A DAMAGE/REPAIR MODEL FOR ALVEOLAR BONE REMODELING

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1. ABSTRACT

Tooth movements obtained through orthodontic appliances result from a complex biochemical process of bone structure and density adaptation to its mechanical environment, called bone remodeling. This process is mechanically far from linear elasticity. It leads to permanent deformation due to biochemical actions. The proposed biomechanical constitutive law (inspired from Doblaré's model [1]) is based on a Continuum Damage Mechanics material coupled to an elasto-visco-plastic law and considering a damage variable not as actual damage but as a measure of bone density. It is formulated as to be used explicitly for alveolar bone, whose remodeling cells, opposite to most bones, are believed to be triggered by the pressure state of the bone matrix.

2. INTRODUCTION

For most bones [2, 3], the physiological remodeling process takes place in order to adjust the amount of tissue and its topology according to long term loading conditions, following what is called ``Wolff's law'' of bone adaptation. The bone therefore adapts its density in such a way to achieve an homeostatic state of stresses. Besides the density change, remodeling also occurs to change the bone topology, mainly in trabecular bone for which the trabeculae align to the stresses principal directions. Contrary to the majority of bones, alveolar bone remodeling seems on a macroscopic scale to depend mainly on the pressure state [4, 5, 6]. One can observe apposition on the tension side of a tooth when loaded with an abnormal mechanical environment, such as the one obtained with orthodontics appliances, as well as resorption on the compression side. The actual mechanical stimulus for such a difference is not quite clear and uniform among biology and biochemistry literature (see among others [3, 7, 8]). Some works focus on the periodontal ligament (PdL), a complex membrane of high vascularity seated between the teeth and the bone, non linear response (see among others [4, 9, 10, 11]). Its non linearity and different behavior in traction and compression leads to opposite loading conditions of the bone on each side of the tooth. However, when no non-linearities are considered in the PdL, no difference in the stress level can be observed. Instead of focusing on the PdL response, this work concentrates at first on the bone behavior during remodeling. We suppose the pressure state (traction or compression) of the bone matrix as the key stimulus to differentiate apposition and resorption in overloaded conditions.

Within the diverse approaches that have been adopted to model bone remodeling processes, most of them are qualified as phenomenological models. They are models that do not try to predict the evolution of the microstructure and biological constitution

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of the bone as consequence of the mechanical environment but whose goal is to predict the mechanical behavior of bone, taking into account the acting loads, its microstructure and the constraints imposed by other organs. Most of these models admit the existence of a given mechanical stimulus that produces bone apposition or resorption in such a way that the stimulus tends to a certain physiological level in the long-term. The model which is proposed in this work is built on a damage/repair based model stated first by Doblaré and co-workers [1, 12]. This model as been chosen as a working base because it is one of the few models whose stimulus variation is justified through thermodynamical concepts of continuum mechanics.

3. MATERIAL AND METHODS

Doblaré and co-workers adapted Stanford's remodeling model [13, 14] so as to formulate it in a Continuum Damage Mechanics approach. They consider an anisotropic damage coupled to a linear elastic material (reduced to isotropy, the damage variable can be written: $d=1-E/E_0$). The undamaged material is considered as the ideal situation of bone with null porosity and perfect isotropy. The process of bone resorption increases the damage while bone apposition reduces damage (repair process). In analogy to plasticity, a remodeling stimulus is identified with the variable thermodynamically associated with damage. Doblaré and co-workers therefore establish two damage criteria (Eq. (1) in their isotropic formulation), representing the domain of the remodeling stimulus for which damage is not modified (the lazy zone) both for resorption and formation.

$$\begin{aligned} g_f &= U-(1+\omega)U \\ g_r &= \frac{1}{U}-\frac{1}{(1-\omega)U} \end{aligned} \qquad (1)$$

In (1), U is function of a strain energy density as well as the density and the number of cycles considered for the applied loads and U^* is a reference homeostatic value of U. Using consistency condition, they can explicit the damage variation, Eq. (2), as proportional to a remodeling rate such as the one proposed by the Stanford group and presented on Figure 1 and to a specific surface, function of the bone's density.

$$\dot{d} \propto k\, S_v\, \dot{r} \qquad (2)$$

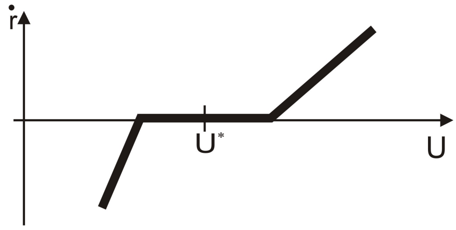

Figure 1 : Stanford remodeling rate

As stated previously, this type of bone remodeling model cannot be applied to alveolar bone if no PdL nonlinearities are considered. We propose a model that uses the same approach of Doblaré's, both for the damage definition and variation and the damage criteria. However, in accordance to the observation of a pressure dependent phenomenon, the remodeling rate definition is modified (Figure 2, Eq. (3)) taking into account the pressure state.

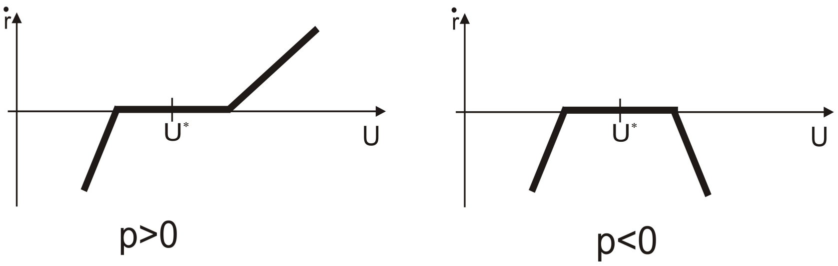

Figure 2 : Pressure dependent model.

$$\dot{r} = \begin{cases} c_f \dfrac{g_f}{\rho^{2-\beta/2}} & if \;\; g_f > 0 \; and \; p > 0 \\ -c_r \dfrac{g_f}{\rho^{2-\beta/2}} & if \;\; g_f > 0 \; and \; p < 0 \\ -c_r \dfrac{g_r}{\rho^{2-\beta/2}} & if \;\; g_r > 0 \\ 0 & if \;\; g_r < 0 \; and \; g_f < 0 \end{cases} \qquad (3)$$

In (3), c_r and c_f are two remodeling constants, β is such that $E = B\rho^{\beta}$ (E being Young's modulus) and p is the pressure, positive in traction.

The damage used is virtual and is a measure of bone density. There is no actual damage of the tissue. The damage evolution is proportional to the defined remodeling rate so that repair will occur in the case of tissue formation, for overloaded traction conditions. Damage will increase in the case of tissue resorption, both in the case of overloaded compression conditions and underloaded conditions. Opposite to Doblaré's approach, this continuum damage model is coupled to a visco-plastic material behavior. One can therefore consider permanent deformation of the bone as a process due to both permanent strains (and therefore plasticity-like, although it is clear that the relevant inelastic process is different from that of the classical metal plasticity) and change in bone density. It can also be used to describe a fracture process with a plasticity-like yield function to model the envelope of bone failure.

Once the remodeling model has been formulated, we need to check its ability to achieve qualitative results close to the ones obtained in experimental tests of actual alveolar bone. This is accomplished in the next section in which the model is applied to the study of the remodeling behavior of the alveolar bone in the case of orthodontics treatment.

4. RESULTS

In this section, we consider the potential of this pressure dependent model to predict the density evolution of bone tissue. As an example, we present a 2D model of a tooth surrounded by its parodontal tissue, on the crown of which a pressure load is applied in the vestibulo-lingual direction. The aim is to predict the bone density and its evolution from an initial ideal situation (isotropic material with uniform density distribution) when loaded by forces that characterize the orthodontics appliances. Neither this problem nor the starting situation are "real" problems, therefore, the homeostatic values are not relevant.

The tooth geometry is idealized, with a parabolic root surrounded by a constant thickness periodontal ligament as well as trabecular and cortical bone. The 2D discretization used here is shown in Figure 3. The root is 12 mm high and 6 mm wide at the collar. The tooth and the PdL mechanical behavior is elastic. The cortical layer as well as the trabecular bone mechanical behavior is elasto-plastic with a continuum damage model. The damage evolution follows the remodeling law proposed in this work.

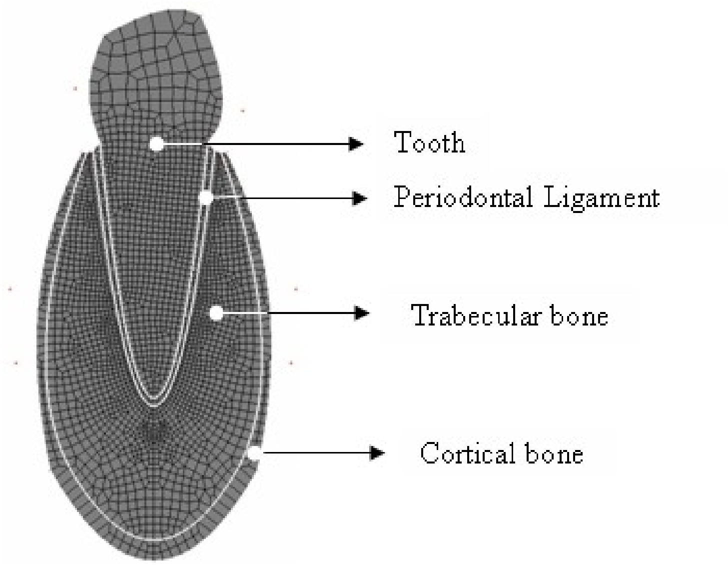

Figure 3 : Geometry and mesh $E_{tooth} \approx 20GPa, \nu_{tooth} = 0.3, E_{pdl} = 0.6MPa, \nu_{pdl} = 0.45$, $\rho_{cortical} = 1.99g/cc, \rho_{trabecular} = 1.15g/cc$

Finite Element analysis is performed, using finite strains code METAFOR, considering a plane strain state as well as a quasi-static analysis. The basal bone junction is restrained in both the vertical and horizontal directions. The mesh is composed of 2670 nodes and 2600 linear quadrangular elements. Loading is performed using two levels of pressure value (corresponding to 1.0N and 2.0N force) as well as different sets of remodeling constants (Table 1).

Force	$c_r = c_f$ $[\mu m/day]$	c_r , c_f $[\mu m/day]$
1.0N	2.0	1.0, 4.5
2.0N	1.0	0.5, 2.25

Table 1 : Remodeling constants used for trabecular bone.

The obtained tooth movement is a rotation around a center of rotation situated at one third of the root length starting from the apex for small loads (center of rotation is getting apical while load increases). The rotation angle is almost null for small forces, corresponding to initial tooth mobility, while it reaches about 1.5° for higher forces. The initial tooth mobility depends only on the applied load and not on the remodeling model used. However the possibilities of long term tooth movement due to bone density change varies strongly with the type of model used (1 constant or 2 constants model) as well as with the homeostatic value U^* and the number of cycle considered.

When using the same remodeling constant for both resorption and formation, one does not exactly gets symmetric values for damage variation (Figure 4, top row) because of its dependence on the damage value (damage variation increases while damage decreases). In the case of two different remodeling constants (Figure 4, bottom row), the one used in resorption, c_r , is the restrictive one because it is the one increasing the damage value. The constant used in formation, c_f , can be increased almost at will (as numerical convergence is concerned, not on a biological point of view). However, if too high, the effective stiffness will increase and so will the stress. The stress state around the apex will be modified and affect the resorption side as well.

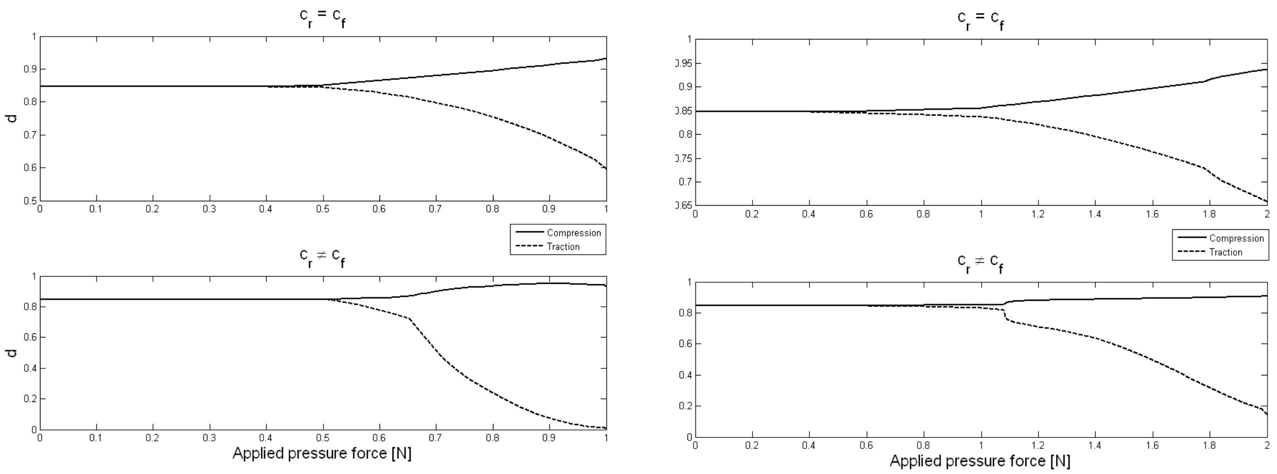

Figure 4 : Damage variation at the apex – top row: $c_r = c_f$, bottom row : $c_r \neq c_f$, left : low force, right: higher force, plain line : compression side, dashed line : traction side.

Concerning the stress state of the periodontal ligament, it is interesting to notice that the hydrostatic stresses are several times higher than the shear stresses (up to 8 times for low forces). For the lower load, the later ones vary from 0 to 0.025MPa along the tooth root (highest values reached at the collar) while the hydrostatic stresses' range goes from 0 to 0.2MPa both in compression and traction (highest values reached at the level of the center of rotation). The pressure is as expected the key stress in the PdL. Qualitatively analyzing the results of this model, as can be expected of an orthodontic treatment, the results show that while higher loads lead to higher initial displacement, density variation of the bone can be observed only in the apex region while it is observed on the entire length of the root for smaller loads.

5. DISCUSSION

The present study introduced a numerical model for the simulation of orthodontic tooth movement based on the assumption that bone remodeling processes during tooth movement are controlled by the elastic energy density as well as the pressure state of the alveolar bone. In spite of the necessary idealizations, a reliable qualitative prediction of bone density variation around the tooth is possible for porosities variations from 0% to almost 70%, starting from an homogenized porosity of 45%. Next investigations will have to focus on the functional description of the orthodontic “remodeling law”. So far anisotropic phenomena cannot be considered while bone's intrinsic anisotropy leads to directionality dependent remodeling processes. The simulation presented allows a prediction of tooth movement's initiation. It should be completed by long term tooth movement issues such as actual bone loss and creation with rupture-like criteria for loss or creation of elements.

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