actually increases this risk twofold [13]. A few years ago, Brian Maki using footswitches, reported that decreased speed, reduced step length and increased duration of the double stance period were independent fall predictors. However in the same study the author highlighted that step-to-step variability was not related to the risk of falls but rather associated with the fear of falling [20]. Unfortunately, a comparison between these three studies remains difficult without a complete and detailed description of the three different populations and the patients' potential history of falls prior to the evaluation.

Using GaitRite®, numerous authors have demonstrated the relevance of analyzing gait during dual task, i.e. asking the subject to walk while simultaneously completing a cognitive task. Three parameters, gait speed step length and variability of gait parameters, seem good predictors for the risk of falls in this condition [6,10].

Other teams have studied older adults at the onset of cognitive disorders or Alzheimer's disease and have reported the influence of memory and executive functions [1] but also the importance of divided attention capacities [32] on the variability of gait parameters, mainly gait speed [15]. Most recently, a significant relationship was reported between an increased CV for step time (i.e., time needed to complete a gait cycle) in older adults with frontotemporal dementia, compared to older adults suffering from AD and to healthy subjects [2].

In a survey of 400 subjects without cognitive disorders, a cognitive assessment and instrumental study of their gait parameters were performed at the beginning of the study and after 5 years. At the 5-year visit, the authors described a significant correlation ($P = 0.02$) between the evolution of gait speed, step length and global decline in cognitive and executive functions, once variables had been adjusted for age, sex and level of education [36].

A recent study reported the reliability of the results obtained using a triaxial accelerometer device (Dynaport®) based on two 18- meter walking courses. Furthermore, the results obtained were correlated with falls and functional status [5] in older adults. In this context, our team analyzed gait parameters with a triaxial portable accelerometer, the Locométrix®. This device, fitted around the patient's waist with an elastic belt, recorded the accelerations in the antero-posterior, medio-lateral and cranio-caudal axes. Gait parameters were analyzed in a population of healthy older adults, adults with MCI and adults with Alzheimer's disease (AD). In all three groups, the subjects had to walk along a straight, 40-meter long corridor, in a simple task (walking) and dual task (walking while counting backwards). In both conditions, the Locométrix® differentiated the three populations based on their gait profile related to their cognitive status [11].

The instrumental gait analysis seems quite relevant for discriminating one or several spatiotemporal variables that could increase the risk of falling and help identify subjects with cognitive disorders based on their gait profile and thus discern those at risk of developing dementia.

These methods for studying gait parameters have shown that age was not the only element to explain gait variability. In fact, factors such as cognitive disorders, histories history of falls, given walking speed and gait conditions (single or dual task) do have an impact on gait in older populations and remain useful indicators to evaluate the risk of falls or predict cognitive evolution. Then, considering these parameters and in order to assess gait it is essential to refine gait conditions, population characteristics (not only age but history of falls, medications,

comorbidities, sensory impairments, neurologic disorders. . .), required walking speed and evaluation technic used. It seems important to note that results from a study with a specific technic conducted on a specific population and using a specific protocol cannot be extrapolated to other situations. In light of the above, teams who are studying gait do encounter some general difficulties in geriatric populations: design of the study and finding a balance between the need to highlight some results with a strong statistical power and the relevance of underling this result in a ''real-life'' population.

1.3.3. Gait initiation

More recently, several teams have tried with various techniques to study the precise onset of GI. Indeed, this moment seems particularly dangerous for older adults presenting a risk of falls. GI is a voluntary transition from a stable double stance position to a continuously unstable posture when walking, which requires anticipatory postural adjustments to shift the center of mass towards the supporting side to lift-up the leg [33]. As showed by single-photon or positron emission tomography [21] or repetitive transcranial magnetic stimulation [16] GI requires the activation of the SMA and basal ganglia also seem to be playing a crucial role as demonstrated in subjects with in Parkinson's disease (PD) subject [44]. GI is often evaluated using force platforms. In the past years, several teams have studied GI in a population of healthy older adults [22], in older people with fear of falling [34], fallers [23], subjects with PD [29] and AD [45]. GI seems to vary according to the population studied. Further studies should be conducted to validate and refine these observations.

1.4. Discussion and conclusion

Several studies have reported the relevance of gait assessment in older populations. New imaging techniques and instrumental analyses enable to underline the relationship between anatomy and metabolism of the central structures and spatiotemporal gait parameters. Gait seems to be an indicator of ''successful aging'' by predicting cognitive and functional changes. Unfortunately, and despite very useful elements provided by these new techniques, they remain quite confidential, restrained to an experimental setting and are not widely available in clinical practice. Step length, gait speed and variability seem essential indicators of gait quality. Moreover, studying gait in dual task condition could bring information on the risk of falls, and cognitive reserve allocated for gait organization. Furthermore, in this elderly population it seems quite relevant to conduct systematic gait evaluations in single and dual task conditions associated with a long-term follow-up.

Finally, GI seems a crucial element to evaluate safety in older adults. Studying this specific moment, could better refine the underlying mechanisms involved in gait and balance and work towards improving them.

In our opinion, clinical tests remain essential for evaluating quickly and efficiently the gait and balance of older adults, available tests include the 10-meter walking test to evaluate speed, the Timed-up-and-go test, the short physical performance battery including the Sit-to-Stand test, the Berg or Tinetti balance tests test should remain the best way to perform a quick useful study of the balance of older subjects. Concerning dual task condition, the Stop walking when talking test represent a very good predictor of risk [19]. Indeed, comparing gait speed in single and dual task is a relevant screening tool to evaluate the risk of falls; with a low 40% sensitivity but a high 96% specificity [35].

In choosing a systematic method for gait assessment, the most important thing would be to select some easy to perform test, reproducible under the same conditions and well known by the examiner in order to be systematically repeated. Finally, a hand grip test or Martin's vigorimeter could be relevant to evaluate muscle mass in elderly populations and predict the evolution of their and future functional status. In the future, the use of these new technics, enhancing ''central'' or ''peripheral'' markers with a predictive value for functional and cognitive decline, might improve the detection of people at risk of falling and enable teams to propose multidisciplinary screening evaluation and monitoring follow-ups. These studies might permit the design of adapted and specific gait but also cognitive rehabilitation protocols, in a population of frail older adults.

Disclosure of interest

The authors declare that they have no conflicts of interest concerning this article.

2. Version française

2.1. Introduction

Au cours de ces dernières années, l'étude de la marche chez le sujet âgé a suscité beaucoup d'attention. En effet, une altération de la marche peut entraîner des chutes et augmenter le risque de placement, de déclin cognitif et la survenue de comorbidités [39]. C'est pourquoi les changements de la marche peuvent être considérés comme un indicateur de fragilité. Des travaux récents montrent que les altérations de certains paramètres de la marche peuvent être considérées comme un signe de fragilité cognitive [11,37]. Le concept de « démarche sénile physiologique » est donc contesté. Le questionnement actuel du clinicien est : – La « démarche sénile physiologique » est vraiment physiologiquement normale ou est-elle le reflet d'un processus pathologique débutant, comme un trouble cognitif, pouvant entraîner des chutes ? – Et qu'en est-il d'un éventuel seuil permettant de différencier la personne âgée en bonne santé de la personne âgée fragile ?

Il est admis que la marche est le résultat d'une commande motrice émanant du système nerveux central et exécutée au niveau du système nerveux périphérique. L'exécution de la commande motrice est possible grâce à l'implication du système vestibulaire, du système nerveux périphérique et du système cardiovasculaire mais également par une bonne qualité musculaire et ostéoarticulaire.

Pour étudier la marche, il est nécessaire d'étudier les aspects cliniques. Une liste exhaustive des antécédents (chirurgies, traitements médicaux, conditions préexistantes), examen physique et tests cliniques sont essentiels à la compréhension de la mobilité de la personne âgée. De plus, il existe deux voies d'analyse de l'organisation spécifique de la marche : l'analyse anatomique et fonctionnelle du système nerveux central qui organise la commande motrice et la mesure des paramètres spatiotemporels de la marche. L'organisation de la commande motrice peut actuellement être précisée grâce aux techniques d'imagerie par résonance magnétique (IRM), IRM fonctionnelle (IRMf), et tomographie par émission de positons (PET). Quant à l'étude précise des paramètres de marche, les tests cliniques utilisés en pratique quotidienne ont rapidement montré leurs limites comparé aux analyses instrumentales des paramètres de marche [11,43]. En effet, les tapis de marche et les accéléromètres permettant l'analyse des paramètres spatiotemporels de la marche, ont été récemment utilisés chez le sujet âgé.

L'objectif de cet article est de discuter des nouvelles techniques d'analyse de la marche chez le sujet âgé. La maladie de Parkinson ainsi que d'autres pathologies incluant des modifications de la marche déjà décrites lors de publications spécifiques, ne sont pas abordées.

2.2. Méthode

Le but de cet article n'est pas de mener une revue systématique de la littérature. Les auteurs proposent ici une revue thématique basée sur les articles récents et d'autres précédemment connus des auteurs. Les articles ont été sélectionnés selon :

- leur contribution à une large revue thématique ;
- l'expertise des principaux auteurs concernant l'étude de la marche ;
- l'aspect pratique des connaissances et découvertes citées dans l'article.

Ensuite, les auteurs ont analysé la bibliographie de chaque article afin de sélectionner d'autres références contributives. Les recherches bibliographiques ont été menées au sein de la Cochrane Library avec les mots clés « gait AND older » (marche ET âgé). Nous avons identifié ainsi huit articles dont deux décrivaient spécifiquement la rééducation de la marche chez la personne âgée. Dans la même base de données et en utilisant les mots « gait AND elderly » (marche ET personnes âgées), nous n'avons retrouvé qu'un seule article portant sur la rééducation fonctionnelle uniquement. Aucun article n'a été retrouvé sur la contribution de l'imagerie et des techniques d'analyse instrumentale de la marche. Les auteurs ont ensuite cherché via MEDLINE avec les mêmes mots clés en limitant avec « articles complets uniquement, Humains, Anglais, Français, Âge : +65 ans, publiés au cours des dix dernières années ».

Nous avons retrouvé plus de 900 références. Les plus intéressantes étaient sélectionnées en fonction des critères décrits plus haut. Initialement, 47 résumés furent considérés et seulement 12 articles finalement retenus.

2.3. Résultats

2.3.1. Apport de l'imagerie

2.3.1.1. L'imagerie à résonance magnétique. Le Tableau 1 résume les différentes études décrites.

Dans un groupe de 48 patients âgés et cognitivement sains (âge moyen de 81 ans), le volume de l'hippocampe a été mesuré par IRM volumétrique et son métabolisme par tomographie par émission de positons [46]. Les auteurs ont montré une corrélation entre la longueur du pas et le volume hippocampique ainsi qu'une relation entre la variabilité de la longueur du pas et le métabolisme hippocampique.

Concernant l'influence des lésions vasculaires, une étude récente vient confirmer les résultats précédemment obtenus [24,26]. Les auteurs ont évalué la mobilité chez 331 adultes sans antécédents d'AVC, démence ou maladie de Parkinson. Dans ce groupe de personnes âgées en bonne santé, les lésions vasculaires de la substance blanche et les AVC silencieux étaient reliés à une vitesse de marche plus lente, une longueur de pas plus courte et une durée de phase de double support augmentée [30]. De plus, la même étude montrait un lien entre augmentation de la variabilité de la longueur du pas et augmentation de la prévalence des infarctus y compris pour les noyaux gris centraux [31].

L'IRM morphologique soutient également la théorie « vasculaire » pour expliquer les modifications de la marche et certains troubles cognitifs apparaissant avec l'âge. En effet, dans deux publications similaires, le rôle des lésions athérosclérotiques dans le développement de la démence chez les patients MCI a été mis en évidence [14,28].

En considérant les études précédentes, les lésions vasculaires semblent un facteur de risque significatif du développement d'altérations de la marche et/ou d'un déclin cognitif. Le risque de chutes dépend de la topographie des lésions vasculaires présentes dans la région frontale, responsable des fonctions exécutives et de l'organisation de la commande motrice. Une autre étude souligne l'importance des lésions sous-corticales non seulement dans la région frontale mais également dans sept régions spécifiques : frontale, temporale, lobes pariétaux et occipitaux, noyaux gris centraux, capsule interne, capsule externe et le cervelet. L'analyse statistique multivariée montre que seules les lésions vasculaires sous-corticales constituent un facteur de risque prédictif de chutes dans une population âgée déterminée [12].

Selon les auteurs, le mécanisme sous-jacent pourrait être une hypoperfusion sous-corticale interrompant les réflexes à longue boucle des voies motrices de la substance blanche profonde et des fibres motrices issues des aires corticales médianes.

En résumé, il semble que l'IRM morphologique souligne une corrélation entre présence de lésions vasculaires, atrophie de l'hippocampe, changements de la marche et leurs implications dans l'évolution cognitive des sujets âgés MCI.

Tableau 1
Résumé des nouvelles connaissances grâce à l'imagerie.

Premier auteur et références bibliographiques	Observations
Zimmerman et al., Brain Res, 2009	Corrélation entre la longueur du pas et le volume hippocampique ainsi qu'entre la variabilité de la longueur du pas et le métabolisme hippocampique
Onen et al., Neurosci Lett 355, 2004	Lien entre lésions vasculaires dans des zones spécifiques et troubles de la marche chez
Onen et al., Brain Res, 2008	la personne MCI
Onen F., J Neuroradiol, 2005	Rôle de la microangiopathie et circulation du liquide céphalorachidien dans les troubles de la marche chez la personne MCI
Rosano et al., Neuroepidemiology, 2006	Lésion vasculaire de la substance blanche et AVC silencieux sont associés avec une marche ralentie, un pas raccourci et une augmentation de la durée de la phase de double support
Rosano et al., Neuroepidemiology, 2007	Infarctus des noyaux gris centraux sont associés à la variabilité de la longueur du pas
Pugh et Lipsitz, Neurobiol Aging, 2002	L'âge lui-même est associé à la microangiopathie
Henry-Feugeas et al., Clin Interven Aging, 2008	Rôle des lésions cérébrales athérosclérotiques dans le développement de la démence chez la personne MCI
Guerini et al., Arch Phys Med Rehabil, 2008	Hypoperfusion sous-corticale interrompant les réflexes à longue boucle des voies motrices de la substance blanche profonde et des fibres motrices issues des aires corticales médiane

MCI : *mild cognitive impairment*.

2.3.1.2. Imagerie fonctionnelle. Le Tableau 2 résume les principales études.

Les structures impliquées dans le contrôle de la marche peuvent être analysées à l'aide de nouvelles techniques comme la TEMP/SPECT-HMPAO, une tomographie d'émission monophotonique avec technétium, la spectroscopie proche infrarouge NIRS), la stimulation magnétique transcranienne (SMT), l'électroencéphalographie (EEG), l'IRMf et le PET-scan. En fonction des limites techniques, l'IRM fonctionnelle et le PET-scan sont les méthodes d'imagerie les plus souvent utilisées [4] et c'est pour cela que nous avons choisi de parler uniquement de ces deux techniques. De plus, certaines études décrites dans ce paragraphe concernent une population jeune car aucune étude similaire n'est retrouvée au sein d'une population âgée.

L'IRMf n'utilise pas de traceur radioactif, mais détecte les changements de concentration de la désoxyhémoglobine qui reflètent les différences d'activations entre plusieurs conditions. Cette technique différentie les zones corticales et sous-corticales activées par les tâches demandées par l'examinateur. L'IRMf (nécessitant que le patient reste immobile) ne peut étudier l'activation cérébrale liée directement à la véritable action mais elle analyse l'activation secondaire obtenue quand le sujet se représente mentalement cette action. En 2006, Bakker et al. [3] ont montré que le temps nécessaire à la représentation mentale de la marche était comparable à celle d'une marche réelle dans les mêmes conditions (longueur et largeur du pas) chez le sujet jeune. Ils ont conclu à une concordance temporelle forte entre la marche réelle et la marche imaginée et la possibilité d'étudier les mécanismes

Tableau 2
Résumé des nouvelles connaissances grâce à l'imagerie fonctionnelle.

Premier auteur et références bibliographiques	Observations
Bakker et al., J Neural Transm, 2007	IRMf et PET-Scan sont les techniques les plus souvent utilisées dans l'analyse de la marche
Bakker et al., fMRI, Exp Brain Res, 2007	Concordance temporelle forte entre marche réelle et marche imaginée chez le sujet jeune
Personnier et al., Neurosci Lett, 2010	Tendance du sujet âgé à surestimer la durée des mouvements imaginés
Beauchet et al., J Neurol Sci, 2010	Tendance du sujet âgé à sous-estimer la durée des mouvements imaginés
Malouin et al., PETscan, Hum Brain Mapp, 2003	Rôle de l'aire motrice pré-supplémentaire (pré-AMS), de la région jambe du cortex moteur et des noyaux gris centraux dans les mouvements locomoteurs volontaires
Jahn et al., fMRI, Neuroimage, 2004	Station debout : activation des thalamus, putamen gauche, gyrus frontal gauche et vermis Marche : activation de l'AMS, gyrus parahippocampique et fusiforme, aires visuelles occipitales, gyrus frontal inférieur, putamen gauche et vermis
Jahn K., fMRI, Brain Res, 2008	Importance de l'activité dans le gyrus parahippocampique et le gyrus fusiforme quand la marche doit être précise
Deshpande et Patla, Galvanic vestibular stimulations, Brain Res, 2007	Diminution de la capacité d'atténuation de l'information vestibulaire avec l'âge
Zwergal et al., fMRI, Neurobiology of aging, 2012	Inhibition corticale et interaction entre les systèmes sensoriels durant la marche et la phase de support déclinent avec l'âge
La Fouchère, fMRI and PET, Neuroimage, 2010	Rôle des noyaux gris centraux dans l'initiation de la marche
Vidoni et al., fMRI, J Neurol Phys Ther. 2012	Activation diminuée dans les régions motrices accessoires, aire motrice supplémentaires et le cervelet chez les patients aux stades précoces de la maladie d'Alzheimer et augmentation de la coactivation dans le cortex moteur primaire des régions motrices et visuelles chez les sujets atteints de MA

MA : maladie d'Alzheimer ; pré-AMS : aire motrice pré-supplémentaire.

sous-jacents de la marche de cette manière. À notre connaissance, il existe seulement deux études comparant les sujets jeunes et les sujets âgés. Une étude montrait que le sujet âgé surestimait la durée des mouvements imaginés [27]. Et l'autre étude montrait que l'action imaginée était plus rapide que l'action physique elle-même [7]. Lors de la réalisation de ce type d'étude il semble donc nécessaire de comparer le temps nécessaire pour accomplir la tâche et le temps passé à se représenter la tâche. De plus, la population étudiée en imagerie fonctionnelle doit être précisément définie et les conclusions obtenues dans un groupe et selon un protocole bien défini ne peuvent être extrapolées à d'autres groupes ni comparées aux résultats obtenus avec d'autres protocoles.

En 2003, Malouin et al.[21], dans leur étude sur PET scan avec six sujets sains (âge moyen de 59 ans) comparaient l'activation cérébrale durant l'imagerie motrice d'une tâche locomotrice et montraient l'implication de l'aire motrice pré-supplémentaire (pré-AMS), et la région jambes du cortex moteur dans des conditions nécessitant des mouvements locomoteurs (marche, initiation de la marche (IM) et franchissement d'un obstacle). Cette étude montrait également le rôle joué par les noyaux gris centraux dans les mouvements locomoteurs, automatiques par nature.

En 2004, Jahn et al. [17], dans une étude chez 13 sujets jeunes en bonne santé (âge moyen 27,3 ans), comparaient l'activation cérébrale en position debout, durant la marche et la course. La position debout était associée avec une activation au sein des thalamus (plus vers la droite), du putamen gauche, du gyrus frontal gauche et du vermis, de plus il n'existait pas de désactivation des autres régions. La marche, quant à elle, était liée à une activation au sein des aires motrices supplémentaires comme l'AMS, du gyrus parahippocampique et du gyrus fusiforme ainsi qu'au sein des aires visuelles occipitales, du gyrus frontal inférieur, du putamen gauche, du vermis et du lobe antérieur du cervelet.

En 2008, ce même auteur a mené une étude sur 19 jeunes adultes en bonne santé (âge moyen 33,5 ans), en leur demandant de s'imaginer marchant le long d'une ligne incurvée, il a utilisé la même méthode décrite lors de son étude précédente, et il a validé l'importance de l'activation du gyrus parahippocampique et du gyrus fusiforme quand la navigation visuelle est indispensable pour mener à bien un schéma précis de marche. Cependant, des zones de désactivation étaient identifiées dans le gyrus temporal supérieur et le gyrus temporel moyen. Jahn a suggéré que cette désactivation diminuait l'importance des afférences vestibulaires et ainsi évitait un conflit éventuel entre les informations provenant des systèmes visuel et vestibulaire [41]. Cette modulation de l'information du système vestibulaire semble avoir été confirmée dans l'étude de Deshpande et Patla, utilisant une stimulation vestibulaire galvanique au niveau de l'os mastoïde pour souligner l'effet de la perturbation vestibulaire sur la performance locomotrice chez le sujet jeune et sain (de 20 à 35 ans) et la personne âgée en bonne santé (de 65 à 85 ans) [9]. Cette étude soulignait que chez le sujet jeune, l'information vestibulaire pouvait être atténuée pour favoriser l'information visuelle. Cependant, cette capacité d'atténuation diminue avec l'âge et le sujet âgé est plus sensible au déséquilibre quand les informations vestibulaire et visuelle diffèrent. De la même façon, il a été démontré que le mécanisme d'interaction inhibitrice réciproque entre les systèmes sensoriels pendant la marche et la position débout décline chez le sujet âgé [47]. Ces données semblent importantes pour étudier la marche chez la personne âgée atteinte de déficits sensoriels. Enfin, en comparant les résultats de deux techniques d'imagerie fonctionnelle, l'IRMf et le 18F-FDG PET scan, l'auteur a validé le rôle des noyaux gris centraux dans l'IM au sein d'une population de 16 adultes en bonne santé (entre 51 et 73 ans, âge moyen : 61,3 ans $\pm$ 7,8 ans) [18].

Selon Jahn et al., durant la marche, les régions cérébrales ont des rôles spécifiques : le tronc cérébral est impliqué dans le début et la fin de la marche alors que l'aire corticale, les noyaux gris centraux et les thalamus s'activent pendant l'IM et les changements de direction. Les noyaux gris centraux et l'aire corticale sont responsables de la modulation de la marche pour suivre des directions spécifiques ou dans le but d'éviter un obstacle. Le cervelet semble commander le rythme et la vitesse de marche. Certaines aires corticales, comme l'AMS, sont impliquées dans le contrôle cognitif de la marche et leur importance augmente avec l'âge. En effet, alors que l'atténuation de l'information provenant du système vestibulaire est moins efficace avec l'âge, la composante corticale semble plus importante quand la marche tend vers un but précis et durant les changements de direction. Selon des études récentes, l'activité du réseau cortical augmente avec l'âge, expliquant ainsi les difficultés à accomplir plusieurs tâches. Une étude récente confirme que le réseau coordonnant la marche se modifie avec l'âge et lors de certaines pathologies [40]. Dans les stades précoces de la maladie d'Alzheimer (MA), les sujets semblent présenter une diminution de l'activation dans les régions motrices du cortex, l'aire motrice supplémentaire et le cervelet comparé aux sujets sains. De plus, ces mêmes sujets atteints de la MA montrent une augmentation de la coactivation au sein du cortex moteur primaire des aires motrices et visuelles bilatérales.

Les récentes études disponibles se focalisent sur des populations jeunes. Très peu de travaux sont menés sur des populations âgées et surtout des sujets âgés souffrant de troubles cognitifs. Ce manque de données peut être expliqué par différents éléments. D'abord, le nombre restreint de centres disposant de ce type de techniques et la difficulté d'amener une population de personnes âgées à se mobiliser pour de telles études. Un autre facteur limitant étant la difficulté d'interprétation des résultats obtenus au sein de petits groupes et la difficulté d'extrapoler ces résultats à de plus larges populations.

2.3.2. Contribution des paramètres spatiotemporels de la marche

Le Tableau 3 résume les différentes études.

À l'heure actuelle, l'étude précise des paramètres spatio-temporels de la marche nécessite l'utilisation de tapis de marche (comme le GaitRite®) ou de méthodes accélérométriques (Locométrix® ou Dynaport®). De nombreuses équipes utilisent le système GaitRite® (bien que d'autres systèmes

Tableau 3
Contribution des paramètres spatiotemporels de la marche.

Auteur et références bibliographiques	Observations
Bridenbaugh et Kressig, Gerontology, 2011	Méthode instrumentale plus précise que les tests cliniques
Wang et al., J Neural Transm, 2009	La composante attentionnelle de la marche augmente avec l'âge
Verghese et al., J Gerontol A Biol Sci Med Sci, 2009	Vitesse et variabilité de la marche sont deux indicateurs du risque de chutes
Hausdorff et al., Arch Phys Med Rehabil, 2001	Une augmentation de la longueur du pas au cours d'un cycle de pas double le risque de chutes
Maki, J Am Geriatr Soc, 1997	La variabilité de la marche est un indicateur de la peur de tomber et pas un indicateur du risque de chutes
Beauchet et al., JAGS, 2008	La pertinence de l'étude de la vitesse, de la longueur du pas ainsi que la variabilité
Dubost et al., Hum Mov Sci, 2006	de la marche en condition de double tâche en relation avec le risque de chutes
Allali et al., Dement Geriatr Cogn Disord, 2008	L'importance des fonctions exécutives en relation de la variabilité de la marche
Sheridan et al., JAGS, 2003	L'importance de la capacité à diviser l'attention en relation avec la variabilité de la marche
Holtzer et al., Neuropsychology, 2007	Importance des fonctions cognitives en relation avec la variabilité de la vitesse de la marche
Allali et al., Mov Disord, 2010	Profil de marche particulier chez les sujets atteints de MA ou démence fronto-temporale
Verghese et al., J Neurol Neurosurg Psychiatry, 2007	Les changements du profil de marche pourraient aider à prédire l'évolution cognitive des sujets MCI
Bautmans et al., Gait Posture, 2011	Fiabilité entre les résultats obtenus avec le Dynaport® et le GaitRite® au cours de deux parcours de marche de 18 m et en relation avec les chutes et le statut fonctionnel chez la personne âgée
Gillain et al., Ann Phys Rehabil Med, 2009	Un accéléromètre permet de souligner les profils de marche spécifiques selon le statut cognitif chez la personne âgée

existent). Ce tapis de marche existe en plusieurs longueurs et il est équipé de capteurs de pression mesurant les paramètres spatiotemporels de la marche. Les données enregistrées sont la vitesse, la durée du cycle de marche, longueur et largeur de pas et durée de la phase de simple support. Enfin, en calculant le coefficient de variation du temps du cycle de marche et en comparant entre eux chaque cycle de marche, il est possible d'exprimer la « variabilité » de la marche.

Comme récemment rapporté dans une revue sur la pertinence de l'analyse de la marche, ces techniques permettent d'affiner le profil de marche bien plus précisément que les seuls tests cliniques [8]. De plus, au cours des dix dernières années, l'utilisation de ces techniques a montré qu'avec l'âge, la personne doit se concentrer sur l'action de marcher [42]. Certains auteurs ont souligné que la vitesse et la variabilité de la marche sont deux indicateurs du risque de chutes chez la personne âgée (Verghese et al., 2009) ; une variation de 0,017 m de la longueur du pas dans un cycle de pas double ce risque [13]. Quelques années auparavant, Brian Maki à l'aide de capteurs sous chaque pied, rapporte qu'une vitesse réduite, une diminution de la longueur du pas et l'augmentation du temps de phase de support étaient des facteurs indépendants de risque de chutes. Mais, dans la même étude l'auteur souligne qu'une augmentation de la variabilité entre les cycles de marche n'est pas associée au risque de chutes mais uniquement à la peur de tomber [20]. Malheureusement, il est difficile de comparer ces trois études sans disposer des descriptions et caractéristiques exactes des populations étudiées et de leurs antécédents de chutes avant l'évaluation.

En utilisant le GaitRite® plusieurs auteurs ont relevé l'importance de l'étude de la marche en condition de double tâche, consistant à demander au sujet de marcher tout en exécutant une tâche cognitive. Trois paramètres : la vitesse, la longueur du pas et la variabilité de la marche semblent prédire avec acuité le risque de chutes dans cette condition de double tâche [6,10].

D'autres équipes ayant mené des études sur des patients âgés au début d'un développement de troubles cognitifs ou de MA, ont souligné l'influence de la mémoire, des fonctions exécutives [1] mais également de la capacité à diviser l'attention [32] sur la variabilité des paramètres de la marche et notamment la vitesse [15]. Plus récemment, un lien significatif a été validé entre l'augmentation du coefficient de variation du temps de cycle de marche chez le sujet âgé atteint de démence fronto-temporale en comparaison au sujet sain [2]. Dans une étude sur 400 sujets sains d'esprit, les auteurs ont mis en place une évaluation cognitive et instrumentale des paramètres de marche à l'inclusion et à cinq ans. Après cinq ans, les auteurs rapportent une corrélation significative ($p = 0{,}02$) entre l'évolution de la vitesse et de la longueur du pas et le déclin des fonctions cognitives et exécutives, après ajustement pour l'âge, le sexe et le niveau d'éducation [36].

Utilisant un accéléromètre triaxial (Dynaport®), fixé autour de la taille du patient à l'aide d'une ceinture élastique et enregistrant les accélérations dans les axes antéro-postérieur, médio-latéral et rostro-caudal, une équipe a montré la fiabilité et la reproductibilité des paramètres de marche obtenus lors de deux aller-retour sur un parcours de 18 m de long dans corridor rectiligne. Enfin, les performances obtenues par les volontaires âgés participant ont montré une corrélation avec les chutes et le statut fonctionnel [5] des sujets. Dans ce contexte, notre équipe a souhaité analyser les paramètres de marche à l'aide d'un accéléromètre triaxial portable le Locométrix® au sein d'une population de sujets âgés sains, de sujets MCI et de sujets atteints de MA. Les sujets des trois groupes devaient marcher au milieu d'un corridor droit et sans obstacle de 40 m de long, lors d'une tâche simple (marche) et d'une tâche double (marcher tout en comptant à l'envers). Dans les deux tests, le

Locométrix® était capable de différentier les trois populations à partir de leur profil de marche lié au statut cognitif [11].

L'étude instrumentalisée de la marche semble utile car elle permet d'identifier une ou plusieurs variables spatiotemporelles augmentant le risque de chutes et ainsi aider à discriminer les patients avec troubles cognitifs en fonction de leur profil de marche et ainsi identifier les personnes à risque de développer une démence.

Ces méthodes instrumentales d'étude des paramètres de la marche montrent que l'âge n'est pas le seul élément pouvant expliquer la variabilité de la marche. En effet, les troubles cognitifs, antécédents de chutes ainsi que vitesse et conditions de marche demandées (simple ou double tâche) sont des facteurs essentiels ayant un impact sur la marche de la personne âgée et peuvent aider à estimer le risque de chutes et prédire l'évolution cognitive. Ainsi, en prenant en compte les paramètres de la marche il convient de préciser les conditions nécessaires à l'évaluation de la marche, les caractéristiques de la population (non seulement l'âge mais également antécédents de chutes, traitements, comorbidités, déficits sensoriels, troubles neurologiques…), vitesse de marche demandée, technique utilisé. Dans ce contexte, les résultats d'une étude réalisée à l'aide d'une technique spécifique et se rapportant à une population déterminée ne peuvent s'extrapoler à d'autres situations. Dans ce contexte, les équipes étudiant la marche rencontrent des difficultés dans une population gériatrique, tout d'abord avec la conception de l'étude et ensuite pour trouver un équilibre (ou pas) entre la nécessité de souligner des résultats à forte puissance statistique et l'intérêt de montrer cet impact dans une population « plus réaliste ».

2.3.3. Initiation de la marche

Plus récemment, plusieurs équipes ont tenté, à l'aide de différentes techniques, d'étudier le moment spécifique d'IM. En effet, ce moment semble particulièrement dangereux chez la personne âgée ayant un risque élevé de chutes. L'IM est une transition volontaire entre un état statique stable (phase de support) et une posture continuellement instable durant la marche qui nécessite des ajustements posturaux anticipateurs pour glisser le centre de masse vers le côté d'appui afin de pouvoir lever la jambe [33]. Comme souligné à l'aide de la TEMP ou PET-scan [21] et la stimulation magnétique transcranienne répétitive (rTMS), l'IM nécessite l'activation de l'AMS mais également des noyaux gris centraux qui semblent également impliqués, comme rapporté chez le sujet parkinsonien [16,44].

L'IM est souvent analysée à l'aide d'une plateforme de force. Au cours des dernières années, plusieurs équipes ont étudié l'IM dans une population générale âgée [22], chez la personne âgée ayant peur de tomber [34], chez la personne âgée avec antécédents de chutes [23] ainsi que chez des patients avec maladie de Parkinson [29] et MA [45]. L'IM montre des modifications spécifiques liées à la population étudiée. Des études complémentaires seraient pertinentes pour confirmer et affiner ces observations, bien que l'IM soit l'objet d'un grand intérêt scientifique.

2.4. Discussion et conclusion

Plusieurs études montrent la pertinence de l'évaluation de la marche chez la personne âgée. Les nouvelles techniques d'imagerie et d'analyses instrumentales permettent d'objectiver le lien entre l'anatomie des structures centrales, leur métabolisme et les paramètres spatiotemporels de la marche. La marche semble être un indicateur « de vieillissement réussi » pouvant prédire les changements cognitifs et fonctionnels. Malheureusement et malgré les informations essentielles recueillies, ces techniques restent confinées au niveau expérimental et sont peu utilisées en pratique clinique. La vitesse, la longueur du pas et la variabilité de la marche semblent des indicateurs importants de la qualité de marche. De plus, l'étude de la marche en condition de double tâche pourrait fournir des éléments sur le risque de chutes et la réserve cognitive attribuée pour l'organisation de la marche. Ainsi, il semble pertinent de procéder à une évaluation systématique de la marche en simple et double tâche ainsi qu'un suivi chez la personne âgée. Enfin, l'IM représente un élément crucial pour déterminer la sécurité de la personne âgée, c'est pourquoi l'étude de ce moment pourrait identifier les mécanismes sous-jacents à l'équilibre dynamique et la marche afin d'essayer de les améliorer.

Selon nous, les tests cliniques de la marche restent essentiels comme la vitesse de marche sur 10 m le « Timed Up and Go test ». Le Short Physical Performance Battery Test qui comprend le test « Sit-to-stand », les test de Berg ou Tinetti, reste le meilleur moyen d'étudier de façon rapide et utile l'équilibre du sujet âgé. En ce qui concerne la double tâche, le « stop walking when talking » test semble un prédicteur important du risque de chutes chez le sujet âgé [19]. En effet, la comparaison de la marche en simple et double tâche avec une tâche de parole, est un outil prédictif du risque de chute avec une sensibilité assez basse de 40 % mais une spécificité de 96 % [35]. Pour le choix d'une méthode d'évaluation de la marche, l'important est de sélectionner un test facile à reproduire dans les mêmes conditions et bien maitrisé par l'examinateur. Enfin, un test de force de préhension ou vigorimètre de Martin, pourraient être de bons indicateurs de masse musculaire pour mesurer le statut fonctionnel présent et à venir du sujet âgé.

À l'avenir, l'utilisation de ces nouvelles techniques et l'amélioration des indicateurs des systèmes centraux et périphériques pourrait améliorer l'identification du sujet à risque de chutes afin de leur proposer un suivi multidisciplinaire. Ces travaux permettront de mettre en place des protocoles de rééducation spécifique, de la marche et de la fonction cognitive, adaptés au sujet âgé fragile.

Déclaration d'intérêts

Les auteurs déclarent ne pas avoir de conflits d'intérêts en relation avec cet article.

References

[1] Allali G, Assal F, Kressig RW, Dubost V, Herrmann FR, Beauchet O. Impact of impaired executive function on gait stability. Dement Geriatr Cogn Disord 2008;8626:364–9.

[2] Allali G, Dubois B, Assal F, Lallart E, de Souza LC, Bertoux M, et al. Frontotemporal dementia: pathology of gait? Mov Disord 2010;25:731–7.
[3] Bakker M, de Lange FP, Stevens JA, Toni I, Bloem BR. Motor imagery of gait: a quantitative approach. Exp Brain Res 2007;179:497–504.
[4] Bakker M, Verstappen CC, Bloem BR, Toni I. Recent advances in functional neuroimaging of gait. J Neural Transm 2007;114:1323–31.
[5] Bautmans I, Jansen B, Van Keymolen B, Mets T. Reliability and clinical correlates of 3D-accelerometry based gait analysis outcomes according to age and fall-risk. Gait Posture 2011;33:366–72.
[6] Beauchet O, Annweiler C, Allali G, Berrut G, Herrmann FR, Dubost V. Recurrent falls and dual task-related decrease in walking speed: is there a relationship? JAGS 2008;56:1265–9.
[7] Beauchet O, Annweiler C, Assal F, Bridenbaugh S, Hermann FR, Kressig RW, et al. Imagined Timed Up & Go test: a new tool to assess higher-level gait and balance disorders in older adults? J Neurol Sci 2010;294:102–6.
[8] Bridenbaugh SA, Kressig RW. Laboratory review: the role of gait analysis in seniors'mobility and fall prevention. Gerontology 2011;57:256–64.
[9] Deshpande N, Patla AE. Visual-vestibular interaction during goal directed locomotion: effects of aging and blurring vision. Exp Brain Res 2007;176:43–53.
[10] Dubost V, Kressig RW, Gonthier R, Hermann FR, Aminian K, Najafi B, et al. Relationship between dual-task related changes in stride velocity and stride time variability in healthy older adults. Hum Mov Sci 2006;25:372–82.
[11] Gillain S, Warzee E, Lekeu F, Wojtasik V, Maquet D, Croisier JL, et al. The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects on a comparison with other clinical tests used in single and dual-task conditions. Ann Phys Rehabil Med 2009;52:453–74.
[12] Guerini F, Frisoni G, Marrè A, Turco R, Bellelli G, Trabucchi M. Subcortical vascular lesions predict falls at 12 months in elderly patients discharged from rehabilitation ward. Arch Phys Med Rehabil 2008;89:1522–7.
[13] Hausdorff JM, Rios DA, Edelberg HK. Gait variability and falls risk in community-living older adults: a 1-year prospective study. Arch Phys Med Rehabil 2001;82:1050–6.
[14] Henry-Feugeas MC, Onen F, Schouman-Claeys E. Classifying late-onset dementia with MRI: is arteriosclerotic brain degeneration the most common cause of Alzheimer's syndrome? Clin Interv Aging 2008;3:189–99.
[15] Holtzer R, Friedman R, Lipton RB, Katz M, Xue X, Verghese J. The relationship between specific cognitive functions and falls in aging. Neuropsychology 2007;21:540–8.
[16] Jacobs JV, Lou JS, Kraakevik JA, Horak FB. The supplementary motor area contributes to the timing of the anticipatory postural adjustment during step initiation in participants with and without Parkinson's disease. Neuroscience 2009;164:877–85.
[17] Jahn K, Deutschländer A, Stephan T, Strupp M, Wiesmann M, Brandt T. Brain activation patterns during imaging stance et locomotion in functional magnetic resonance imaging. Neuroimage 2004;22:1722–31.
[18] La Fougère C, Zwergal A, Rominger A, Förster S, Fesl G, Dieterich M, et al. Real versus imagined locomotion: a [18F]-FDG PET-fMRI comparaison. Neuroimage 2010;50:1589–98.
[19] Lundin-Olsson L, Nyberg L, Gustafson Y. "Stop walking when talking" as a predictor of falls in elderly people. Lancet 1997;349:617.
[20] Maki BE. Gait changes in older adults: predictors of falls or indicators of fear. J Am Geriatr Soc 1997;45:313–20.
[21] Malouin F, Richards CL, Jackson PL, Dumas F, Doyon J. Brain activations during motor imagery of locomotors-related tasks: a PET study. Hum Brain Mapp 2003;19:47–62.
[22] Martin K, Blizzard L, Garry M, Thomson R, McGinley J, Srikanth V. Gait initiation in older people – Time to first lateral movement may be the measure of choice. Gait Posture 2011;34:374–8.
[23] Mbourou GA, Lajoie Y, Teasdale N. Step length variability at gait initiation in elderly fallers and non-fallers, and young adults. Gerontology 2003;49:21–6.
[24] Onen F, Henry-Feugeas MC, Baron G, De Marco G, Godon-Hardy S, Peretti I, et al. Leukoaraiosis and mobility decline: a high resolution magnetic resonance imaging study in older people with mild cognitive impairment. Neurosci Lett 2004;355:185–8.
[25] Onen F, Feugeas MC, De Marco G, Baron G, Ravaud P, Legrain S.. et al. Cerebrospinal fluid MR dynamics and risk of falls in the elderly. J Neuroradiol 2005;32:3–9.
[26] Onen F, Henry-Feugeas MC, Roy C, Baron G, Ravaud P. Mobility decline of unknow origin in mild cognitive impairment: a MRI-based clinical study of the pathogenesis. Brain Res 2008;1222:79–85.
[27] Personnier P, Kubicki A, Laroche D, Papaxanthis C. Temporal features of imagined locomotion in normal aging. Neurosci Lett 2010;476:146–9.
[28] Pugh KG, Lipsitz LA. The microvascular frontal subcortical syndromes of aging. Neurobiol Aging 2002;23:421–31.
[29] Roemmich RT, Nocera JR, Vallabhajosula S, Amano S, Naugle KM, Stegemöller EL, et al. Spatiotemporal variability during gait initiation in Parkinson's disease. Gait Posture 2012;36:340–3.
[30] Rosano C, Brach J, Longstreth Jr WT, Newman AB. Quantitative measures of gait characteristics indicate prevalence of underlying subclinical structural brain abnormalities in high functioning older adults. Neuroepidemiology 2006;26:52–60.
[31] Rosano C, Brach J, Studenski S, Longstreth Jr WT, Newman AB. Gait variability is associated with subclinical brain vascular abnormalities in high-functioning older adults. Neuroepidemiology 2007;29:193–200.
[32] Sheridan PL, Solomont J, Kowall N, Hausdorff JM. Influence of executive function on locomotor function: divided attention increases gait variability in Alzheimer's disease. JAGS 2003;51:1633–7.
[33] Tateuchi H, Ichihashi N, Shinya M, Oda S. Anticipatory postural adjustments during lateral step motion in patient with hip osteoarthritis. J Appl Biomech 2011;27:32–9.
[34] Uemura K, Yamada M, Nagai K, Tanaka B, Mori S, Ichihashi N. Fear of falling is associated with prolonged anticipatory postural adjustment during gait initiation under dual-task conditions in older adults. Gait Posture 2012;35:282–6.
[35] Verghese J, Buschke H, Viola L, Katz M, Hall C, Kuslansky G, et al. Validity of divided attention task in predicting falls in older individiuals: a preliminary study. JAGS 2002;50:1572–6.
[36] Verghese J, Wang C, Lipton RB, Holtzer R, Xue X. Quantitative gait dysfunction and risk of cognitive decline and dementia. J Neurol Neurosurg Psychiatry 2007;78:929–35.
[37] Verghese J, Robbins M, Holtzer R, Zimmerman M, Wang C, Xue X, et al. Gait dysfunction in mild cognitive impairment syndromes. JAGS 2008;56:1244–51.
[38] Verghese J, Holtzer R, Lipton RB, Wang C. Quantitative gait markers and incident fall risk in older adults. J Gerontol A Biol Sci Med Sci 2009;8:896–901.
[39] Verghese J, Ambrose AF, Lipton RB, Wang C. Neurological gait abnormalities and risk of falls in older adults. J Neurol 2010;257:392–8.
[40] Vidoni ED, Thomas GP, Honea RA, Loskutova N, Burns JM. Evidence of altered corticomotor system connectivity in early-stage alzheimer's disease. J Neurol Phys Ther 2012;36:8–16.
[41] Wagner J, Stephan T, Kalla R, Brückmann H, Strupp M, Brandt T, et al. Mind the bend: cerebral activation associated with mental imagery of walking along a curved pad. Exp Brain Res 2008;191:247–55.
[42] Wang J, Wai Y, Weng Y, Ng K, Huang YZ, Ying L, et al. Functional MRI. in the assessment of cortical activation during gait related imaginary tasks. J Neural Transm 2009;116:1087–92.
[43] Warzee E, Elbouz L, Seidel L, Petermans J. Evaluation of gait by 115 inpatients in geriatric department of CHU Liège. Ann Gerontol 2010;3:1–9.
[44] Wichmann T, Delong MR. Deep-brain stimulation for basal ganglia disorders. Basal Ganglia 2011;1:65–77.
[45] Wittwer JE, Andrews PT, Webster KE, Menz HB. Timing variability during gait initiation is increased in people with Alzheimer's disease compared to controls. Dement Geriatr Cogn Disord 2008;26:215–25.
[46] Zimmerman ME, Lipton RB, Pan JW, Hetherington HP, Verghese J. MRI- and MRS- derived hippocampal correlates of quantitative locomotor function in older adults. Brain Res 2009;1291:73–81.
[47] Zwergal A, Linn J, Xiong G, Brandt T, Strupp M, Jahn K. Aging of human supraspinal locomotor and postural control in fMRI. Neurobiol Aging 2012;33:1073–84.

Disponible en ligne sur ScienceDirect www.sciencedirect.com
Elsevier Masson France EM|consulte www.em-consulte.com
ELSEVIER MASSON
Annals of Physical and Rehabilitation Medicine 52 (2009) 453–474
ANNALS OF PHYSICAL AND REHABILITATION MEDICINE

Original article / Article original

The value of instrumental gait analysis in elderly healthy, MCI or Alzheimer's disease subjects and a comparison with other clinical tests used in single and dual-task conditions

Apports de l'analyse instrumentale de la marche de sujets âgés sains, MCI et alzheimer et comparaison avec d'autres tests cliniques utilisés en simple et double tâche

S. Gillain [a,*], E. Warzee [a], F. Lekeu [a,c], V. Wojtasik [a,c], D. Maquet [b], J.-L. Croisier [b], E. Salmon [c], J. Petermans [a]

[a] *Département de gériatrie, CHU de Liège, Liège, Belgium*
[b] *Département des sciences de la motricité, CHU de Liège, Liège, Belgium*
[c] *Département de neurologie, CHU de Liège, Liège, Belgium*

Received 16 March 2007; accepted 3 October 2008

Abstract

In the elderly, gait disorders and cognitive frailty may influence each other and increase the risk of falling. The aim of the present study was to determine gait parameters in elderly people with different cognitive profiles (controls, individuals with mild cognitive impairment [MCI] and Alzheimer's disease [AD] patients) with the Locometrix® three-axis accelerometer and establish whether or not this tool is more useful than conventional clinical tests (the timed "get up and go" test, the pull test and the single-leg balance test). Study subjects were all over 65, living at home and free of known gait impairments. A neuropsychological battery was applied to 14 control subjects, 14 MCI subjects and six AD patients. A motor evaluation (in single- and dual-task paradigms) was performed with three conventional clinical tests and the Locometrix® (standardized gait). Our results showed that in a single-task paradigm, the Locometrix® was more accurate than validated, conventional tests and generated a characteristic gait profile for each of the three cognitive profiles. In a dual-task paradigm, the gait of MCI subjects more closely resembled that of AD patients than that of control subjects. We conclude that the Locometrix® is a high-performance tool for defining gait profiles, which correspond to given cognitive profiles. The use of a dual-task paradigm is a good way to screen for gait abnormalities in MCI.

Keywords: Alzheimer's disease; Mild cognitive impairment; Gait; Locometrix®; Test

Résumé

Avec l'avancée en âge, modifications des paramètres de marche et fragilisation cognitive semblent s'influencer mutuellement majorant dangereusement le risque de chute.
Objectifs. – Le but de ce travail est d'étudier les paramètres de marche de sujets âgés de profils cognitifs différents (sains, *mild cognitive impairment* [MCI] et maladie d'Alzheimer [MA]) grâce à un accéléromètre triaxial, le Locométrix®, et de déterminer si cet instrument est plus pertinent que certains tests cliniques classiques (*timed up and go*, *pull test* et *one-leg balance*).
Patients et méthode. – Nous avons étudié des sujets volontaires de plus de 65 ans, vivant au domicile et ne présentant pas de plainte sur la qualité de leur marche. Une évaluation neuropsychologique a permis de distinguer 14 sujets sains, 14 sujets souffrant de MCI et six sujets présentant une MA. L'évaluation motrice, réalisée en tâche simple et en tâche double, comprend les trois tests cliniques précités et un parcours de marche standardisé et analysé par le Locométrix®.

* Corresponding author. Service de gériatrie, CHU Notre-Dame des Bruyères, rue de Gaillarmont, 4032 Chênée, Belgium.
E-mail address: s.gillain@gmail.com (S. Gillain).

1877-0657/$ – see front matter © 2009 Published by Elsevier Masson SAS.
doi:10.1016/j.rehab.2008.10.004

Résultats. – Utilisé en simple tâche, le Locométrix® permet, contrairement aux tests cliniques validés, de distinguer pour chaque profil cognitif, un profil de marche correspondant soit une diminution de la fréquence des pas propre aux MCI et une diminution de la vitesse de marche, de la longueur des pas et de la régularité de la marche propres aux MA. En double tâche, le Locométrix® révèle que le profil de marche des sujets MCI présente plus de similitudes avec le profil des sujets MA qu'avec celui des sujets sains.
Conclusion. – Le Locométrix® est un outil performant qui, utilisé en simple et double tâche, permet de mieux définir le profil de marche des sujets avec troubles cognitifs y compris les sujets MCI. Dans ce context, le Locométrix® apparaît être un outil plus performant que les tests cliniques utilisés jusqu'alors.

Mots clés : Alzheimer ; MCI ; Marche ; Locométrix® ; Test

1. English version

1.1. Introduction

In the western world, population ageing and the current demographic forecasts mean that detection and treatment of the main geriatric syndromes are becoming increasingly important. Falls and functional and cognitive decline are decisive factors in the fragility of the elderly.

With age, gait parameters change. The main modifications include a reduction in gait speed (due to a reduction in stride length) [12], stride irregularity (due to an increase in lateral deviations), a shortening of the single-support phase and a lengthening of double-support phase [27,39,47,54]. Some authors have suggested that these changes are intended to make the elderly person's gait safer and more balanced [40]. However, the prevalence of falls increases with age (from 33% at the age of 65 to about 50% at 85 years) [55] and the rate of recurrence within the year is high, since one in two subjects fall again [56]. Hence, age per se does not appear to be the only explanatory factor for the appearance of falls in a given subject. In fact, the presence of cognitive impairment constitutes a risk factor for falling [20] and is independent of other comorbid factors. When compared with an elderly population free of cognitive disorders, subjects suffering from Alzheimer's disease (AD) present a lower gait speed [61], a lower stride length and greater cycle-to-cycle gait variability [38].

The motor components of gait also change with age; automatic (involuntary, subcortical) components weaken, with a corresponding increase in voluntary (cortical, ''attentional'') components [7–65]. Hence, many studies have shown that the performance of a cognitive task while walking perturbs not only postural balance [29,30,35] but also the gait parameters related to the risk of falling [11,16]; the additional cortical task competes with the cortical component of gait and thus favours the appearance of underlying gait disorders. AD patients present impairments in executive processes and (depending on the disease stage) working memory, making it difficult or even impossible to manage two actions simultaneously [6]. In these subjects, the performance of gait alone monopolizes the available executive resources. In a dual-task paradigm (TS), the AD patients' executive capacities are soon overloaded [13,49] and thus gait disorders appear rapidly during cognitive decline. Moreover, many authors [9,63,61,1] have recently shown that certain gait profile changes precede the diagnosis of cognitive decline per se. These observations prompted us to wonder whether early screening for gait disorders might be useful not only for preventing falls and functional decline but also for detecting cognitive deficits at an early stage. To this end, we decided to study the relationship between gait profiles and cognitive profiles in different populations of elderly people: AD patients and subjects presenting isolated, mild cognitive impairment (MCI).

1.2. Study objectives

The aim of this work was to study gait parameters in three elderly populations with different cognitive profiles: healthy control subjects (free of memory impairments), MCI subjects and AD patients.

To achieve this, we used the Locometrix® system – a tri-axial accelerometer that enables the evaluation of body movements in all three spatial planes. In order to validate this tool in the present application, each subject also performed three standardized gait tests (the single-leg balance test, the pull test and the timed ''get up and go'' test).

We thus sought to determine:

- whether or not the Locometrix® is a pertinent tool for studying gait in elderly subjects;
- the technique's efficacy compared with conventional clinical tests of balance and gait.

Moreover, we also hypothesized that the tool would enable better gait profile definition in the three populations.

1.3. Population, materials and methods

1.3.1. Population

The healthy controls were recruited from among non-institutionalized volunteers aged over 65 who had not experienced a fall or hospitalization in the preceding six months, were able to walk without assistance and had made no complaints concerning the quality of their gait or balance. The medically related exclusion criteria were as follows:

- vascular stroke with motor or sensory sequelae;
- Parkinson's disease;

- non-compensated diabetes, non-compensated arterial hyper- or hypotension;
- any cardiac or respiratory disease which could cause gait-limiting weakness or dyspnea;
- a hip or knee prosthesis;
- a fracture of the leg or ankle which would have impaired gait;
- arthritis or another invalidating bone/joint disease.

The use of benzodiazepine or an antidepressant or small doses of neuroleptics (lacking motor repercussions) was tolerated. Patients needing glasses and/or a hearing aid were eligible but the subjects had to be completely satisfied with the performance of these sensory aids. All subjects were informed of the study objectives and procedures and had given their informed consent in writing.

The AD and MCI patients were recruited from among those attending Liège University Hospital's Memory Centre (headed by Professor E. Salmon). Memory disorders were diagnosed according to standard medical imaging and neuropsychological evaluation methods. The diagnostic criteria applied in our centre are as follows: MCI patients present a confirmed, isolated cognitive disorder but which has not had a major impact on their activities of daily living (ADL). The subjects had to have undergone the neurological, neuropsychological and neuroimaging evaluations required for diagnosis, with a clinical dementia rating (CDR) of under 0.5 [36]. A distinction was drawn between anamnestic MCI subjects (i.e., presenting a decrease in memory performance of 1.5 standard deviations at most, when compared with the mean value for a population of the same age and social level) and subjects with ''multiple'' MCI (i.e., those also presenting an impairment in another nonmemory-related cognitive function) [41,64]. Other exclusion criteria included mental retardation, less than four regular years of education, cranial trauma, epilepsy, cancer, depression, substance abuse or any other major organic disease. On inclusion, none of the patients was taking any medication likely to influence their cognitive performance. Their score in the Mini Mental State Examination (MMSE) [18] had to be 24 or more.

Demented patients had been diagnosed as having probable AD according to the criteria defined by the National Institute of Neurological and Communicative Disorders and Stroke/Alzheimer Disease and Related Disorders Association (NINCDS-ADRDA) [34]. The exclusion criteria were the same as those applied to the MCI subjects, with the exception of the CDR (which had to be 1 or more). Mild dementia is defined as an MMSE score of 20 or more.

Healthy control subjects were all over the age of 65 and none presented a known memory or cognitive disorder during the recruitment screening session. The exclusion criteria were the same as those given above, with a CDR of 0.

1.3.2. Medical evaluation

The medical evaluation included an interview (to establish the subject's full personal medical history) and a comprehensive clinical examination (height, weight and a cardiorespiratory, abdominal and above all neurological examination), the goal of which was to check for the absence of exclusion criteria. For the AD patients, family members and/or carers were questioned on living conditions, falls and medications; if required, we performed a telephone interview with the patient's family doctor or consulting physician. In addition, each subject was scored on the ADL scale [26], the Instrumental Activity of Daily Living (IADL) scale [28], the Mini Nutritional Assessment (MNA) [46], the 15-item Geriatric Depression Scale (GDS) [50] (in order to exclude any underlying depressive pathologies), a pain visual analogue scale (VAS) [17] and, lastly, the Score Hospitalier d'évaluation du risque de perte d'autonomie (SHERPA) hospital score for evaluating the overall fragility profile and risk of loss of personal independence [14]. The goal was to ensure that the gait test results and the neuropsychological assessment would not be influenced by any organic, affective or functional factors.

1.3.3. The neuropsychological assessment

In order to check that none of the healthy volunteers presented any memory impairments, all underwent an MMSE and were scored on the Mattis scale. In order to ensure similar evaluation conditions for each group, the MCI and AD subjects also underwent these tests. Other tests were also performed in order to detect potential executive or praxis disorders likely to interfere with the gait tests.

The comprehensive evaluation performed by each subject thus included:

- the MMSE [18] for orientation, learning, working memory (via mental arithmetic and word spelling), object naming, understanding simple instructions and copying a drawing. The 30-point score has a cognitive impairment cut-off at 24 (26 for subjects having received higher education);
- the Mattis scale [31] is another set of global cognitive evaluations which complements the MMSE. It investigates frontal and subcorticofrontal functions more broadly and can thus detect types of dementia other than AD. It explores attention, memory, verbal and motor initiation, conceptual abilities and visuoconstructive praxis. The cognitive impairment cut-off is 123 out of 144, which can be modulated depending on the subject's education level;
- a French version of the Grober and Buschke 16-item free recall/cued recall test [57], which examines episodic memory. The subject has to memorize four sheets of four words belonging to 16 different semantic categories. First, the examiner prompts semantic encoding of the words by asking the subject to identify the word corresponding to the stated semantic category (i.e., ''Which one is the flower?''→ ''The daffodil''). When the four words have been identified, the sheet is removed and the subject's task is to immediately recall these four words. This procedure is repeated for the other three sheets. After the 16 words have been learnt, the subjects perform a 20-second interference task (counting backwards). Next, the subject undergoes an initial free recall phase in which he/she must remember as many of the 16 words as possible. For the words which were not been remembered freely, the examiner provides a cue (the name of the category to which the

word belongs). Three free recall and then cued recall trials are performed, followed by a recognition memory task for the words studied. After a 20-minute interference task, a further free recall/cued recall task is set;
- Rey's complex figure test [45] in order to evaluate visuoconstructive and visuospatial organisation abilities. Scoring is both qualitative and quantitative (out of 36);
- the computerized alertness and divided attention sub-tests from the Test for Attentional Performance (TAP) battery [66] during the alertness test, the subject has to press a response key as rapidly as possible when a cross appears in the centre of a screen. This test records the subject's reaction time (in milliseconds) and is performed under two different conditions (in the presence and absence of a warning sound). During the divided attention test, the subject must share his attentional resources between visual stimuli and auditory stimuli. The reaction time (in milliseconds) and the number of correct answers and omissions are recorded. The reaction times are also recorded in two single-task conditions: the appearance of a visual stimulus only and the appearance of auditory stimuli only.

1.3.4. Motor evaluation

The gait tests performed were the pull test, the single-leg balance test, the timed "get up and go" test and a 30-metre stable gait test during which parameters were recorded using the Locometrix®.

The pull test is a test of static balance [37]. The subject stands with his/her feet together. The examiner (standing behind the subject) puts his/her hands on the subject's shoulders and pulls. The subject's body axis is thus displaced backwards. The examiner scores the movements required to maintain balance and reestablish the axis' initial position (toe retropulsion, toe elevation, forward arm movements and any backward steps). We chose this test because it is quick to perform and does not require any specific equipment [3].

The one-leg balance test is also a test of static balance. The subject stands on the leg of his/her choice (with his/her arms by his/her side) and fixes an eye-level visual marker for as long as possible. We chose this test because it is easy to administer and is highly predictive of a low functional level and the risk of falling [59].

The timed "get up and go" test evaluates gait and dynamic balance. The subject sits in a chair with armrests. He/she must stand up (without using the armrests), walk in a straight line for 3 m, turn around, walk back to the chair in a straight line and then sit down.

The studied parameters are:

- the time needed to cover the distance ("normal" if 10 s or less, "fragility" if between 11 and 20 s and a significant risk of falling if over 20 s) [43];
- a qualitative evaluation of gait borrowed from the original "get up and go" test [32].

According to this scoring system, the score of 0 corresponds to the highest possible gait quality and a score of −5 corresponds to the lowest possible quality. We chose this test because it is easy to administer, does not require any particular material or training and appears to be closely correlated with the subject's overall motor status [44]. Furthermore, it has excellent specificity and sensitivity for identifying the risk of falling in the elderly [52] and has excellent interexaminer reproducibility [48,53].

The Locometrix® system was developed in France by the National Agricultural Research Institute (INRA) [24], in collaboration with Dr Bernard Auvinet from the Rheumatology and Physical Medicine Department at Laval Hospital. It features a system comprising two perpendicularly arranged accelerometers placed near the subject's centre of gravity. In practice, the sensor is placed behind the subject's back (at the L3 level) using an elastic, abdominal belt. The sensors thus have craniocaudal and mediolateral measurement axes. A portable data logger records the sensor's accelerations at a frequency of 50 Hz and the data are then transferred to a computer for statistical spreadsheet analysis [4]. During the test, the subject walks up and back along a straight 40-m track (i.e., 80 m in all) at a freely chosen pace and cadence and in a well-lit, straight corridor with flat, level ground and which is free of obstacles or visual/auditory distractions. Two timing lines are located 5 m after the starting line and 5 m before the 40-m line, respectively. By passing in front of these photoelectric cells, the subject automatically starts/stops the chronometer. The gait speed thus calculated (30 m divided by the completion time) is thus a "stable" gait speed that excludes the effects of gait initiation and termination. During the test, the evaluator walks at a standardized distance behind the test subject (usually 3 m) in order to monitor the countdown spoken aloud during the dual-TS and without influencing the subject's gait. Analysis of the gait parameters reported by the accelerometer is performed during a stable gait period of 20.48 s [15]. This period of stable gait is selected by the software after detection of the acceleration curves in the various axes and identification of the period of gait during which these curves vary least (i.e., changes in "stable" curves). The gait variables analyzed are as follows [4]:

- gait speed measured using a timing line and expressed in metres per second;
- the stride frequency or number of cycles per second (Hz) is calculated from the craniocaudal acceleration following application of a Fourier transform;
- stride length is deduced from the equation (speed = frequency × stride length) and is expressed in metres;
- stride regularity measures the similarity (in terms of duration and amplitude) of the shape of craniocaudal acceleration curves from one step to another;
- stride symmetry: the similarity (in terms of duration and amplitude) of the shape of craniocaudal acceleration curves when comparing right and left strides, specifically.

According to Dr Auvinet, the gait speed and stride length parameters are gender-dependent and are greater in men (even when normalized for height). Symmetry and regularity may be

age- and gender-independent. In a study comparing elderly fallers and non-fallers, Dr Auvinet observed that the Locometrix® variables that most discriminated between the two populations were (in decreasing order of importance) gait speed, stride regularity, stride frequency and the symmetry of left steps versus right steps. Analysis of gait regularity is particularly pertinent for early detection of gait degradation [5].

A preliminary study performed on a sample of 265 control subjects (based on six 40-m trials and measurement of the intraobserver reproducibility by comparing the coefficients of variation for the different trials) established that only the second trial should be studied. Hence, in order to maintain similar walking conditions for all tasks, all trials were performed three times and only the results of the second trial were processed. Moreover, in order to make the gait tests as sensitive as possible, we asked the subjects to perform them in a single motor-TS (i.e., the gait test alone) and in a dual-TS (i.e., addition of a cognitive task, to be performed during the gait test). The choice of the cognitive task influences the change in gait parameters [10]. In fact, verbal fluency calls on semantic memory [21] (which is relatively unimpaired at the onset of AD), whereas the countdown task calls on working memory [22] (which is impaired early in the course of the disease). According to Beauchet, the countdown is the additional cognitive task that mostly perturbs gait parameters in a dual-TS [10,11]. This is why we chose a countdown from 50 as the cognitive task.

During this dual task, the patient did not receive any instructions concerning the priority to be given to one task or the other. Furthermore, the order of the pull test, single-leg balance test and the timed "get up and go" test were randomized in both the single- and dual-TS, so that the sequence of tasks would differ as much as possible from one subject to another (in order to limit any possible effects of learning or tiredness on the results).

Lastly, the time required to complete the test, the number of steps and the number of countdown errors were independently noted by two operators. Moreover, all the tests were filmed in order to review the trials in slow motion and better observe lateral deviations and stops. The number of countdown errors and the number of stops performed during the motor task were noted in order to determine the task which had been spontaneously prioritized by the subject.

Over half a day, each participant underwent a medical and functional evaluation, a neuropsychological screen and a motor test.

We were thus able to evaluate three groups of subjects with different cognitive profiles: 14 healthy controls, 14 MCI subjects and six AD patients. The low sample reflects the size of our Memory Centre and the severity of our recruitment criteria, rather than an intentional restriction of the number of subjects to be observed. At this stage, it is important to note that this was a preliminary clinical study designed to evaluate:

- the feasibility of applying the Locometrix® in our centre and to our population;
- different aspects of the gait study protocol in elderly patients presenting confirmed or nascent cognitive disorders.

Our results are presented as means and standard deviations (median) for continuous variables and as percentage frequencies for categorical variables. Intergroup comparisons were performed by using an analysis of variance (ANOVA) or a Kruskal-Wallis test, depending on whether the continuous data were normally distributed or not. For categorical variables, the comparison was performed by using chi-square tests. The significance threshold was set to an alpha risk of 5% (i.e., $p < 0.05$). All calculations were performed with SAS software (version 9.1, SAS Institute, Cary, NC, USA).

1.4. Results

1.4.1. Characteristics of the study population

Tables 1 and 2 summarize the results of the medical evaluation performed on all subjects in the three populations.

The IADL scale scores were calculated by considering the sum of the results obtained in the items applicable to each subject, divided by the sum of the maximum possible scores in the applicable items. Hence, the lower the value, the greater the degree of personal independence. Likewise, the GDS score is the result of the total score divided by the number of items for which the patient has responded.

Lastly, it should be noted that there were no significant intergroup differences in leg length – a factor which could otherwise have influenced our analysis of spatiotemporal variables.

1.4.2. Results of the neuropsychological evaluation

As with the results of the medical evaluation, the results of the neuropsychological screen are reported in Table 3.

1.4.3. Results of the gait tests performed in a single-TS

All the data for the three validated clinical tests and the Locometrix® trial obtained in a single-TS are presented in Tables 4–7. The pull test parameters did not show any significant intergroup differences. During the one-leg balance test, body movements differentiated the AD patients from the healthy controls. However, the MCI subjects presented intermediate values that did not differentiate them from the other two groups.

The timed "get up and go" test revealed a significant difference between the gait profile of the AD patients (with a

Table 1
Medical assessment.

Parameters	Controls	MCI	AD
Age (years)	73.53	72.85	73.66
Woman (%)	21	21	9
Man (%)	19	21	9
BMI (absolute value)	27.4	23.6	24.1
Katz scale (/24)	6	6	6
Lawton scale (calculated)	0.25	0.26	0.295
GDS (/15)	0.08	0.18	0.26
MNA (/30)	28.6	25.2	25.6
EVA (/10)	0.6	0.7	1.2
SHERPA (/11.5)	0.9	0.6	0.8

Table 2
Medical characteristics.

Habits	Tobacco, 6% Alcohol 1 or 2 glasses wine/day, 27% Physical activity at least 1 h/week, 66%
Sensory	Subjects needing glasses, 72% Subjects needing hears prothesis, 18%
Medical past history	Arterial hypertension or myocardial thrombosis, 57% AIT, nevralgie trijumeau ou migraine, 10% Diabetes, 28% Dyslipidemia, 28% Hypothyroidism, 14% Moderate arthrosis, 27% Anxiety, mood disorders, 9%
Sedative Drugs	Antidepressant, 12%; a ISSR in 75% of cases Sleeping drugs, 28%, a BZD in 66% of cases

longer completion time, a higher number of steps and a lower qualitative score) and those of the other two groups. In contrast, this test did not evidence any significant gait profile differences between MCI subjects and healthy controls.

In a single-TS, the Locometrix® better differentiated between the three populations. In fact, the AD group stood out from the healthy control group in terms of a lower gait speed and stride length and stood out from the MCI subjects in terms of the regularity parameter. In turn, the MCI group had a significantly lower gait frequency than the healthy controls.

1.4.4. Results of the single cognitive task

The countdown number reached (i.e., the rapidity of the countdown) differed significantly for the three populations (Table 8). In contrast, there was no intergroup difference in the number of errors.

Table 3
Results of the neuropsychological evaluation.

Neuropsychological variable	Healthy controls ($m \pm$ S.D.)	MCI subjects ($m \pm$ S.D.)	AD patients ($m \pm$ S.D.)
IADL score	0.26 ± 0.02 (a)	0.26 ± 0.02 (a)	0.31 ± 0.08 (b)
Years in education	13.71 ± 3.73 (a)	13.64 ± 3.30 (a)	9.33 ± 3.78 (a)
MMSE	28.21 ± 1.58 (a)	26.71 ± 1.68 (a)	22.83 ± 2.14 (b)
Total Mattis score	139.28 ± 4.79 (a)	135.57 ± 5.54 (a, b)	129 ± 11.30 (b)
Mattis attention subscore	36.55 ± 0.85 (a)	36 ± 0.96 (a)	35 ± 2.37 (a)
Mattis initiation subscore	34.71 ± 2.37 (a)	33.71 ± 3.85 (a)	31.66 ± 6.06 (a)
Mattis construction subscore	6 ± 0.00 (a)	6 ± 0.00 (a)	5 ± 0.90 (b)
Mattis conception subscore	38.07 ± 1.27 (a)	37.78 ± 1.25 (a,b)	36.16 ± 2.32 (b)
Mattis memory subscore	24 ± 1.24 (a)	22.07 ± 2.56 (a)	21.16 ± 4.40 (a)
G and B – total FR	28.35 ± 5.87 (a)	19.07 ± 9.51 (b)	12 ± 10.49 (c, b)
G and B – total IFR	45.64 ± 2.47 (a)	38.64 ± 8.51 (b, c)	30 ± 9.59 (c)
G and B – difference in FR	11.57 ± 2.95 (a)	6.14 ± 3.96 (b, c)	3.75 ± 3.86 (c)
G and B – difference in IFR	15.35 ± 1.01 (a)	13.21 ± 2.89 (a, b)	9.75 ± 5.19 (b)
G and B – Rec	16 ± 0.00 (a)	15.15 ± 1.46 (a)	15 ± 1.41 (a)
Rey Figure time (s)	189.64 ± 95.10 (a)	189.6 ± 98.59 (a)	375.8 ± 184.93 (b)
Rey Figure score	32.57 ± 3.27 (a)	30.6 ± 3.93 (a)	23.66 ± 6.82 (b)
AS alert – mean RT	240.35 ± 36.24 (a)	292.26 ± 81.74 (a, b)	355.5 ± 90.34 (b)
AS alert – correct resp.	40 ± 0.00 (a)	39.85 ± 0.36 (a)	40 ± 0.00 (a)
AS alert – omissions	0.07 ± 0.27 (a)	0.07 ± 0.27 (a)	0 ± 0.00 (a)
AS alert – incorrect resp.	1.15 ± 0.77 (a)	1.36 ± 0.50 (a)	1 ± 0.63 (a)
AS alert – early resp.	1.82 ± 2.25 (a)	0.92 ± 1.14 (a)	1.5 ± 1.05 (a)
SS alert – mean RT	252.14 ± 60.50 (a)	295.56 ± 63.03 (a)	392.66 ± 113.42 (b)
SS alert – correct resp.	40 ± 0.00 (a)	39.85 ± 0.36 (a)	40 ± 0.00 (a)
SS alert – omissions	0.21 ± 0.80 (a)	0.07 ± 0.27 (a)	0.66 ± 1.63 (a)
SS alert. – incorrect resp.	1.21 ± 0.58 (a)	0.71 ± 0.47 (a)	1.33 ± 1.03 (a)
SS alert – early resp.	0 ± 0.00 (a)	0 ± 0.00 (a)	0 ± 0.00 (a)
Alertness index	0.007 ± 0.14 (a)	0.004 ± 0.14 (a)	0.084 ± 0.11 (a)
Div. att. – ST squares RT	939.07 ± 117.30 (a)	980 ± 180.27 (a)	1227 ± 274.47 (b) (a)
Div. att. – ST corr. resp. squares	14.64 ± 2.06 (a)	15.28 ± 2.46 (a)	12.5 ± 3.78 (a)
Div. att. – ST RT sounds	598.28 ± 109.16 (a)	600.35 ± 81.42 (a)	700.83 ± 212.18 (a)
Div. att. – ST corr. resp. sounds	15.93 ± 0.27 (a)	15.85 ± 0.36 (a,b)	15 ± 1.67 (b)
Div. att. – DT RT squares	909.07 ± 100.47 (a)	944.78 ± 128.34 (a,b)	1120.67 ± 298.45 (b)
Div. att. – DT corr. resp. squares	15.4 ± 1.92 (a)	14.57 ± 2.06 (a)	13.83 ± 3.13 (a)
Div. att. – DT omissions squares	2 ± 1.92 (a)	2.42 ± 2.06 (a)	3.16 ± 3.13 (a)
Div. att. – DT RT sounds	650 ± 79.66 (a)	710.57 ± 119.81 (a)	720.5 ± 82.18 (a)
Div. att. – DT corr. resp. sounds	15.35 ± 0.63 (a)	15.5 ± 0.65 (a)	13.33 ± 3.88 (a, b)
Div. att. – DT omissions sounds	0.57 ± 0.65 (a)	0.5 ± 0.65 (a,)	2.66 ± 3.88 (b)
Div. att. – DT mean RT squares + sounds	770.25 ± 63.69 (a)	821.67 ± 115.38 (a)	1048.92 ± 277.64 (b)
Div. att. corr. resp. squares + sounds	30.4 ± 2.31 (a)	30.2 ± 2.49 (a)	27.2 ± 6.24 (a)
Div. att. omissions squares + sounds	2.3 ± 2.67 (a)	2.8 ± 2.49 (a)	5.8 ± 6.24 (a)

A different letter indicates a significant difference ($p < 0.05$) between the groups.
IFR: indicate free report; fr: free report; rec: recognition; rt: reaction time; resp: response; corr: correct; div att: divided attention.

Table 4
Results for the *pull test* performed in a single-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Number of toe retropulsions	0.57 ± 0.51 (a)	0.64 ± 0.50 (a)	0.50 ± 0.55 (a)
Number of toe elevations	0.50 ± 0.52 (a)	0.71 ± 0.47 (a)	0.50 ± 0.55 (a)
Number of forward arm movements	0.21 ± 0.43 (a)	0.21 ± 0.43 (a)	0.17 ± 0.41 (a)
Number of steps back	0.21 ± 0.58 (a)	0.36 ± 0.84 (a)	1 ± 1.10 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 5
Results for the one-leg balance test performed in a single-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Time (s)	22.11 ± 8.92 (a)	22.91 ± 10.24 (a)	13.91 ± 9.52 (a)
Arm movements per second	0.22 ± 0.26 (a)	0.10 ± 0.12 (a)	0.11 ± 0.11 (a)
Body movements per second	0.04 ± 0.05 (a)	0.1 ± 0.17 (a, b)	0.20 ± 0.45 (b)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 6
Results for the timed "get up and go" test performed in a single-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Time (s)	9.25 ± 1.35 (a)	10.56 ± 1.56 (a)	16.03 ± 5.49 (b)
Stops	0 ± 0 (a)	0 ± 0 (a)	0 ± 0 (a)
Number of strides	10.64 ± 1.29 (a)	11.14 ± 1.23 (a)	13.33 ± 2.80 (b)
Qualitative evaluation	−0.07 ± 0.27 (a)	−0.21 ± 0.43 (a)	−1.5 ± 1.97 (b)

A different letter represents a significant difference ($p < 0.05$) between the groups.

1.4.5. Results of the gait tests performed in a dual-TS

For obvious practical reasons, the pull test was not performed in a dual-TS (subjects do not have time to count backwards during the pull). For the one-leg balance test performed in a dual-TS (Table 9), none of the studied variables presented significant differences between the three groups. When performed as part of a dual-TS, the timed "get up and go" test (Table 10) presented some significant differences – mainly between the AD patients on one hand and the remaining two groups on the other.

1.4.6. Gait parameters reported by the Locometrix® in a dual-TS

The gait speed was able to differentiate between the three groups (Table 11). Furthermore, the MCI subjects presented a lower stride frequency than the healthy controls, whereas the AD patients had a lower stride length and lower gait regularity than the other two groups.

Regarding the cognitive task performed in a dual-TS (Table 12), the three populations presented significantly different countdown values, whereas the number of stops did not differ significantly.

1.4.7. Comparison of the single-task/dual-task results in each population

The Tables 13–15 summarize the calculated difference between the gait parameters in the single task and in the dual task, which explains the minus sign in front of some values.

In the healthy control group (Table 13), performing an additional cognitive task led to a significant decrease in gait "speed" and the "frequency" and a significant "increase in symmetry". For the MCI group (Table 14), gait "speed" and "frequency" and stride "length" and "regularity" decreased. In contrast, there was no significant difference in "symmetry".

Table 7
Locometrix® parameters in a single-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Speed (m/s)	1.4 ± 0.13 (a)	1.22 ± 0.15 (a, b)	1.02 ± 0.36 (b)
Stride frequency (Hz)	1 ± 0.08 (a)	0.9 ± 0.05 (b)	0.95 ± 0.17 (a, b)
Stride length (m)	1.41 ± 0.10 (a)	1.36 ± 0.13 (a, b)	1.13 ± 0.45 (b)
Regularity (dimensionless)	276 ± 35 (a, b)	287 ± 29 (a)	227 ± 82 (b)
Symmetry (dimensionless)	202.79 ± 31.06 (a)	224 ± 25 (a)	209 ± 77 (a)
Stops	0 ± 0 (a)	0 ± 0 (a)	0 ± 0 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 8
Results for the cognitive task in a single-task paradigm (TS).

	Simple cognitive task		
	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Number of errors	0.07 ± 0.27 (a)	0.29 ± 0.61 (a)	0.50 ± 0.84 (a)
Number of figures counted down per second	1.25 ± 0.25 (a)	1.05 ± 0.19 (a, b)	0.85 ± 0.33 (b)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 9
Results for the one-leg balance test performed in a dual-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Time (s)	19.42 ± 10.78 (a)	21.10 ± 10.36 (a)	12.72 ± 10.87 (a)
Arm movements per second	0.28 ± 0.25 (a)	0.23 ± 0.23 (a)	0.42 ± 0.37 (a)
Body movements per second	0.11 ± 0.16 (a)	0.08 ± 0.11 (a)	0.09 ± 0.11 (a)
Number of figures counted down per second	1.19 ± 0.42 (a)	0.87 ± 0.31 (a)	0.76 ± 0.29 (a)
Number of errors	0.14 ± 0.36 [(a)]	0.14 ± 0.36 [(a)]	0.60 ± 0.89 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 10
Results for the timed "get up and go" test performed in a dual-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Time (s)	10.61 ± 1.72 (a)	12.25 ± 2.17 (a)	22.03 ± 7.74 (b)
Stops	0 ± 0 (a)	0.14 ± 0.53 (a)	0.83 ± 1.17 (b)
Number of strides	11.36 ± 1.65 (a)	11.57 ± 1.22 (a)	15 ± 5.66 (b)
Qualitative evaluation	−0.14 ± 0.36 (a)	−0.71 ± 1.07 (a)	−2.17 ± 2.04 (b)
Number of figures counted down per second	1.15 ± 0.14 (a)	0.97 ± 0.19 (a, b)	0.80 ± 0.30 (b)
Number of errors	0.29 ± 0.83 (a)	0.29 ± 0.61 (a)	0.17 ± 0.41 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 11
Locometrix® parameters in a dual-task paradigm (TS).

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Speed (m/s)	1.3 ± 0.14 (a)	1.05 ± 0.21 (b)	0.74 ± 0.26 (c)
Stride frequency (Hz)	0.94 ± 0.07 (a)	0.81 ± 0.13 (b)	0.81 ± 0.23 (a, b)
Stride length (m)	1.38 ± 0.15 (a)	1.30 ± 0.12 (a)	1 ± 0.42 (b)
Regularity (dimensionless)	258 ± 38 (a)	224 ± 47 (a)	139 ± 81 (b)
Symmetry (dimensionless)	250 ± 35 (a)	216 ± 21 (a)	206 ± 61 (a)
Stops	0 ± 0 (a)	0.07 ± 0.27 (a)	0.17 ± 0.41 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

For the AD patients (Table 15), our statistical analysis demonstrated a significant decrease in gait "speed" and "frequency" and stride "length" and "regularity".

In summary, when moving from a single task to a dual task, the MCI subjects and the AD patients presented the same gait profile modifications.

Lastly, comparison of the three groups in terms of the single-task/dual-task change in gait parameters (Table 16) showed that symmetry improved in the healthy control group but worsened in the AD and MCI subjects. This transition also meant that the MCI subjects' gait profile became more similar to that of the AD patients.

Table 12
Results for the cognitive task in a dual-task paradigm (TS).

	Simple cognitive task		
	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Number of errors	0.07 ± 0.27 (a)	0.29 ± 0.61 (a)	0.50 ± 0.84 (a)
Number of figures counted down per second	1.25 ± 0.25 (a)	1.05 ± 0.19 (a, b)	0.85 ± 0.33 (b)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 13
Comparison of the TS–TD change in the control group.

	Control group ($m \pm$ S.D.)
Speed (m/s)	-0.10 ± 0.11 (a) (b)
Stride frequency (Hz)	-0.05 ± 0.06 (a) (b)
Stride length (m)	-0.02 ± 0.11 (a) (a)
Regularity (dimensionless)	-17.86 ± 35.45 (a) (a)
Symmetry (dimensionless)	47.5 ± 33.46 (a) (b)
Stops	0 ± 0 (a) (a)
Number of errors	0.50 ± 1.16 (a) (a)
Number of figures counted down per second	-0.03 ± 0.16 (a) (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 14
Comparison of the TS–TD change in the MCI group.

	MCI group ($m \pm$ S.D.)
Speed (m/s)	-0.17 ± 0.16 (a) (b)
Stride frequency (Hz)	-0.09 ± 0.11 (a) (b)
Stride length (m)	-0.06 ± 0.06 (a) (b)
Regularity (dimensionless)	-63.4 ± 60.3 (a) (b)
Symmetry (dimensionless)	-7.14 ± 36.9 (a) (a)
Stops	0.07 ± 0.27 (a) (a)
Number of errors	0.36 ± 1.08 (a) (a)
Number of figures counted down per second	-0.10 ± 0.17 (a) (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 15
Comparison of the TS–TD change in the AD group.

	AD group ($m \pm$ S.D.)
Speed (m/s)	-0.28 ± 0.27 (a) (a)
Stride frequency (Hz)	-0.14 ± 0.09 (a) (b)
Stride length (m)	-0.13 ± 0.22 (a) (a)
Regularity (dimensionless)	-88.33 ± 89.75 (a) (a)
Symmetry (dimensionless)	-3.16 ± 45.55 (a) (a)
Stops	0.17 ± 0.41 (a) (a)
Number of errors	2.16 ± 3.92 (a) (a)
Number of figures counted down per second	-0.14 ± 0.21 (a) (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

1.4.8. Correlations between motor and neuropsychological parameters in the MCI subjects

During the analysis of the correlation between these gait variables and the MCI subjects' performance in the neuropsychological tests, we observed that the gait speed in a dual-TS was positively correlated with the total MMSE score; this would mean that (in MCI subjects) the higher the global cognitive level, the faster the gait (Table 17). Furthermore, in a single-TS, the stride frequency was positively correlated with the overall Mattis score and the Mattis initiation subscore. This seems to imply that the higher the overall cognitive performance level and the better the initiation (a process which is related to executive function, with a frontal anatomical source location), the higher the stride frequency. Gait speed and stride frequency thus appear to be related to global cognitive function. Stride frequency is also related to executive function.

Moreover, the coefficient of variation for symmetry (the change observed when moving from a single task to a dual task) is positively correlated with the lengthening of the reaction time in a divided attention paradigm. Hence, the greater the difficulty subjects have in sharing their attentional resources between two tasks (i.e., visual and auditory tasks), the higher the coefficient of variation of symmetry and the stronger the decrease in symmetry when moving to a dual-TS. These results prompt a hypothesis whereby gait symmetry is not automatic but, in fact, requires attentional resources.

1.5. Discussion

1.5.1. Discussion of the parameters obtained in a single-TS

Used in a single-TS, the three validated clinical tests were not able to significantly differentiate between the three cognitive profile groups. This observation is easily explainable in terms of:

- the nature of the parameters studied by these tests;
- the examiner-dependant (and thus relatively subjective) quantification or qualification involved.

However, when used under the same conditions, the Locometrix® enabled definition of a gait profile for each group. In fact, AD patients presented significantly lower gait speed and stride length regularity compared with the two other groups. The MCI subjects presented a lower stride frequency than did healthy controls. These results need to be confirmed in a larger population. In fact, to the best of our knowledge, no other group has used this tool in a clinical study applied to elderly people with differing cognitive profiles.

Table 16
Comparison of the TS–TD variations in the Locometrix® parameters.

	Control group ($m \pm$ S.D.)	MCI group ($m \pm$ S.D.)	AD group ($m \pm$ S.D.)
Speed (m/s)	-0.10 ± 0.11 (a)	-0.17 ± 0.16 (a)	-0.28 ± 0.27 (a)
Stride frequency (Hz)	-0.05 ± 0.06 (a)	-0.09 ± 0.11 (a)	-0.14 ± 0.09 (a)
Stride length (m)	-0.02 ± 0.11 (a)	-0.06 ± 0.06 (a)	-0.13 ± 0.22 (a)
Regularity (dimensionless)	-17.86 ± 35.45 (a)	-63.4 ± 60.3 (a)	-88.33 ± 89.75 (a)
Symmetry (dimensionless)	47.5 ± 33.46 (a)	-7.14 ± 36.9 (b)	-3.16 ± 45.55 (b)
Stops	0 ± 0 (a)	0.07 ± 0.27 (a)	0.17 ± 0.41 (a)
Number of errors	0.50 ± 1.16 (a)	0.36 ± 1.08 (a)	2.16 ± 3.92 (a)
Number of figures counted down per second	-0.03 ± 0.16 (a)	-0.10 ± 0.17 (a)	-0.14 ± 0.21 (a)

A different letter represents a significant difference ($p < 0.05$) between the groups.

Table 17
Correlation between neuropsychological variables and Locométrix® variables.

Locométrix® variables	MMSE	Overall Mattis score	Mattis initiation score	Mean reaction time in a divided attention task
Speed	**$r = 0.51$ (TS); $r = 0.55$ (TD)**	$r = 0.40$ (TS); $r = 0.38$ (TD)	$r = 0.40$ (TS); $r = 0.36$ (TD)	**$r = -0.66$ (TS); $r = -0.27$ (TD)**
Stride length	**$r = 0.59$ (TS); $r = 0.65$ (TD)**	$r = 0.15$ (TS); $r = 0.39$ (TD)	$r = 0.04$ (TS); $r = 0.25$ (TD)	**$r = -0.36$ (TS); $r = -0.22$ (TD)**
Stride rate	$r = 0.28$ (TS); $r = 0.28$ (TD)	**$r = 0.57$ (TS); $r = 0.27$ (TD)**	**$r = 0.63$ (TS); $r = 0.35$ (TD)**	$r = -0.78$ (TS); $r = -0.19$ (TD)
Regularity	$r = 0.42$ (TS); $r = 0.20$ (TD)	$r = 0.17$ (TS); $r = -0.02$ (TD)	$r = 0.01$ (TS); $r = -0.05$ (TD)	$r = -0.47$ (TS); $r = -0.17$ (TD)
Symmetry	$r = 0.46$ (TS); $r = -0.20$ (TD)	$r = 0.32$ (TS); $r = -0.24$ (TD)	$r = 0.36$ (TS); $r = -0.35$ (TD)	$r = -0.48$ (TS); $r = 0.40$ (TD)
CV Symmetry TS–DT	$r = -0.45$	$r = -0.41$	$r = -0.51$	**$r = 0.58$**

The data in bold type had a p-value < 0.05.

1.5.2. Discussion of the parameters obtained in a dual-TS

When performed as part of a dual-TS, the conventional clinical tests (the one-leg balance test, the pull test and the timed "get up and go" test) were not able to significantly differentiate the MCI subjects from the other two populations. This observation is similar to that made by Pettersson et al. in 2004 [42].

Only the timed "get up and go" test was rendered more sensitive by the addition of an additional task and was thus better able to detect gait profile perturbations in AD patients.

When evaluated with the Locometrix® in a dual-TS, the healthy controls presented gait profile modifications (relative to the profile in a single-TS) similar to those found in other studies performed with analogous dual-TS [11,23].

Similarly, in a dual-TS, the AD patients' gait parameters revealed significant differences in gait speed, stride length, stride cycle frequency and stride regularity. This observation is identical to that made by Sheridan [51] and Cocchini et al. [13].

Concerning MCI, a novel and interesting element of this work is the observation that in a dual-TS, the MCI subject's gait profile is more similar to that of AD patients than that of healthy subjects. Our results contrast with those obtained by Pettersson et al. [42]. This difference can doubtless be explained by the measurement tools used: the Tinetti test, the timed "get up and go" test, the Berg balance test and a 10-m gait track in Pettersson's work and an accelerometer monitoring movement in all three spatial planes in the present work.

1.5.3. Discussion of the single task/dual task transition for each group

For the healthy control group, the performance of a cognitive task in addition to the motor task led to a significant decrease in gait speed and stride frequency. This can be explained by an increase in the attention component of gait with age and would confirm the fact that the performance of an additional cognitive task decreases the amount of attentional resource that even a healthy subject can allocate to gait. Hence, gait speed and stride frequency may be influenced by the subject's attention status. In terms of symmetry, this parameter is improved by the addition of cognitive task, since it increases from 200 (an absolute value measured by the accelerometer, according to the measurement parameters determined by the device manufacturer) in a single-TS to 250 (again, an absolute value obtained in the same way) in a dual-TS. This can be explained by the fact that a regular countdown gives rhythm to the subject's gait. In fact, some authors suggest that gait can be improved by the use of a metronome. Thus, in a study on Parkinson's disease patients, McIntosh et al. [33] demonstrated that the presence of a metronome set to a faster rhythm than the subjects' own cadence increased gait speed and stride length. Enzenberger et al. [18] reported the same findings in similar subjects and pointed out that these positive results are not solely obtainable in a hospital setting.

In our group of MCI subjects, the transition from a single task to a dual task induced a decrease in gait speed, stride length and stride frequency and a decrease in absolute regularity values. However, the transition did not modify gait symmetry.

To explain the decrease in regularity observed in the MCI subjects, our hypothesis is that the regularity parameter studied by the Locometrix® relates to attention components of gait. The reduced attention capacities of the MCI subjects may be "overloaded" in a dual-TS and no longer allow the degree of regularity achieved in a single-TS. When combined with the clinical impression that the MCI subjects are more at risk of falling and, indeed, fall more often than healthy subjects, this observation corroborates Auvinet's suggestion that regularity is a particularly pertinent Locometrix® parameter [5].

Lastly, during the transition from single-task to dual-task conditions, the subjects in the AD group presented a significant decrease in gait speed, stride frequency, stride length and gait regularity. These results are in agreement with literature reports. In fact, Sheridan and his group [51] mentioned a decrease in speed but, above all, an increase in step-to-step variability (as measured with in-shoe sensors recording the ground contact times). Consequently, the authors concluded as to difficulty in regulating variation of the gait cycle. Visser [62] noted a decrease in stride length and frequency and an increase in double-support time. Likewise, Cocchini et al. [13] observed a reduction in gait speed, combined with a decrease in precision when the subjects were having a conversation at the same time. More recently, in a population of subjects presenting dementia (AD, vascular dementia or mixed dementia) and frontal impairment, Allali et al. [2] showed that adding a cognitive task resulted in a significant increase in the mean gait cycle time (the time needed to perform two consecutive steps) and an increase in the latter's coefficient of variation.

In summary, two novel original elements emerge from the transition from a single-TS to a dual-TS:

- the performance of MCI subjects becomes more similar to that of AD patients;
- healthy controls increase the symmetry of their gait (probably as a result of the rhythmic cue provided by the countdown).

1.5.4. Discussion concerning performance of a cognitive task in a dual-TS

When comparing changes in the cognitive task and motor task for the transition to a dual-TS, we observed a change in the gait parameters and a slowing of the countdown but no difference in the number of errors. This observation underpins the hypothesis whereby the subjects gave priority to the cognitive task. Although it is true that the attentional component of the gait increases with age, gait remains a relatively unconscious phenomenon for the subject. In contrast, the countdown is a conscious action and so, according to our hypothesis, the subject may focus his/her remaining attentional resources on this countdown task rather than on the maintenance of gait parameters.

Furthermore, Verghese et al. [60] have observed a similar situation for the priority given to a cognitive task during a study of elderly, non-demented subjects.

This study included a few elements that may have influenced the results and their interpretation. The sample size (34 subjects), the medications being taken (nearly all the patients were taking benzodiazepine sleeping pills) and the fact that falls were excluded from the patient screening interview are factors that should be taken into account in the interpretation of our observations. Lastly, the patient's state of fatigue (after having taken part in a half-day evaluation) certainly has an influence. At this stage, we believe that it is essential to pursue this work with a larger sample, in order to confirm the data (notably those regarding the AD population). Moreover, it would be useful to determine reference values for the Locometrix® variables in a population of healthy controls and for different age classes, in order to better identify changes that are strictly related to a subject's cognitive status.

1.6. Conclusions

The aim of this preliminary study was firstly to determine the potential of a new tool for the evaluation of gait parameters, the Locometrix®. We demonstrate clearly that in comparison with other validated clinical tests (the one-leg balance test, the timed ''get up and go'' test and the pull test), the Locometrix® usefully investigates gait parameters in elderly people. In fact, when used in a single-TS and then in a dual-TS, the Locometrix® is able to define a truly specific gait profile for each cognitive profile. Furthermore, gait speed and stride frequency appear to be positively correlated with the subject's global cognitive level. Likewise, stride frequency appears also related to the subject's Mattis initiation subscore. Lastly, continuation of this work should enable us to identify and confirm the existence of one or several gait parameters in MCI subjects, which could serve as predictive clinical factors of progression towards dementia in this population.

2. Version française

2.1. Introduction

Dans le monde occidental, suite au vieillissement de notre population et compte tenu des projections démographiques actuelles, la détection et le traitement des grands syndromes gériatriques prennent toute leur importance. Parmi les éléments de fragilité, les chutes ainsi que le déclin fonctionnel et cognitif sont des facteurs déterminants.

Avec l'âge, les paramètres de la marche changent. Les principales modifications rencontrées sont une réduction de la vitesse de la marche due à une réduction de la longueur des pas [12], une irrégularité du pas par augmentation des latérodéviations, une réduction de la phase d'appui unipodal et un allongement de l'appui bipodal [27,39,47,54]. Selon certains auteurs, ces changements ont pour but de rendre la marche des personnes âgées plus sûre, moins déséquilibrée [40]. Toutefois, la prévalence des chutes augmente avec l'âge et passe de 33 % à 65 ans à environ 50 % à 85 ans [55] et le taux de récidive dans l'année est élevé pour une personne sur deux rechutant [56]. L'âge « per se » ne semble donc être pas le seul facteur expliquant l'apparition des chutes chez un sujet. En effet, la présence d'un déficit cognitif constitue un facteur de risque de chute [20] indépendant des autres éléments de comorbidité. Ainsi, les sujets atteints de la maladie d'Alzheimer (MA) présentent, par rapport à une population âgée sans trouble cognitif, une diminution de la vitesse de marche [61], une diminution de la longueur des pas ainsi qu'une augmentation de la variabilité des composantes du pas d'un cycle de pas par rapport à un autre [38].

Les composantes motrices de la marche se modifient elles aussi avec l'âge ; la composante automatique (involontaire, sous-corticale) diminue au profit de la composante volontaire (corticale dite aussi attentionnelle) [7–65]. Ainsi, de nombreuses études ont montré que, lors de la marche, l'ajout d'une tâche cognitive entraînait non seulement une perturbation dans le maintien de l'équilibre postural [29,30,35] mais aussi dans les paramètres de marche conditionnant le risque de chute [11,16] ; la tâche corticale supplémentaire concurrençant la composante corticale de la marche favorisant ainsi l'apparition des troubles de la marche sous-jacents. Les sujets MA présentent un déficit des processus exécutifs et, selon l'avancement de la maladie, un déficit de la mémoire de travail entraînant une difficulté voire une incapacité de gérer simultanément deux actions [6]. Chez ces sujets, la marche en simple tâche mobilise les ressources exécutives. En situation de double tâche (TD), les sujets MA présentent donc rapidement un dépassement de leurs capacités exécutives [13,49]. Ainsi, les troubles de la marche apparaissent rapidement lors d'un déclin cognitif. De plus, récemment, de nombreux auteurs [9,63,61,1] ont montré que certaines modifications du profil de la marche précèdent le moment du diagnostic du déclin cognitif. Ces observations ouvrent la réflexion sur l'intérêt du dépistage précoce des troubles de la marche, non seulement afin de prévenir les chutes et le déclin fonctionnel mais également pour dépister précocement les déficits cognitifs. Dans ce contexte,

nous avons voulu étudier des populations de personnes âgées aux profils cognitifs différents afin d'évaluer la relation existant entre profils de marche et profils cognitifs, non seulement chez les sujets souffrant de MA mais également chez les sujets présentant un déficit cognitif isolé tel qu'observé dans le *mild cognitive impairment* (MCI).

2.2. Objectifs de l'étude

Le but de ce travail est d'étudier les paramètres de marche de trois populations au profil cognitif différent, soit des sujets sans déficit mnésique (sains), des sujets présentant un MCI et des sujets souffrant de MA.

Pour ce faire, nous avons utilisé le Locométrix®, un accéléromètre triaxial, permettant d'étudier les mouvements du corps effectués selon les trois plans de l'espace. Afin de valider cet outil dans cette application, chaque sujet a également réalisé trois tests de marche standardisés (le *one-leg balance*, le *pull test* et le *timed up and go*).

Nous souhaitons ainsi déterminer si le Locométrix® est un outil pertinent afin d'étudier la marche de personnes âgées et son efficacité par rapport aux tests cliniques classiques d'équilibre et de marche. De plus, nous pensons également que cet outil permettra de mieux définir le profil de marche des trois populations.

2.3. Population, matériel et méthode

2.3.1. Population

Les sujets sains ont été recrutés parmi des personnes volontaires de plus de 65 ans, vivant au domicile, n'ayant pas eu de chute ou d'hospitalisation les six mois précédents, se déplaçant sans aide à la marche et ne présentant aucune plainte concernant la qualité de leur marche ou leur équilibre. Les critères d'exclusion d'ordre médical étaient :

- l'accident vasculaire cérébral ayant laissé des séquelles motrices ou sensitives ;
- la maladie de Parkinson ;
- le diabète non équilibré ;
- l'hyper- ou l'hypotension non équilibrée ;
- toute pathologie cardiaque ou respiratoire qui entraînerait une faiblesse ou une dyspnée limitant la marche ;
- une prothèse de hanche ou de genou ;
- une fracture de jambe ou de cheville qui aurait laissé une incapacité à la marche ;
- de l'arthrose ou autre pathologie ostéoarticulaire invalidante.

La prise d'une benzodiazépine ou d'un antidépresseur et de petites doses de neuroleptique sans répercussion motrice est tolérée. Les patients peuvent porter des lunettes et/ou un appareil auditif mais ils doivent être pleinement satisfaits par leur utilisation. Ils doivent être avertis des objectifs et des conditions de l'étude et accepter de signer le consentement éclairé.

Les sujets MA et MCI ont été recrutés parmi les patients du centre de la mémoire du CHU de Liège (Pr E. Salmon). Le diagnostique de trouble mnésique a été réalisé selon les méthodes habituelles d'imagerie et d'évaluation médicale et neuropsychologique. Les critères diagnostiques appliqués dans notre centre sont les suivants : les patients MCI présentent un trouble cognitif isolé confirmé mais ce déficit n'entraîne aucune répercussion majeure sur leurs activités de la vie journalière. Les patients doivent avoir bénéficié des évaluations neurologiques, neuropsychologiques et de neuro-imagerie nécessaires au diagnostique et leur Clinical Dementia Rating (CDR) ne peut dépasser 0,5 [36]. Une différence est faite entre les MCI anamnestiques (qui présentent une baisse des performances mnésiques de 1,5 déviation standard maximum par rapport à la moyenne d'une population de même âge et de même niveau social) et les MCI « multiples » (qui présentent également un déficit dans une autre fonction cognitive non mnésique) [41,66]. Les critères d'exclusion sont le retard mental, une scolarisation réussie inférieure à quatre ans, un trauma crânien, une épilepsie, un cancer, une dépression, un abus de médicaments ou toute autre pathologie organique majeure. Lors de l'inclusion, les patients doivent être libres de toute médication ayant une influence sur leur performance cognitive. Le Mini Mental State Examination (MMSE) [18] réalisé doit être supérieur ou égal à 24.

Les patients déments ont été diagnostiqués comme MA probable selon les critères définis par le National Institute of Neurological and Communicative Disorders and Stroke/Alzheimer Disease and Related Disorders Association (NINCDS-ADRDA) [34]. Les critères d'exclusion appliqués sont identiques à ceux appliqués pour les MCI hormis le CDR qui est égal ou supérieur à 1. Un MMSE supérieur ou égal à 20 définit une démence légère.

Les sujets témoins (sains) ont au moins 65 ans et ne présentent aucun trouble mnésique ou cognitif lors de l'évaluation réalisée lors de leur recrutement.

Les critères d'exclusion utilisés sont identiques aux critères cités précédemment et leur score au CDR est égal à 0.

2.3.2. Évaluation médicale

L'évaluation médicale comprend une anamnèse (antécédents médicochirugicaux et médications) et un examen clinique complet (prise des paramètres, examen cardiorespiratoire, abdominal et surtout neurologique) dont le but est de vérifier l'absence de critère d'exclusion. Pour les sujets MA, une hétéroanamnèse concernant les conditions de vie, la présence de chute ou les médications a été réalisée en concertation avec la famille et si nécessaire à l'aide d'un contact téléphonique au médecin traitant. De plus, chacun d'eux a bénéficié d'une échelle Activity of Daily Living (ADL) [26] afin de mesurer leur autonomie au domicile – d'une échelle Instrumental Activity of Daily Living (IADL) [28] vérifiant leur autonomie dans les activités instrumentales de la vie journalière – d'un MNA (Mini Nutritional Assesment) [46] permettant d'évaluer leur statut nutritionnel – d'une Geriatric Depression Scale (GDS) à 15 items [51] afin d'exclure toute pathologie dépressive sous-jacente – d'une échelle visuelle analogique (EVA) [17], afin de s'assurer de l'absence de douleur lors des évaluations – d'un score hospitalier d'évaluation du risque de

perte d'autonomie (SHERPA) [14] et, enfin, afin d'évaluer globalement leur risque de perte d'autonomie et leur profil de fragilité. Le but de ces échelles est de s'assurer qu'aucune cause organique, thymique ou fonctionnelle ne vienne perturber les résultats des tests de marche et de l'évaluation neuropsychologique.

2.3.3. Évaluation neuropsychologique

Afin de vérifier que nos volontaires sains ne présentent aucun déficit mnésique, ils ont bénéficié d'un MMSE et d'une échelle de Mattis. Et afin de conserver des conditions d'évaluation similaires pour chaque groupe, les sujets MCI et MA ont eux aussi été soumis à ces épreuves. D'autres tests ont été également réalisés pour discerner d'éventuels troubles exécutifs ou praxiques susceptibles d'interférer avec les tests de marche.

L'évaluation complète réalisée par chaque sujet comprend donc :

- le MMSE [18] qui étudie l'orientation, l'apprentissage, la mémoire de travail par le calcul mental et un mot à épeler, la dénomination d'objets, la compréhension d'ordres simples et la copie d'un dessin. Cotée sur 30 points, son seuil pathologique se situe à 24 (26 pour les sujets ayant réalisé des études supérieures) ;
- l'échelle de Mattis [31] qui est une autre batterie d'évaluation cognitive globale complémentaire au MMSE. Elle explore plus largement les fonctions frontales et sous-corticofrontales et peut donc dépister d'autres types de démences que la MA. Elle explore l'attention, la mémoire, l'initiation verbale et motrice, les capacités conceptuelles et les praxies visuoconstructives. Le seuil pathologique est de 123/144, à moduler selon le niveau d'éducation du sujet ;
- une version française du test de Grober et Buschke (1987) appelée « Épreuve de rappel libre/rappel indicé à 16 items (RL/RI 16) » [61] et qui explore les capacités de mémoire épisodique. La tâche du sujet est de mémoriser quatre fiches de quatre mots appartenant à 16 catégories sémantiques différentes. Dans un premier temps, l'expérimentateur induit un encodage sémantique des mots en demandant au sujet d'identifier le mot correspondant à sa catégorie sémantique (« quelle est la fleur ? » → « la jonquille »). Lorsque les quatre mots d'une fiche ont été identifiés, la fiche est retirée et la tâche du sujet est de rappeler immédiatement ces quatre mots. Cette procédure est répliquée pour les trois fiches suivantes. Après cette phase d'apprentissage des 16 mots, les sujets entreprennent une tâche interférente de 20 secondes (comptage à rebours). Ensuite, le sujet est soumis à un premier essai de rappel libre au cours duquel il doit rappeler un maximum de mots qu'il vient d'étudier. Pour les mots qui n'ont pas été rappelés librement, l'examinateur fournit un indice (le nom de la catégorie à laquelle le mot appartient). Trois essais de rappel libre, puis de rappel indicé sont réalisés. S'ensuit une tâche de mémoire de reconnaissance des mots étudiés. Après une tâche interférente de 20 minutes, un nouvel essai de rappel libre et de rappel indicé est à nouveau propose ;
- la copie de la figure de Rey [45] afin d'évaluer les capacités d'organisation visuoconstructive et visuospatiales. La cotation est qualitative et quantitative (/36) ;
- les sous-tests informatisés d'alerte et d'attention divisée de la batterie TEA [68] ou tests d'évaluation de l'attention. Au cours du test de l'alerte phasique, le sujet doit appuyer le plus rapidement possible sur une clé réponse lorsqu'une croix apparaît au centre d'un écran. Ce test enregistre les temps de réaction du sujet (ms) et est réalisé dans deux conditions différentes (avec et sans avertisseur sonore). Au cours du test d'attention divisée, le sujet doit partager ses ressources attentionnelles entre des stimuli visuels et des stimuli auditifs. Les temps de réaction (ms) ainsi que les réponses correctes et les omissions sont enregistrés. Les temps de réaction ont également été enregistrés dans deux conditions simples : face à l'apparition de stimuli visuels seuls et face à l'apparition de stimuli auditifs seuls.

2.3.4. Évaluation motrice

Les tests de marche réalisés sont le *pull test*, le one-leg balance, le *timed up and go* et un parcours de 30 m de marche stabilisée dont les paramètres sont enregistrés au moyen du Locométrix®.

Le *pull test* est une épreuve d'équilibre statique [37]. Le sujet se tient debout les deux pieds joints. L'examinateur, placé derrière le sujet, pose ses mains sur ses épaules et lui applique une traction. L'axe corporel du sujet est alors déporté vers l'arrière. On note les mouvements nécessaires pour conserver l'équilibre et rétablir la position dans son axe initial. Sont comptabilisés les mouvements de rétropulsion, d'élévation des orteils, d'antepulsions des bras et les éventuels pas en arrière. Nous avons choisi ce test car il est rapide et ne nécessite aucun équipement particulier [3].

L'appui monopodal ou one-leg balance est également un test d'équilibre statique. Le sujet se tient debout en appui monopodal, sur la jambe de son choix, les bras le long du corps, fixant un repère visuel placé à hauteur du regard et ce le plus longtemps possible. Nous avons choisi ce test car il est facile à administrer et parce qu'il présente une excellente valeur prédictive du risque de chute ainsi que d'un faible niveau fonctionnel [59].

Le *timed up and go* est un test évaluant la marche et l'équilibre dynamique. Le sujet est assis sur une chaise avec accoudoirs. Il doit se lever sans l'aide des accoudoirs, marcher en ligne droite sur une longueur de trois mètres, faire demi-tour, puis revenir selon le même parcours et s'asseoir sur la chaise.

Les paramètres étudiés sont, d'une part, le temps nécessaire pour réaliser le parcours (normal si inférieur ou égal à 10 secondes, fragilité si temps nécessaire entre 11 et 20 secondes et important risque de chute si temps supérieur à 20 secondes) [43] et, d'autre part, une évaluation qualitative de la marche reprise du test « up and go » original [32]. Selon ce système de cotation le score de 0 est une qualité idéale, tandis qu'un score de − 5 est la plus basse cotation. Nous avons choisi ce test parce qu'il est facile, ne demande pas de matériel ni de formation particulière et parce qu'il semble bien corrélé au statut moteur global du sujet [44]. De plus, il peut identifier

avec d'excellentes spécificité et sensibilité les personnes âgées à risque de chute [52] et sa reproductibilité interexaminateur est jugée excellente [48,54].

Développée en France par l' Institut nationale de Recherche Agricole (INRA) [24], en collaboration avec le Dr Auvinet (rhumatologie et médecine physique), le Locométrix® comporte un capteur composé de deux accéléromètres disposés perpendiculairement à proximité du centre de gravité du sujet. En pratique, ce capteur est placé à l'aide d'une ceinture abdominale élastique, dans le dos du sujet et à hauteur de L3. Les axes de mesure des capteurs sont craniocaudal et médiolatéral. Un boîtier d'enregistrement portatif recueille les accélérations mesurées à la fréquence de 50 Hz, ces données sont ensuite transférées à un ordinateur sous la forme d'un tableau d'analyse statistique [4]. Lors du test, le sujet réalise, à sa vitesse de marche habituelle et confortable, un aller–retour sur une distance rectiligne de 40 m (soit 80 m au total) dans un couloir rectiligne bien éclairé, au sol régulier et plat et sans aucun obstacle ni distraction visuelle ou auditive. Deux lignes de chronométrage sont placées respectivement à 5 m après la ligne de départ et 5 m avant la ligne d'arrivée des 40 m. Le passage du sujet devant ces lignes déclenche automatiquement le chronomètre par l'intermédiaire d'une cellule photoélectrique). La vitesse de marche ainsi calculée (30 m par temps chronométré) est donc une vitesse de marche « stabilisée » ne tenant pas compte des effets de démarrage et d'arrêt de la marche. Durant le test, l'évaluateur est situé à une distance normalisée du sujet testé (fixée à 3 m en arrière) afin de suivre le décompte réalisé à voix haute durant la TD sans influencer la marche du sujet évalué. L'analyse des paramètres de marche à partir de l'accéléromètre s'effectue au cours d'une période de marche stabilisée de 20,48 secondes [15]. Cette période de marche « stabilisée » est sélectionnée par le logiciel lui-même qui repère les courbes d'accélérations dans les différents axes et sélectionne les données de la période de marche durant laquelle les variations de ces courbes sont les moins importantes (évolution des courbes « stable »). Les données (variables de marche) ainsi analysées sont [4] :

- la vitesse de marche mesurée à l'aide d'une ligne de chronométrage et exprimée en mètre par seconde ;
- la fréquence des pas ou nombre de cycle par seconde (Hz) est calculée sur base de l'accélération craniocaudale suivant l'application de la transformée de Fournier ;
- la longueur de pas est déduite de la relation vitesse = fréquence × longueur de pas et s'exprime en mètres ;
- la régularité des pas mesure la similitude en durée et en amplitude de la morphologie des courbes d'accélération craniocaudale des pas successifs les uns par rapport aux autres.

La symétrie mesure la similitude en durée et en amplitude de la morphologie des courbes d'accélération craniocaudale concernant cette fois les demi-pas droits et gauches. Selon le Dr Auvinet, les paramètres vitesse de marche et longueur des pas seraient dépendants du sexe et, même une fois rapportés à la taille, ces deux paramètres seraient plus élevés chez l'homme. La symétrie et la régularité seraient elles toutes deux indépendantes de l'âge et du sexe. Lors d'une étude comparant des sujets âgés chuteurs et non chuteurs, le Dr Auvinet constate que les variables Locométrix® permettant le plus de discriminer les deux populations sont dans l'ordre décroissant : la vitesse de marche, la régularité des pas, la fréquence des pas, puis la symétrie des demi-pas gauches sur les demi-pas droits. L'étude de la régularité de la marche serait particulièrement pertinente pour en déceler précocement la dégradation [5].

Une étude préliminaire réalisée au sain d'un échantillon de 265 sujets « témoins », basée sur six trajets de 40 m et portant sur la reproductibilité intraobservateur par comparaison des coefficients de variation entre les différents passages ayant établi que seul le deuxième passage devait être étudié et afin de conserver des conditions de passage similaires pour l'ensemble des épreuves, tous les tests ont été réalisés trois fois et seuls les résultats du deuxième passage ont été retenus. De plus, afin de sensibiliser les tests de marche, nous avons demandé aux sujets de les réaliser en tâche motrice simple (test de marche réalisé seul) mais aussi en tâche double (DT) (soit le test de marche auquel est ajouté une tâche cognitive à réaliser simultanément). Le choix de la tâche cognitive conditionne l'évolution des paramètres de marche [10]. En effet, la fluence verbale fait appel à la mémoire sémantique [21] (relativement préservée au début de la MA), alors que le décompte mobilise la mémoire de travail [22] (précocement atteinte chez les sujets MA). Selon Beauchet et al., le compte à rebours est la tâche cognitive additionnelle qui perturbe le plus les paramètres de la marche en TD [10,11]. C'est pourquoi nous avons choisi une tâche de décompte à partir de 50 comme tâche cognitive.

Lors de cette TD aucune consigne n'a été donnée au patient concernant la priorité à donner à une tâche par rapport à l'autre. De plus, l'ordre de passage du *pull test*, one-leg balance et *timed up and go*, en simple et en TD, ont été distribués de façon à ce que la succession d'examens soit la plus différente possible d'un sujet à l'autre afin de limiter les possibles influences d'apprentissage ou de fatigue lors de l'analyse de l'ensemble des résultats.

Enfin, le temps nécessaire à l'épreuve, le nombre de pas, le nombre d'erreurs de décompte ont été relevés de manière indépendante par deux opérateurs. De plus, les épreuves ont toutes été filmées afin de revoir les épreuves au ralenti et mieux observer les latérodéviations et les arrêts. Les erreurs effectuées dans le décompte et le nombre d'arrêts réalisés dans la tâche motrice ont été relevés afin d'observer à quelle tâche le sujet a spontanément donné la priorité.

En une demi-journée, chaque participant a bénéficié d'une évaluation médicale et fonctionnelle, d'un bilan neuropsychologique et d'une évaluation motrice.

Nous avons pu ainsi évaluer trois groupes de personnes de profil cognitif différent : 14 sujets « sains », 14 sujets « MCI » et six sujets présentant une MA. L'étroitesse de notre échantillon reflète d'avantage la taille de notre centre et la sévérité de nos critères de recrutement plus qu'une volonté de limiter le nombre de sujets observés. À ce stade, il est également important d'expliquer qu'il s'agit d'une étude clinique

préliminaire destine, d'une part, à évaluer la faisabilité de l'application du Locométrix® dans notre centre et à notre population et, d'autre part, à tester des différents aspects du protocole d'étude de la marche des patients âgés présentant des troubles cognitifs confirmés ou débutants.

Les résultats sont présentés sous forme de moyennes et d'écarts-types (médianes) pour les variables continues et sous forme de fréquences (pourcentages) pour les variables catégorisées. Les comparaisons entre groupes ont été réalisées à l'aide de modèles Anova ou des tests de Kruskal-Wallis, selon la distribution normale ou anormale des données continues. En cas de variables catégorisées, la comparaison a été réalisée à l'aide des tests Chi carré. Les différences entre les valeurs obtenues ont été considérées comme significatives pour un niveau d'incertitude de 5 %, soit un $p < 0{,}05$. Les calculs ont été réalisés à laide du logiciel SAS version 9.1.

2.4. Résultats

2.4.1. Caractéristiques de la population étudiée

Les Tableaux 1 et 2 résument les résultats de l'évaluation médicale dont a bénéficié l'ensemble des sujets des trois populations.

Les scores de l'échelle IADL ont été calculés en considérant, pour chaque sujet, la somme des résultats obtenus aux items applicables divisée par la somme des scores maximum des items applicables. Donc plus le chiffre est petit, plus l'autonomie est grande. Il en est de même pour la GDS ; le score obtenu est le résultat du score total divisé par le nombre d'items auquel le patient a répondu.

Notons enfin qu'il n'existe pas de différence significative entre les groupes concernant la longueur de jambs, ce qui aurait pu influencer l'analyse des variables spatiotemporelles.

2.4.2. Résultats de l'évaluation neuropsychologique

Comme pour les résultats de l'évaluation médicale, les résultats du bilan neuropsychologique sont rapportés sous forme d'un tableau (Tableau 3).

2.4.3. Résultats des tests de marche réalisés en tâche simple (TS)

L'ensemble des données obtenues en simple tâche lors des trois tests cliniques validés et du parcours étudié par le Locométrix® est présenté dans les Tableaux 4–7.

Tableau 1
Tableau d'évaluation médicale.

Variables médicales	Sains	MCI	MA
Âge (années)	73,53	72,85	73,66
Femme (%)	21	21	9
Hommes (%)	19	21	9
BMI (valeur absolue)	27,4	23,6	24,1
Échographie ADL (/24)	6	6	6
Échographie IADL (score calculé)	0,25	0,26	0,295
GDS (/15)	0,08	0,18	0,26
MNA (/30)	28,6	25,2	25,6
EVA (/10)	0,6	0,7	1,2
SHERPA (/11,5)	0,9	0,6	0,8

Tableau 2
Tableau d'évaluation médicale tous groupes confondus.

Hygiène de vie	Tabagisme, 6 % Alcool 1 à 2 verres de vin/jour, 27 % Activité physique d'1 h par semaine, 66 %
Afférences sensorielles	Trouble visuel compensé par le port de lunettes, 72 % Trouble auditif léger éventuellement compensé par une prothèse auditive, 18 %
Antécédents médicaux	HTA ou infarctus, 57 % AIT, névralgie trijumeau ou migraine, 10 % Diabète, 28 % Dyslipidémie, 28 % Hypothyroïdie, 14 % Arthrose modérée non invalidante, 27 % Trouble d'humeur de type anxiété modérée, 9 %
Médications	Antidépresseur, 12 %, un SSRI dans 75 % des cas Somnifère, 28 %, une BZD dans 66 % des cas

Les paramètres étudiés par le *pull test* ne présentent aucune différence significative selon les groupes.

Lors du *one-leg balance*, les mouvements du corps différencient les sujets MA des sujets sains mais les sujets MCI présentent des valeurs intermédiaires ne permettant pas de les différencier des deux autres groupes.

Le Timed get up and go permet de distinguer de manière significative le profil de marche des sujets MA des deux autres groupes (allongement du temps de parcours, augmentation du nombre de pas et score qualitatif plus faible). En revanche, ce test ne relève pas de différence significative entre le profil de marche des MCI et celui des sujets sains.

Le Locométrix® en TS permet de mieux différencier les trois populations. En effet, le groupe MA se distingue des sains par une diminution de la vitesse de marche et une diminution de la longueur du pas. Et il se différencie des MCI par le paramètre régularité. Les MCI quant à eux présentent par rapport aux sains une réduction significative de la fréquence des pas.

2.4.4. Résultats de la TS cognitive

Le nombre de chiffres décomptés, soit la rapidité du décompte, présente des résultats significativement différents dans les trois populations (Tableau 8). En revanche, le nombre d'erreurs ne présente pas cette différence.

2.4.5. Résultats des tests de marche réalisés en TD

Pour des raisons de réalisation évidentes (les sujets n'ont pas le temps de décompter lors d'une poussée), le *pull test* n'a pas été réalisé en TD. Concernant le *one-leg balance* réalisé en TD (Tableau 9) aucune des variables étudiées ne présente de différence significative entre les trois groupes. Réalisé en TD le *Timed up and go* (Tableau 10) présente des différences significatives surtout entre les sujets MA et les sujets des deux autres groupes.

2.4.6. Les paramètres de marche étudiés par le Locométrix® en condition de TD

Les trois groupes se différentient sur base de la vitesse de marche (Tableau 11).

Tableau 3
Évaluation neuropsychologique.

Variable neuropsychologique	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « MA » ($m \pm$ S.D.)
IADL	0,26 ± 0,02 (a)	0,26 ± 0,02 (a)	0,31 ± 0,08 (b)
Années d'étude	13,71 ± 3,73 (a)	13,64 ± 3,30 (a)	9,33 ± 3,78 (a)
MMSE	28,21 ± 1,58 (a)	26,71 ± 1,68 (a)	22,83 ± 2,14 (b)
Mattis totale	139,28 ± 4,79 (a)	135,57 ± 5,54 (a, b)	129 ± 11,30 (b)
Mattis attentionnelle	36,55 ± 0,85 (a)	36 ± 0,96 (a)	35 ± 2,37 (a)
Mattis initiation	34,71 ± 2,37 (a)	33,71 ± 3,85 (a)	31,66 ± 6,06 (a)
Mattis construction	6 ± 0,00 (a)	6 ± 0,00 (a)	5 ± 0,90 (b)
Mattis conception	38,07 ± 1,27 (a)	37,78 ± 1,25 (a, b)	36,16 ± 2,32 (b)
Mattis mémoire	24 ± 1,24 (a)	22,07 ± 2,56 (a)	21,16 ± 4,40 (a)
G et B – RL total	28,35 ± 5,87 (a)	19,07 ± 9,51 (b)	12 ± 10,49 (c, b)
G et B – RLI total	45,64 ± 2,47 (a)	38,64 ± 8,51 (b,c)	30 ± 9,59 (c)
G et B – RL différent	11,57 ± 2,95 (a)	6,14 ± 3,96 (b,c)	3,75 ± 3,86 (c)
G et B – RLI différent	15,35 ± 1,01 (a)	13,21 ± 2,89 (a,b)	9,75 ± 5,19 (b)
G et B – Rec	16 ± 0,00 (a)	15,15 ± 1,46 (a)	15 ± 1,41 (a)
Rey copie sec	189,64 ± 95,10 (a)	189,6 ± 98,59 (a)	375,8 ± 184,93 (b)
Rey copie score	32,57 ± 3,27 (a)	30,6 ± 3,93 (a)	23,66 ± 6,82 (b)
Alerte AS–TR moyenne	240,35 ± 36,24 (a)	292,26 ± 81,74 (a,b)	355,5 ± 90,34 (b)
Alerte AS – rep correcte	40 ± 0,00 (a)	39,85 ± 0,36 (a)	40 ± 0,00 (a)
Alerte AS – omissions	0,07 ± 0,27 (a)	0,07 ± 0,27 (a)	0 ± 0,00 (a)
Alerte AS – Rep abérrante	1,15 ± 0,77 (a)	1,36 ± 0,50 (a)	1 ± 0,63 (a)
Alerte AS – rep antic	1,82 ± 2,25 (a)	0,92 ± 1,14 (a)	1,5 ± 1,05 (a)
Alerte SS – TR moyenne	252,14 ± 60,50 (a)	295,56 ± 63,03 (a)	392,66 ± 113,42 (b)
Alerte SS – rep correcte	40 ± 0,00 (a)	39,85 ± 0,36 (a)	40 ± 0,00 (a)
Alerte SS – omission	0,21 ± 0,80 (a)	0,07 ± 0,27 (a)	0,66 ± 1,63 (a)
Alerte SS – rep abérrante	1,21 ± 0,58 (a)	0,71 ± 0,47 (a)	1,33 ± 1,03 (a)
Alerte SS – rep antic	0 ± 0,00 (a)	0 ± 0,00 (a)	0 ± 0,00 (a)
Index alerte	0,007 ± 0,14 (a)	0,004 ± 0,14 (a)	0,084 ± 0,11 (a)
Att Div – TS carrés TR	939,07 ± 117,30 (a)	980 ± 180,27 (a)	1227 ± 274,47 (b, a)
Att Div – TS rep corr carrés	14,64 ± 2,06 (a)	15,28 ± 2,46 (a)	12,5 ± 3,78 (a)
Att Div – TS TR sons	598,28 ± 109,16 (a)	600,35 ± 81,42 (a)	700,83 ± 212,18 (a)
Att Div – TS rep correcte sons	15,93 ± 0,27 (a)	15,85 ± 0,36 (a,b)	15 ± 1,67 (b)
Att Div – TD TR carrés	909,07 ± 100,47 (a)	944,78 ± 128,34 (a,b)	1120,67 ± 298,45 (b)
Att Div – TD rep correcte carrés	15,4 ± 1,92 (a)	14,57 ± 2,06 (a)	13,83 ± 3,13 (a)
Att Div – TD rep om carrés	2 ± 1,92 (a)	2,42 ± 2,06 (a)	3,16 ± 3,13 (a)
Att Div – TD TR sons	650 ± 79,66 (a)	710,57 ± 119,81 (a)	720,5 ± 82,18 (a)
Att Div – TD rep correcte sons	15,35 ± 0,63 (a)	15,5 ± 0,65 (a)	13,33 ± 3,88 (a, b)
Att Div – TD rep om sons	0,57 ± 0,65 (a)	0,5 ± 0,65 (a,)	2,66 ± 3,88 (b)
Att Div – TD moy TR corrés + sons	770,25 ± 63,69 (a)	821,67 ± 115,38 (a)	1048,92 ± 277,64 (b)
Att Div rep corr carrés + sons	30,4 ± 2,31 (a)	30,2 ± 2,49 (a)	27,2 ± 6,24 (a)
Att Div rep om carrés + sons	2,3 ± 2,67 (a)	2,8 ± 2,49 (a)	5,8 ± 6,24 (a)

Une lettre différente représente une différence significative ($p < 0,05$) entre les groupes.
Rec : reconnaissance ; sec : seconde ; antic : attention divisée ; corr : correcte, réponse correcte ; om : ommise, réponse ommise.

Les MCI présentent une perte de la fréquence des pas par rapport aux sujets sains. Et les sujets MA présentent une diminution de la longueur des pas et de la régularité de la marche par rapport aux deux autres groupes.

Concernant la tâche cognitive réalisée en DT (Tableau 12), les trois populations présentent un nombre de chiffres décomptés significativement différent. Le nombre d'arrêts ne détecte pas de différence entre les trois groupes.

2.4.7. Comparaison des résultats TS–DT dans chaque population

Les (Tableaux 13–15) reprennent le calcul de la différence entre les paramètres de marche révélés lors de la TS et lors de la DT, ce qui explique le signe (−) devant certaines valeurs.

Dans le groupe de sujets « sains » (Tableau 13), la réalisation d'une tâche cognitive complémentaire a un effet significatif sur la « vitesse » et la « fréquence » des pas qui « diminuent » alors

Tableau 4
Résultats du *pull test* réalisé en tâche simple (TS).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Nombre de rétropulsion des orteils	0,57 ± 0,51 (a)	0,64 ± 0,50 (a)	0,50 ± 0,55 (a)
Nombre d'élévation des orteils	0,50 ± 0,52 (a)	0,71 ± 0,47 (a)	0,50 ± 0,55 (a)
Nombre d'antépulsion des bras	0,21 ± 0,43 (a)	0,21 ± 0,43 (a)	0,17 ± 0,41 (a)
Nombre de pas en arrière	0,21 ± 0,58 (a)	0,36 ± 0,84 (a)	1 ± 1,10 (a)

Une lettre différente représente une différence significative ($p < 0,05$) entre les groupes.

Tableau 5
Résultats du *one-leg balance* réalisé en tâche simple (TS).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Temps (s)	22,11 ± 8,92 (a)	22,91 ± 10,24 (a)	13,91 ± 9,52 (a)
Mouvements de bras par seconde	0,22 ± 0,26 (a)	0,10 ± 0,12 (a)	0,11 ± 0,11 (a)
Mouvements de corps par seconde	0,04 ± 0,05 (a)	0,1 ± 0,17 (a, b)	0,20 ± 0,45 (b)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 6
Résultats du *timed up and go* réalisé en tâche simple (TS).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Temps (s)	9,25 ± 1,35 (a)	10,56 ± 1,56 (a)	16,03 ± 5,49 (b)
Arrêts	0 ± 0 (a)	0 ± 0 (a)	0 ± 0 (a)
Nombre de pas	10,64 ± 1,29 (a)	11,14 ± 1,23 (a)	13,33 ± 2,80 (b)
Évaluation qualitative	− 0,07 ± 0,27 (a)	− 0,21 ± 0,43 (a)	− 1,5 ± 1,97 (b)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 7
Paramètres Locométrix® en tâche simple (TS).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Vitesse (m/s)	1,4 ± 0,13 (a)	1,22 ± 0,15 (a, b)	1,02 ± 0,36 (b)
Fréquence de pas (Hz)	1 ± 0,08 (a)	0,9 ± 0,05 (b)	0,95 ± 0,17 (a, b)
Longueur de pas (m)	1,41 ± 0,10 (a)	1,36 ± 0,13 (a, b)	1,13 ± 0,45 (b)
Régularité (sans dimension)	276 ± 35 (a, b)	287 ± 29 (a)	227 ± 82 (b)
Symétrie (sans dimension)	202,79 ± 31,06 (a)	224 ± 25 (a)	209 ± 77 (a)
Arrêts	0 ± 0 (a)	0 ± 0 (a)	0 ± 0 (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

que la « symétrie augmente ». Pour le groupe « MCI » (Tableau 14), ce sont non seulement la « vitesse » et la « fréquence » mais également la « longueur » et la « régularité des pas » qui sont diminuées. En revanche, il n'existe pas de différence significative en ce qui concerne la symétrie.

Pour les patients « Alzheimer » (Tableau 15), l'analyse statistique met également en évidence une baisse significative de la vitesse de marche, la « fréquence, la longueur et la régularité » des pas.

En résumé, lors du passage de la TS à la DT, les sujets MCI et les sujets MA présentent donc les mêmes modifications de leur profil de marche.

Enfin, la comparaison de l'évolution des paramètres de marche entre TS et TD entre les différents groupes (Tableau 16), montre que la symétrie est améliorée chez les sains, alors qu'elle se dégrade chez les MA et les MCI. Cet élément rapproche également le profil de marche des MCI de celui des MA.

2.4.8. Corrélations entre les paramètres moteurs et les paramètres neuropsychologiques des MCI

Lors de l'analyse de la corrélation entre ces variables de la marche et les performances aux tests neuropsychologiques des MCI, d'une part, la vitesse de marche en DT est positivement corrélée au score total du MMSE ; cela signifierait que, chez les MCI, plus le niveau cognitif global est élevé, plus ils sont rapides (Tableau 17). D'autre part, en TS, la fréquence du pas est positivement corrélée au score de la Mattis et au score en initiation. Cela semble signifier que plus le niveau cognitif global est bon et l'initiation importante (processus relevant des fonctions exécutives, de siège anatomique frontal), plus la fréquence du pas est élevée. La vitesse de marche et la fréquence du pas semblent donc liées au fonctionnement cognitif global. La fréquence étant de plus aussi liée au fonctionnement exécutif.

De plus, le coefficient de variation de la symétrie (variation observée lors du passage de la TS à la DT) est positivement

Tableau 8
Résultats de la tâche cognitive en tâche simple (TS).

	Tâche simple cognitive		
	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Nombre d'erreurs	0,07 ± 0,27 (a)	0,29 ± 0,61 (a)	0,50 ± 0,84 (a)
Nombre de chiffres décomptés par seconde	1,25 ± 0,25 (a)	1,05 ± 0,19 (a,b)	0,85 ± 0,33 (b)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 9
Résultats du *one-leg balance* réalisé en double tâche (TD).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Temps (s)	19,42 ± 10,78 (a)	21,10 ± 10,36 (a)	12,72 ± 10,87 (a)
Mouvements de bras par seconde	0,28 ± 0,25 (a)	0,23 ± 0,23 (a)	0,42 ± 0,37 (a)
Mouvements de corps par seconde	0,11 ± 0,16 (a)	0,08 ± 0,11 (a)	0,09 ± 0,11 (a)
Nombre de chiffres décomptés par seconde	1,19 ± 0,42 (a)	0,87 ± 0,31 (a)	0,76 ± 0,29 (a)
Nombre d'erreurs	0,14 ± 0,36 (a)	0,14 ± 0,36 (a)	0,60 ± 0,89 (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 10
Résultats du *timed up and go* réalisé en double tâche (TD).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Temps (s)	10,61 ± 1,72 (a)	12,25 ± 2,17 (a)	22,03 ± 7,74 (b)
Arrêts	0 ± 0 (a)	0,14 ± 0,53 (a)	0,83 ± 1,17 (b)
Nombre de pas	11,36 ± 1,65 (a)	11,57 ± 1,22 (a)	15 ± 5,66 (b)
Évaluation qualitative	− 0,14 ± 0,36 (a)	− 0,71 ± 1,07 (a)	− 2,17 ± 2,04 (b)
Nombre de chiffres décomptés par seconde	1,15 ± 0,14 (a)	0,97 ± 0,19 (a, b)	0,80 ± 0,30 (b)
Nombre d'erreurs	0,29 ± 0,83 (a)	0,29 ± 0,61 (a)	0,17 ± 0,41 (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 11
Paramètres Locométrix® en double tâche (TD).

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Vitesse (m/s)	1,3 ± 0,14 (a)	1,05 ± 0,21 (b)	0,74 ± 0,26 (c)
Fréquence de pas (Hz)	0,94 ± 0,07 (a)	0,81 ± 0,13 (b)	0,81 ± 0,23 (a, b)
Longueur de pas (m)	1,38 ± 0,15 (a)	1,30 ± 0,12 (a)	1 ± 0,42 (b)
Régularité (sans dimension)	258 ± 38 (a)	224 ± 47 (a)	139 ± 81 (b)
Symétrie (sans dimension)	250 ± 35 (a)	216 ± 21 (a)	206 ± 61 (a)
Arrêts	0 ± 0 (a)	0,07 ± 0,27 (a)	0,17 ± 0,41 (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

corrélé à l'allongement du temps de réaction en attention divisée. Donc, plus les sujets ressentent des difficultés à partager leurs ressources attentionnelles entre deux tâches (visuelle et auditive), plus le coefficient de variation de la symétrie est élevé, c'est-à-dire plus la symétrie diminue lors du passage en TD. Ces résultats conduisent à l'hypothèse que la symétrie de la marche ne serait pas automatique mais demanderait des ressources attentionnelles.

2.5. Discussion

2.5.1. Discussion des paramètres obtenus en TS

Utilisés en simple tâche, les trois tests cliniques validés n'ont pas permis de différencier significativement les trois sous-groupes de profils cognitifs. Cette constatation est facilement explicable par la nature des paramètres étudiés par ces tests mais aussi par leur quantification ou qualification examinateur dépendante et donc relativement subjective. En revanche, le Locométrix®, utilisé dans les mêmes conditions, a permis de mieux définir le profil de marche de chaque groupe. En effet, les sujets MA présente une diminution significative de la vitesse et de la régularité de marche et de longueur des pas par rapport aux deux autres groupes. Tandis que les sujets MCI présentent une diminution de la fréquence des pas par rapport aux sujets sains. Ces résultats mériteraient d'être confirmés sur une population plus importante. En effet, à notre connaissance, aucune autre équipe jusqu'à maintenant a déjà utilisé cet outil dans une étude clinique appliquées à des personnes âgées de profils cognitifs différents.

Tableau 12
Résultats de la tâche cognitive en double tâche (TD).

	Tâche simple cognitive		
	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Nombre d'erreurs	0,07 ± 0,27 (a)	0,29 ± 0,61 (a)	0,50 ± 0,84 (a)
Nombre de chiffres décomptés par seconde	1,25 ± 0,25 (a)	1,05 ± 0,19 (a, b)	0,85 ± 0,33 (b)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 13
Comparaison de l'évolution TS–TD dans le groupe témoin.

	Groupe « témoin » ($m \pm$ S.D.)
Vitesse (m/s)	− 0,10 ± 0,11 (a) (b)
Fréquence de pas (Hz)	− 0,05 ± 0,06 (a) (b)
Longueur de pas (m)	− 0,02 ± 0,11 (a) (a)
Régularité (sans dimension)	− 17,86 ± 35,45 (a) (a)
Symétrie (sans dimension)	47,5 ± 33,46 (a) (b)
Arrêts	0 ± 0 (a) (a)
Nombre d'erreurs	0,50 ± 1,16 (a) (a)
Nombre de chiffres décomptés par seconde	− 0,03 ± 0,16 (a) (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 14
Comparaison de l'évolution TS–DT dans le groupe *mild cognitive impairment* (MCI).

	Groupe « MCI » ($m \pm$ S.D.)
Vitesse (m/s)	− 0,17 ± 0,16 (a) (b)
Fréquence de pas (Hz)	− 0,09 ± 0,11 (a) (b)
Longueur de pas (m)	− 0,06 ± 0,06 (a) (b)
Régularité (sans dimension)	− 63,4 ± 60,3 (a) (b)
Symétrie (sans dimension)	− 7,14 ± 36,9 (a) (a)
Arrêts	0,07 ± 0,27 (a) (a)
Nombre d'erreurs	0,36 ± 1,08 (a) (a)
Nombre de chiffres décomptés par seconde	− 0,10 ± 0,17 (a) (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 15
Comparaison de l'évolution TS–DT dans le groupe de la maladie d'Alzheimer (MA).

	Groupe « MA » ($m \pm$ S.D.)
Vitesse (m/s)	− 0,28 ± 0,27 (a) (a)
Fréquence de pas (Hz)	− 0,14 ± 0,09 (a) (b)
Longueur de pas (m)	− 0,13 ± 0,22 (a) (a)
Régularité (Sans dimension)	− 88,33 ± 89,75 (a) (a)
Symétrie (sans dimension)	− 3,16 ± 45,55 (a) (a)
Arrêts	0,17 ± 0,41 (a) (a)
Nombre d'erreurs	2,16 ± 3,92 (a) (a)
Nombre de chiffres décomptés par sec	− 0,14 ± 0,21 (a) (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

2.5.2. Discussion des paramètres obtenus en TD

Réalisés en condition de TD, les tests cliniques « classiques » (*one-leg balance*, *pull test* et *timed up and go*) n'ont pas permis de distinguer significativement les sujets MCI des deux autres populations Cette observation est semblable à celle faite par Pettersson et al. en 2004 [42].

Seul le *timed up and go* a été sensibilisé par l'ajout de la TD permettant ainsi de mieux percevoir les perturbations du profil de marche des MA.

Évalués par le Locométrix®, les sujets sains en TD présentent (par rapport à leur profil en TS) des modifications du profil de marche comparables à celles relevées dans d'autres travaux réalisés en conditions similaires de TD [11,23].

De même, les paramètres de marche des DA en TD montrent une différence significative concernant la vitesse de la marche, la longueur des pas, la fréquence des cycles de pas et leur régularité. Cette observation est identique à celle faite par Sheridan [51] et Cocchini et al. [13].

Concernant les MCI, un élément original et intéressant de ce travail est l'observation qu'en TD leur profil de marche se rapproche plus de celui des MA que de celui des sujets sains. Ces résultats s'opposent à ceux obtenus par Pettersson et al. [42]. Cette différence peut sans doute être expliquée par les outils de mesure utilisés ; Tinetti, *timed up and go*, Berg balance et un parcours de marche de 10 m pour l'équipe de Pettersson ou un accéléromètre étudiant les mouvements dans les trois plans de l'espace dans ce travail.

2.5.3. Discussion passage TS–DT par groupe

Au sein du groupe de sujets sains, la réalisation d'une tâche cognitive complémentaire à la tâche motrice entraîne une diminution significative de la vitesse et de la fréquence des pas, ce qui peut être expliqué par l'augmentation de la composante attentionnelle de la marche avec l'âge ; cela confirmerait le fait que l'ajout d'une tâche cognitive supplémentaire diminue l'importance de l'attention que le sujet sain même peut allouer à sa marche. La vitesse de marche et la fréquence des pas seraient donc influencées par le statut attentionnel du sujet. La symétrie, elle, est améliorée par l'ajout de la tâche cognitive passant de 200 (valeur absolue mesurée par l'accéléromètre selon des paramètres de mesure déterminé par le constructeur) en TS à 250 (valeur absolue obtenue de manière identique) en TD. Cela peut être expliqué par le fait qu'un décompte régulier donne un rythme à la marche du sujet. En effet, d'après certains auteurs, la marche pourrait être améliorée à l'aide d'un métronome. Ainsi, McIntosh et al. [33], dans une étude chez des parkinsoniens, démontrent qu'en faisant marcher les sujets en présence d'un métronome réglé à un rythme plus rapide que leur propre cadence, leur vitesse de marche et leur longueur de

Tableau 16
Comparaison des variations des paramètres Locométrix® TS–TD.

	Groupe « témoin » ($m \pm$ S.D.)	Groupe « MCI » ($m \pm$ S.D.)	Groupe « Alzheimer » ($m \pm$ S.D.)
Vitesse (m/s)	− 0,10 ± 0,11 (a)	− 0,17 ± 0,16 (a)	− 0,28 ± 0,27 (a)
Fréquence de pas (Hz)	− 0,05 ± 0,06 (a)	− 0,09 ± 0,11 (a)	− 0,14 ± 0,09 (a)
Longueur de pas (m)	− 0,02 ± 0,11 (a)	− 0,06 ± 0,06 (a)	− 0,13 ± 0,22 (a)
Régularité (sans dimension)	− 7,86 ± 35,45 (a)	− 63,4 ± 60,3 (a)	− 88,33 ± 89,75 (a)
Symétrie (sans dimension)	47,5 ± 33,46 (a)	− 7,14 ± 36,9 (b)	− 3,16 ± 45,55 (b)
Arrêts	0 ± 0 (a)	0,07 ± 0,27 (a)	0,17 ± 0,41 (a)
Nombre d'erreurs	0,50 ± 1,16 (a)	0,36 ± 1,08 (a)	2,16 ± 3,92 (a)
Nombre de chiffres décomptés par seconde	− 0,03 ± 0,16 (a)	− 0,10 ± 0,17 (a)	− 0,14 ± 0,21 (a)

Une lettre différente représente une différence significative ($p < 0{,}05$) entre les groupes.

Tableau 17
Corrélation entre les variables neuropsychologiques et les variables Locométrix®.

Variables Locométrix®	MMSE	Score total de Mattis	Score initiation Mattis	Temps de réaction moyen en attention divisée
Vitesse	**$r = 0{,}51$ (TS) ; $r = 0{,}55$ (TD)**	$r = 0{,}40$ (TS) ; $r = 0{,}38$ (TD)	$r = 0{,}40$ (TS) ; $r = 0{,}36$ (TD)	**$r = -0{,}66$ (TS) ; $r = -0{,}27$ (TD)**
Longueur des pas	**$r = 0{,}59$ (TS) ; $r = 0{,}65$ (TD)**	$r = 0{,}15$ (TS) ; $r = 0{,}39$ (TD)	$r = 0{,}04$ (TS) ; $r = 0{,}25$ (TD)	$r = -0{,}36$ (TS) ; $r = -0{,}22$ (TD)
Fréquence des pas	$r = 0{,}28$ (TS) ; $r = 0{,}28$ (TD)	**$r = 0{,}57$ (TS)** ; $r = 0{,}27$ (TD)	**$r = 0{,}63$ (TS) ; $r = 0{,}35$ (TD)**	**$r = -0{,}78$ (TS) ; $r = -0{,}19$ (TD)**
Régularité	$r = 0{,}42$ (TS) ; $r = 0{,}20$ (TD)	$r = 0{,}17$ (TS) ; $r = -0{,}02$ (TD)	$r = 0{,}01$ (TS) ; $r = -0{,}05$ (TD)	$r = -0{,}47$ (TS) ; $r = -0{,}17$ (TD)
Symétrie	$r = 0{,}46$ (TS) ; $r = -0{,}20$ (TD)	$r = 0{,}32$ (TS) ; $r = -0{,}24$ (TD)	$r = 0{,}36$ (TS) ; $r = -0{,}35$ (TD)	$r = -0{,}48$ (TS) ; $r = 0{,}40$ (TD)
CV Symétrie TS–DT	$r = -0{,}45$	$r = -0{,}41$	$r = -0{,}51$	**$r = 0{,}58$**

Les données en caractères « gras » présentent une valeur de $p < 0{,}05$.

pas sont augmentées. Enzenberger et al. [18] montrent la même chose chez ces sujets et ajoute que ces résultats positifs ne sont pas obtenus uniquement dans un hôpital.

Dans le groupe des MCI, le passage de le TS à la TD entraîne une diminution de la vitesse de la marche, de la longueur et de la fréquence des pas ainsi qu'une diminution des valeurs absolue de régularité sans modifier la symétrie.

Pour expliquer la diminution de régularité observée chez les MCI, l'hypothèse avancée est que la régularité étudiée par le Locométrix® relève des composantes attentionnelles de la marche. Les capacités attentionnelles réduites des sujets MCI seraient « dépassées » en condition de TD et ne pourraient plus assurer une régularité équivalente à celle développée dans les conditions de simple tâche. Cette observation ainsi que l'impression clinique que les sujets MCI sont plus à risque de chute et chutent plus souvent que les sujets sains viendraient corroborer l'hypothèse d'Auvinet pour qui la régularité semble un paramètre Locométrix® particulièrement pertinent [5].

Enfin, les sujets du groupe MA présentent lors du passage de la TS à la TD une diminution significative de la vitesse de marche, de la fréquence et de la longueur des pas ainsi que de la régularité de la marche. Ces résultats sont en accord avec la littérature. En effet, Sheridan [50] évoque une diminution de vitesse mais surtout une augmentation de variabilité de pas à pas mesurée à l'aide de capteurs placés dans les chaussures, ceux-ci enregistrant le temps de contact au sol. Ils en concluent une difficulté à réguler la variation du cycle de marche. Visser [65] constate une diminution de la longueur de pas, de la diminution de fréquence de pas et une augmentation du temps de double appui. De même, Cocchini et al. [13] retrouvent une réduction de la vitesse de marche associée à une diminution de sa précision lorsque les sujets tenaient une conversation en même temps. Plus récemment, Allali et al. [2] ont montré que l'ajout d'une tâche cognitive au sein d'une population de sujets présentant une démence (MA ou démences vasculaires ou démences mixtes) et un déficit frontal entraînait une augmentation significative de la valeur moyenne du temps du cycle de marche ainsi qu'une augmentation du coefficient de variation de ce temps (le temps du cycle de marche étant le temps nécessaire pour réaliser deux pas consécutifs).

En résumé, il y a deux éléments originaux qui émergent de ce passage de la TS à la TD :

- d'une part, les sujets MCI se rapprochent des performances des MA ;
- d'autre part, les sujets sains améliorent la symétrie de leur marche, probablement grâce au rythme imprimé par le décompte.

2.5.4. Discussion concernant la réalisation de la tâche cognitive en TD

Lorsque l'évolution de la tâche cognitive et de la tâche motrice lors du passage en TD sont comparées, il existe une altération des paramètres de marche et un ralentissement du décompte, mais aucune différence concernant le nombre d'erreurs. Cette observation sous-tend l'hypothèse que les sujets ont donné la priorité à la tâche cognitive. Car s'il est vrai que la composante attentionnelle de la marche augmente avec l'âge, cela reste un phénomène relativement inconscient pour le sujet. Le décompte quant à lui est une action consciente et, selon notre hypothèse, le sujet orienterait sa réserve attentionnelle vers cette action de décompte plutôt que vers le maintien des paramètres de la marche.

J. Verghese et al. [60] ont d'ailleurs observé une similarité dans la priorité donnée à la tâche cognitive lors d'un travail réalisé avec des sujets âgés non déments.

Cette étude comprend quelques éléments pouvant influencer les résultats obtenus et leur interprétation. La taille de l'échantillon (34 sujets), les médications prises (presque tous les patients prenaient une benzodiazépine comme somnifère) et le fait que les chutes soient exclues à l'anamnèse sont des facteurs à prendre en compte dans l'interprétation des observations. Enfin, l'état de fatigue du patient qui participe à une évaluation pendant une demi-journée a certainement une influence. À ce stade, il nous semble donc essentiel de poursuivre ce travail afin de confirmer les données obtenues sur un échantillon plus important (notamment concernant la population MA). De plus, il serait utile de déterminer des valeurs de référence pour les variables Locométrix®, sur une population de sujets sains et de différentes tranches d'âge afin de mieux identifier les modifications strictement dues au statu cognitif des étudiés.

2.6. Conclusions

Le but de cette étude préliminaire était d'abord de déterminer les capacités offertes par un nouvel outil d'évaluation des paramètres de la marche, le Locométrix®. Nous démontrons clairement que le Locométrix® étudie avantageusement les paramètres de la marche des personnes âgées par rapport à

d'autres tests cliniques validés (*one-leg balance*, *timed up and go*, *pull test*). En effet, utilisé conjointement en TS et en TD, le Locométrix® permet de définir un profil de marche bien spécifique à chaque profil cognitif. De plus, la vitesse de marche et la fréquence des pas semblent positivement corrélées au niveau cognitif global du sujet. De même, la fréquence des pas semble également liée au niveau d'initiation présenté par le sujet. Enfin, la poursuite de ce travail devrait nous permettre d'identifier et de confirmer l'existence d'un ou plusieurs paramètres de la marche des sujets MCI comme facteurs cliniques prédictifs de l'évolution de ces sujets vers un syndrome démentiel.

References

[1] Aggarwal NT, Wilson RS, Beck TL, Bienas DA, Bennett. Motor dysfunction in mild cognitive impairment and the risk of incident Alzheimer disease. Arch Neurol 2006;63:1763–9.

[2] Allali G, Kressig RW, Assal F, Hermanne FR, Dubost V, Beauchet O. Changes in gait while backward counting in demented older adults with frontal lobe dysfunction. Gait Posture 2007;26:572–6.

[3] Ann Hunt L, Kapil Sethi D. The pull test: a history. Mov Disord 2006;21:894–9.

[4] Auvinet B, Bileckot R, Alix AS, Chaleil D, Barrey E. Les troubles de la marche chez le patient atteint de fibromyalgie. Rev Rhum 2006;73:932–5.

[5] Auvinet. et al. Chute de la personne âgée : de la nécessité d'un travail en réseau. Rev Med Assoc Malad 2002;33:183–91.

[6] Badeley MA, et al. Attentional control in alzheimer's disease. Brain 2001;124:1492–508.

[7] Beauchet O, et al. Élaboration d'un test clinique de marche du sujet âgé fragile à partir d'une approche cognitive de la locomotion. Ann Readapt Med Phys 2002;45:123–30.

[8] Beauchet O, et al. Influence of a specific cognitive task on spatial-temporal walking parameters in elderly frail individuals. Presse Med 2002;31:1117–22.

[9] Beauchet O, Allali G, Assal F, Berrut G. Troubles de la marche et maladie d'Alzheimer : revue systématique de la littérature. Rev Neurol 2007;163:11pt2. 4S80-4S114.

[10] Beauchet O, Dubost V, Aminian K, Gonthier R, Kressig RW. Dual-task-related gait changes in the elderly: does the type of cognitive task matter? J Mot Behav 2005;37:459–64.

[11] Beauchet O, Dubost V, Gonthier R, Kressig RW. Dual-task-related gait changes in transitionally frail older adults: the type of the walking-associated. Gerontology 2005;51:48–52.

[12] Blanke DJ. Comparison of gait of young men and elderly men. Arch Phys Med Rehabil 1989;69:144–8.

[13] Cocchini G, Della Sala S, Logie RH, Pagani R, Sacco L, Spinnler H. Dual task effects of walking when talking in Alzheimer's disease. Rev Neurol 2004;160:74–80.

[14] Cornette P, Swine C, Malhomme B, Gillet JB, Meert Ph, D'Hoore W. Early evaluation of the risk of functional decline following hospitalization of older patients: development of a predictif tool. Eur J Public Health 2005;16:203–8.

[15] Croisier JL, Crielaard JM, Chapelier D, Bouqueneau A, Maquet D. Gait analysis during an original walking test [Poster présenté au deuxième congrès international « gait and mental function » les 1–3 février 2008 par le département des sciences de la motricité de l'université de Liège].

[16] Dubost V, Kressing RW, Gonthier R, Hermann FR. Relationships between dual-task related changes in stride velocity and stride time variability in healthy older maults. Hum Mov Sci 2006;25:372–82.

[17] Échelle visuelle analogique, Adresse du site Internet : http://www.sniil.fr/html/DSI_echelle_eva.htm.

[18] Enzenberger. et al. Metronome therapy in patients with Parkinson disease. Nervenartzt 1997;68:912–7.

[19] Folstein MF, Folstein SE, McHugh PR. Mini-Mental State: a practical method for grading the state of patients for the clinician. J Psychiatr Res 1975;12:189–98.

[20] Franssen. et al. Equilibrum and limb coordination in mild cognitif impairment and mild Alzheimer's disease. J Am Geriatr Soc 1999.

[21] Hausdorff JM. Stride variability beyond length and frequency. Gait Posture 2004;20:304.

[22] Hittmair-Delazer M, et al. Concepts and facts in calculation. Brain 1994;117:715–28.

[23] Hollman JH, Kovash FM, Kubik JJ, Linbo RA. Age-related differences in spatiotemporal makers of gait stability during dual task walking. Gait Posture 2007;261:113–9.

[24] INRA. Locométrix®. Quand les recherches de l'INRA profitent à la médecine, 2003. Adresse du site Internet : http://www.inra.fr/presse/locometrix_r_quand_les_recherches_de_l_inra_profitent_a_la_medecine.

[25] Jacquot. et al. La chute de la personne âgée. Paris: Édition Masson; 1999.

[26] Katz S, Ford AB, Moskovitz RW, Jackson BA, Jaffee MW. Studies of illness in the aged: the index of MAL: a standardized measure of biological and psychosocial function. JAMA 1963;185:914–9.

[27] Lajoie Y, et al. Upright standing and gait: are these changes in attentional requirements related to normal aging? Exp Aging Res 1996;22:185–98.

[28] Lawton MP, Brody EM. Assessment of older people: self-maintaining and instrumental activities of daily living. Gerontologist 1969;9:179–86.

[29] Maki BE, Zecevic A, Bateni H. Cognitive demandes of executing postural reactions: does aging impede attention switching. Cogn Neurosci Neuropsychol 2001;12–6.

[30] Marsh AP, Geel SE. The effect of age on the attentional demands of postural control. Gait Posture 2000;12:105–13.

[31] Mattis S. Mental status examination for organic mentalsyndrome in the elderly patient. In: Bellak L, Karasu TE, editors. Geriatric psychiatry. New York: Grune and Stratton; 1976. p. 77–121.

[32] Mathias S, Nayak USL, Isaacs B. Balance in elderly patients: the "get up and go" test. Arch Phys Med Rehabil 1986;67:387–9.

[33] McIntosh. et al. Rhythmic auditory-motor facilitation of gaits patterns in patients with Parkinson's disease. J Neurol Neurosurg Psychiatry 1977;62:22–6.

[34] McKhann G, Drachman D, Folstein M, et al. Clinical diagnosis of Alzheimer's disease: report of the NINCDS-ADRDA Work Group under the auspices of Department of health and human services task force on Alzheimer's disease. Neurology 1984;34:939–44.

[35] Melzer I, Benjuya N, Kaplanski J. Age-related changes of postural control: effect of cognitive tasks. Gerontology 2001;47:189–94.

[36] Morris JC. The Clinical Dementia Rating (CDR): Current version and scoring rules. Neurology;43:2412–4.

[37] Munhoz RP, Li JY, Kurtineq M, Piboolnurak P, Constantino A, Fahn S, et al. Evaluation of the pull test technique in assessing postural instability in Parkinson's disease. Neurology 2004;13:125–7.

[38] Nakamura T, et al. Relationship between falls and stride length variability in senile dementia of the Alzheimer type. Gerontology 1996;42:108–13.

[39] Nigg BM, et al. Gait characteristics of the elderly. Clin Biomech 1983;3:79–87.

[40] Pelissier. La marche humaine et sa pathologie. Paris: Édition Masson; 1997.

[41] Petersen RC, Doody R, Kurz A, Mohs RC, Morris JC, Rabins PV, et al. Current concepts in mild cognitive impairment. Arch Neurol 2001;58:1985–92.

[42] Pettersson AF, Olsson E, Wahlund EO. Motor function in subjects with mild cognitive impairment and early Alzheimer disease. Dement Geriatr Cogn Disord 2005;19:299–304.

[43] Podsiadlo D, Richardson S. The timed « up and go »: a test of basic functional mobility for frail elderly persons. J Am Geriatr Soc 2000;48:142–8.

[44] Podsiadlo D, et al. The timed « up and go »: a test of basic functional mobility for frail elderly persons. J Am Geriatr Soc 1991;39:142–8.

[45] Rey A, Osterrieth PA, Taylor LB. Rey Osterrieth complex figure test and Taylor version. In: Spreen O, Strauss E, editors. A compendium of neuropsychological tests: administration, norms, and commentary. New York: Oxford University Press; 1991. p. 157–67.

[46] Rubenstein LZ, Harker JO, Salva A, Guigoz Y, Vellas B. Screening for undernutrition in geriatric practice: developing the short-fom mini nutritional assessment (MNA-SF). J Gerontol 2001;56A:M366–377.

14. Presentations related to this topic

1. Oral presentations

DISCERNER LES SUJETS ÂGÉS À RISQUE DE CHUTE: APPORTS DU DATA MINING. Presented at the Congrès International Francophone de Gériatrie et Gérontologie, Montreux, Suisse, 2018.

THE GAIT PATTERN OF HEALTHY OLD PEOPLE FOR FAST WALKING CONDITION. Presented at the European Union Geriatric Medicine Society Congress, Lisboa, Portugal, 2016.

Short oral presentation based on E-poster: TO ASSESS GAIT PATTERN OF HEALTHY OLD PEOPLE FOR COMFORTABLE WALKING CONDITION. Presented at the International Society of Gerontechnology Congress, Nice, France, 2016.

Short oral presentation based on E-poster: TO ASSESS GAIT PATTERN OF HEALTHY OLD PEOPLE FOR DUAL TASK WALKING CONDITION. Presented at the International Society of Gerontechnology Congress, Nice, France, 2016.

APPORTS DE NOUVELLES TECHNIQUES D'ÉTUDE DE LA MARCHE DE NOS ÂGÉS, Presented at the Congrès International Francophone de Gériatrie et Gérontologie, Nice, France, 2010.

LOCOMETRIX: DIFFÉRENTES VITESSES POUR DIFFÉRENTS PROFILS ? Presented at the francophone annual congress of the Belgian Society of Geriatric, Liège, Belgium, 2010.

THE INFLUENCE OF COGNITIVE DECLINE ON MOTOR PERFORMANCE IN ELDERLY PATIENTS, Presented at the Congres Kinesitherapie En Ergotherapie In De Geriatrie, organized by the Dutch University of Brussels (VUB), Brussels, Belgium, 2009.

LE LOCOMETRIX, UN TEST DE MARCHE PERFORMANT? Presented at the francophone annual congress of the Belgian Society of Geriatric, Liège, Belgium, 2008.

2. *Poster format presentations*

GILLAIN S., Wojtasik V., Dardenne N., Croisier J-L., Brüls O., Bruyère O., Salmon E., Garraux G., Petermans J. PHYSICAL AND MENTAL DETERMINANTS RELATED TO THE FIRST FALL IN HEALTHY OLD PEOPLE. Poster format presented at European Union Geriatric Medicine Society (EUGMS) Congress, Nice, France, 2017.

GILLAIN S., Boutaayamou M., Schwartz C., Demonceau M., Dardenne N., Croisier J-L., Brüls O., Bruyère O., Salmon E., Garraux G., Petermans J. THE GAIT PATTERN OF HEALTHY OLD PEOPLE FOR DUAL TASK WALKING, Poster format presentend at EUGMS Congress, Lisboa, Portugal, 2016.

GILLAIN S., Wojtasik V. Depierreux F. , Schwartz C., Boutaayamou M., Demonceau M., Schmitz X., Dardenne N., Bruyère O., Garraux G., Petermans J. PHYSICAL AND MENTAL DETERMINANTS OF FALLS IN HEALTHY OLD PEOPLE: BASELINE DATE OF THE GABI STUDY. Poster format presented at European Union Geriatric Medicine Society (EUGMS) Congress, Oslo, Norway, 2015.

GILLAIN S., Petermans J., Dardenne N., Beaudart C., Buckinx F., Garraux G., Reginster J.-Y., Bruyère O., Demonceau M., Wojtasik V., Schwartz, C. (2015). COMPARISON OF BODY COMPOSITION, MUSCLE FORCE AND PHYSICAL PERFORMANCES BETWEEN FALLER AND NON-FALLER PEOPLE INCLUDED IN A COHORT OF 100 COMMUNITY DWELLING VOLUNTEERS: THE GABI STUDY. Poster format presented at International Conference in Frailty and Sarcopenia Research (ICFSR), Boston, U.S.A., 2015.

GILLAIN S., Schwartz C., Boutaayamou M., Demonceau, M., Croisier J.-L., Brüls O., Garraux G., Reginster J.-Y., Petermans J. VALIDATION DES PARAMÈTRES DE MARCHE PAR UN SYSTÈME ACCÉLÉROMÉTRIQUE À L'AIDE D'UN SYSTÈME OPTO-ÉLECTRONIQUE 3D. Poster session presented at francophone annual congress of the Belgian Society of Geriatric, Liège, Belgique, 2014.

GILLAIN S., Dramé M., Ricour C., Lekeu F., Wojtasik V., Salmon E., Petermans J. GAIT ANALYSIS AND PREDICTION OF THE CONVERSION OF MILD COGNITIVE IMPAIRMENT SUBJECTS (MCI) INTO ALZHEIMER'S DISEASE (AD). Poster format presented at IX Congress of EUGMS, Venice, Italy, 2013.

GILLAIN S, Lekeu F, Wojtasik V, Salmon E., Petermans J. ARE THE MCI THE MOST REGULAR WALKERS? Poster format presented at IX Congress of EUGMS, Brussels, Belgium, 2012.

GILLAIN S, Warzee E, Lekeu F, Wojtasik V, Maquet D, Croisier J-L, Salmon E, Petermans J., INTEREST OF LOCOMETRIX TO ASSESS GAIT'S PROFILE IN SPECIFIC OLD POPULATIONS,

Poster format presented at International Association of Gerontology and Geriatrics Congress, Paris, France, 2009.

GILLAIN S, Lekeu F, Wojtasik V, Gachet C, Maquet D, Croisier J-L, Salmon E, Petermans J, IS THERE AN INTEREST TO DETERMINE THE GAIT'S PROFILE OF MCI SUBJECTS TO PREDICT THE RISK OF ALZHEIMER DISEASE? GaitRite Meeting, Basel, Swiss, 2009.