

ON THE PERIODONTAL LIGAMENT REPRESENTATION IN ORTHODONTIC TOOTH MOVEMENT MODELISATION.

¹ Marlène Mengoni and ¹ Jean-Philippe Ponthot

¹Non Linear Computational Mechanics, Department of Aerospace and Mechanics, University of Liège, Belgium

SUMMARY

Orthodontic tooth movement (OTM) is the result of bone remodeling at the interface with the periodontal ligament (PDL) around a mechanically loaded tooth in response to a biomechanical stimulus. Modeling of the PDL therefore plays an important role in the process of modeling OTM. However when producing a finite element model from clinical computer tomography data, the PDL cannot be segmented and its geometry is approximated by many authors from the root geometry.

The aim of this study is to propose alternatives to a geometrical representation of the PDL using either simple spring elements between the teeth and alveolar bone or bilateral sticking contact conditions. Results consist in a comparison of the hydrostatic and Von-Mises stresses in the bone along the root as well as the strain energy used in a bone remodeling algorithm when a 1N force is applied to a single rooted tooth crown.

While both models can well represent the pressure (hydrostatic stress) transfer from the tooth to the bone, the bilateral sticking contact conditions show better results to transfer the shear stress as well as the strain energy.

INTRODUCTION

For the past few years, 3D finite element (FE) models based on computer tomography (CT) scans are increasingly used in the field of orthodontics [1,2,3 among many others]. This shows great improvement from the previous decades where most models were derived from 2D then 3D standard geometries.

The production of FE meshes from CT data for orthodontic tooth movement (OTM) modeling requires a description of several important tissues to be segmented. From this segmentation, a surface reconstruction and triangulation followed by a volumetric meshing technique produces the required FE model. However, clinical CT resolution allows only for differentiation of bone (both cortical and alveolar) and teeth (distinction of dentin and enamel). Specially, the surface geometry of the periodontal ligament (PDL) cannot be directly derived from CT images. One of the solutions is to derive a model from μ CT data. However, μ CT technology is not available on a clinical basis, a clinical tool therefore has to be developed. In most recent studies, the PDL is generated using scaling and/or Boolean operations on the teeth and bone interface in order to obtain a thin enclosure [2,3,4]. This approximation is performed despite the fact that most authors agree on the importance of geometrical and material properties of the PDL in the achievement of OTM.

The aim of this study is therefore to propose alternative methods to account for the mechanical role of the PDL without geometrically representing its thickness.

METHODS

A mandible geometry was obtained from the INRIA/GAMMA repository [5], consisting of a surface reconstruction of the mandibular bone and its 14 teeth (crown and root). These surfaces are typically the output of a CT data segmentation and triangulation.

The 2D outline in the mesiodistal plane of the left central incisor was extracted (figure 2) and meshed. From this geometry, four FE models were created ; one for an actual PDL creation (the reference model), one for a spring representation of the PDL, one for a contact representation and a final one with no PDL. The reference model required to duplicate the nodes and meshed curves at the interface and move them normally to the surface to create an enclosure of 0.2 mm thickness for the PDL. The second one required to duplicate the nodes and meshed curves at the interface and create spring elements between the nodes at the same position. The contact model required only duplication of the nodes and meshed curves. The fourth model did not require any pre-processing operations as the tooth and bone were supposed to be bounded. The two (three in the reference model) surfaces (bone and tooth) were finally meshed with linear quadrangles. Material behavior were assumed to be linear (see Table 1), bone Young's modulus depending on the bone apparent density. For the spring model, the spring stiffness takes into account the PDL Young's modulus and the distance between consecutive springs. Some non linearity was to be accounted for as the springs initial length is zero while the PDL thickness is 0.2 mm. For the contact model, bilateral sticking contact conditions allowed for a penetration of half the thickness of the PDL into each surface and a penalty factor accounting for the PDL Young's modulus is used.

Table 1 : Material parameters

	Young's modulus [MPa]	Poisson's ratio [-]
Bone	1700.	0.3
Tooth	20000.	0.3
PDL	0.6	0.45

A pressure representative of a 1N force was applied on the labial side of the tooth crown and the basal line of the bone was considered fixed. As this study is a comparative study for the PDL mechanical behavior, only the tooth initial movement is accounted for, no remodeling algorithm is present. FE analyses were performed using Metafor, a in-house object-oriented non-linear finite element code [6].

RESULTS AND DISCUSSION

The results for the tipping movement simulations of the tooth initial mobility produced by the 1N force are presented in figures 1 and 2. For the three first simulations, the center of rotation is situated about at one third of the root length

from the apex as expected from a tipping movement on a single rooted tooth.

The stress distribution in the bone along the root for all models shows (figure 1 – plain line is the reference model) that the mechanical role of the PDL is of major importance as in the fourth model (no PDL – dashed line) both the stress intensity (hydrostatic and Von-Mises stress) and its distribution are poorly represented.

Models with springs or bilateral contact can both fit the hydrostatic stress distribution. Both these models ensure the transfer of the pressure through the ligament with the same intensity as for the reference model. However, the spring model (dotted line) shows shear intensity half the reference shear on the labial side (side in compression). On the lingual side (side in tension), the shear intensity of the spring model is up to 19% higher than the reference one and its maximum position is 50% less apical than the reference maximal position. The contact model (dashed-dotted line) shows a shear intensity 17% lower in compression and 16% higher in tension with the same position for the maximal value of shear. As the strain energy used for a remodeling algorithm depends on both the shear stress and the hydrostatic pressure, the discussion is the same as for the shear stress.

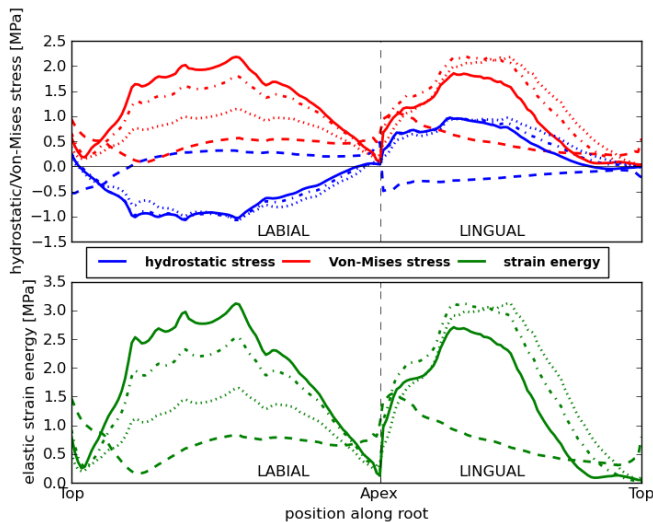


Figure 1: Stress and strain energy in the bone along the root. Plain lines are obtained for a solid PDL, dotted lines for springs, dashed-dotted lines for a bilateral contact condition and dashed lines for no PDL representation.

The obtained displacement can be observed in figure 2 (linguo-labial displacement) for the bilateral contact model. It shows a maximal displacement of the root in the labial side of 36 μm at the apex. This displacement is due only to PDL compression. The close up at the apex shows a penetration of the tooth surface into the bone representative of the PDL deformation in compression. At the collar, the tooth unties from the bone while the PDL is widening.

CONCLUSIONS

Clinical CT data precision does not allow for the PDL surface reconstruction. Extensive preprocessing is often used to create a PDL. This study demonstrated the potential of using customized contact conditions on the bone/tooth interface, as both the hydrostatic and shear stress in the bone could be represented while reducing the preprocessing of the model. In the pre-processing steps, it should also be noticed

that duplication of the meshed nodes (and not only the geometric points) is required for the spring model as compatible meshes of the contour are needed. This therefore allows for a lower number of duplications (here only 23 points represent the interface while 171 meshed nodes are needed). The contact model can therefore also be used if the bone and teeth have been triangulated separately, creating non-compatible meshes.

Further work should customize the contact behavior to retrieve a better shear intensity. We should also consider non linearity of the PDL behavior by using contact laws such as Coulomb's law and extend the methodology to 3D geometries.

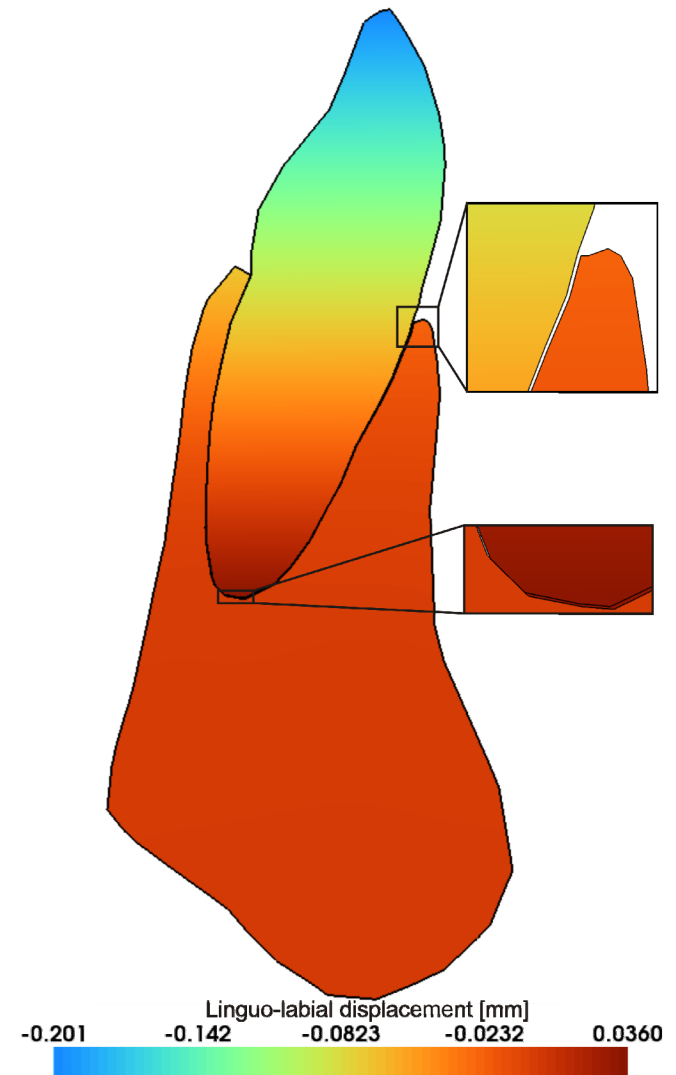


Figure 2: Displacement in the linguo-labial direction for a bilateral contact condition representing the PDL. Close up on the tooth/bone contact at the labial collar and the apex.

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