

The role of periodontal ligament elasticity in periodontal changes – numerical simulation

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ABSTRACT

The clinical applications of oral mucosal biomechanics are particularly important in determining how stimulation can be performed for tissue remodeling, in understanding the level of pressure that can be applied to the painful threshold, in osseous resorption, and the availability of tissue movement.

This study shows, through the finite element, the way in which the periodontal ligament behaves at the level of all dental surfaces in nonlinear model and 3 levels of alveolar bone retraction (0 %, 50%, 75%). From the point of view of geometric construction, each model consists of 3 independent components: the alveolar bone, the periodontal ligament (PL) and the tooth.

The deformities on the lingual surface are very small compared to the other surfaces. The deformity increases progressively with the loss of the supporting bone. From a physiological point of view, the nonlinear model is much closer to reality because it simulates the reaction of the tissue to its immediate loading by a very low rigidity after which, due to the hydrostatic pressure created in the interstitial fluid, the value of tissue rigidity increases exponentially.

Keywords: periodontal ligament, finite element, elastic model, gingival recession, bone loss

INTRODUCTION

The periodontal ligament (PL) is a fibrous tissue, with thicknesses between 0.15 and 0.38 mm. The purpose of PL is to attach the dental root, specifically the root cementum, to the surrounding alveolar bone [1,2]. Its main role is to transmit the forces acting on the teeth to the alveolar bone [3,4].

The clinical applications of oral mucosal biomechanics are particularly important in determining how stimulation can be performed for tissue remodeling, in understanding the level of pressure that can be applied to the painful threshold, in bone resorption, and the availability of tissue movement. In the mastication process, the soft tissues adjacent to the teeth play a decisive role in the distribution of occlusal forces to the maxillary bone area. The distribution is in the form of a functional hydrostatic pressure that occurs in the soft tissues and deter-

mines the pressure centers according to the loading [5-7].

The compression of the mucosa leads to a reduction in blood flow within that area, leading to local pain and discomfort. Also, higher or prolonged compression for a longer period of time can lead to intracellular edema or cellular inflammation. If the hydrostatic pressure reaches a level that exceeds the capillary pressure, then the circulation is reduced, and in its absence bone resorption occurs, a phenomenon that is progressive and harmful to oral health [8,9]. This is due to a lack of essential nutrients.

The removal of mechanical stress generates an elastic recovery of the mucosa, especially in young patients. Reducing the surface pressure will allow the interstitial fluids to irrigate the area again, sometimes generating a higher flow pressure than the initial one. In the elderly, however, the phenom-

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enon does not occur in the same way due to the loss of elasticity of the mucosa as the person is aging. If the mechanical stress is significant and for a long time, even tissue ischemia may occur [10-13].

Studies in the biomechanics of the oral mucosa have shown that the oral mucosa is much more tolerant of cyclic stress than of continuous stress. Thus, the resorption threshold of the alveolar ridge was identified at 19.6 kPa for cyclic stresses and only 6.8 kPa for continuous stress [14,15,6].

These values are of reference, they are strongly influenced by the pathophysiological condition of the individual, to the same extent as the level of continuous pressure applied.

Elasticity is one of the fundamental properties that define the behavior of a material. This is quantified by the modulus of elasticity (Young's modulus) which shows the tendency of a body to deform proportionally with the applied stress. In the case of the mucosa, it is very deformable to compression, presenting modulus of elasticity in a very wide range [16]

Due to the inhomogeneity of the mucosa, its initial rigidity derives from two components: the epithelial layer, blood vessels and collagen fibers on the one hand and the fluids (blood, interstitial fluid) on the other.

MATERIAL AND METHOD

For geometric models, linear elastic materials were defined by their two properties (longitudinal modulus of elasticity and Poisson's ratio), based on the bibliographic study. These are presented in Table I, in this way the property being considered as homogeneously distributed in the model. Because the periodontal ligament (PL) is the key element of this study, the nonlinear material models were considered for it (figure 1). The nonlinear model materializes a nonlinear change of the modulus of elasticity from small to large, a change that takes place from a physiological point of view as the structure is loaded. Thus, at low stress values, the tissue reacts with a strong and less elastic deformation, after which the modulus of elasticity increases due to the hydrostatic pressure that is generated inside the tissue. Such a material model can be simulated as a

hyper elastic one, in this case being opted for a Mooney-Rivlin model defined by 5 parameters.

TABLE 1. Mechanical properties of materials used [14,17-21]

Material	Young Modulus [MPa]	Poisson's ratio[-]
Tooth	18600	0.31
Alveolar bone	13700	0.30
Tooth pulp	2.2	0.45
PDL (E <)	0.05	0.45
PDL (E >)	0.28	0.45
PDL (non-linear Mooney-Rivlin)		0.3

In PL compression it is described by a very low modulus of elasticity up to almost 100% deformation, after which the modulus increases exponentially about 1000 times. This increased value simulates the precontact between the tooth and the alveolar bone. In extension, the modulus of elasticity has a different variation, increasing from 0.05 MPa corresponding to a zero deformation, gradually, up to 50% deformation where it reaches a value 10 times higher. This deformation value in stretching is a breaking point of PL tissue fibers [22].

Poisson's coefficients indicating the PDL deformation ratio for each material model were extracted from the literature [19, 23-25].

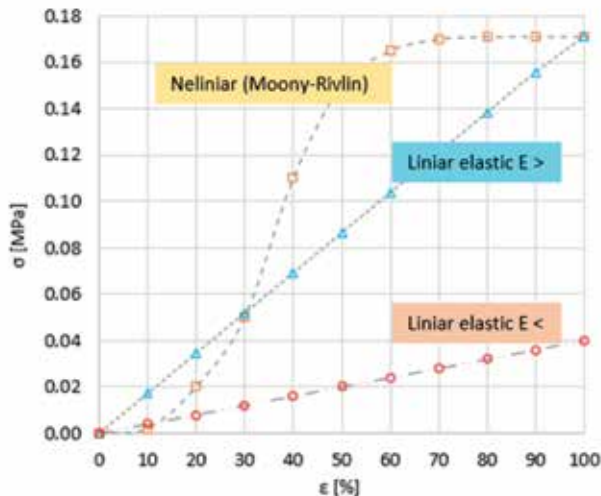


FIGURE 1. Constitutive models for PDL: Linear elastic with E <, linear elastic with E > and nonlinear Mooney-Rivlin

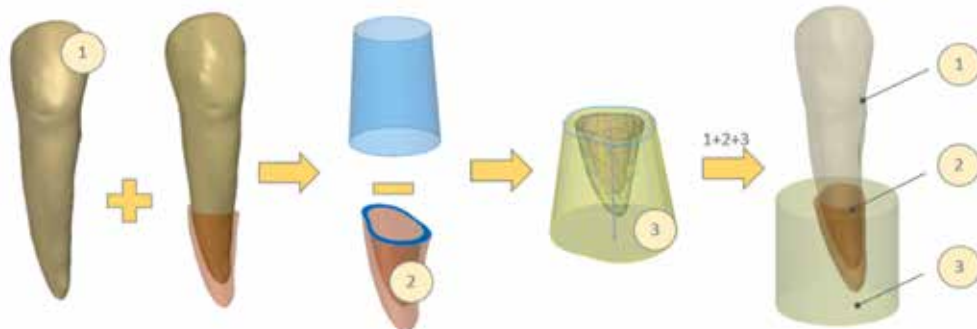


FIGURE 2. CAD operations for obtaining the analysis model

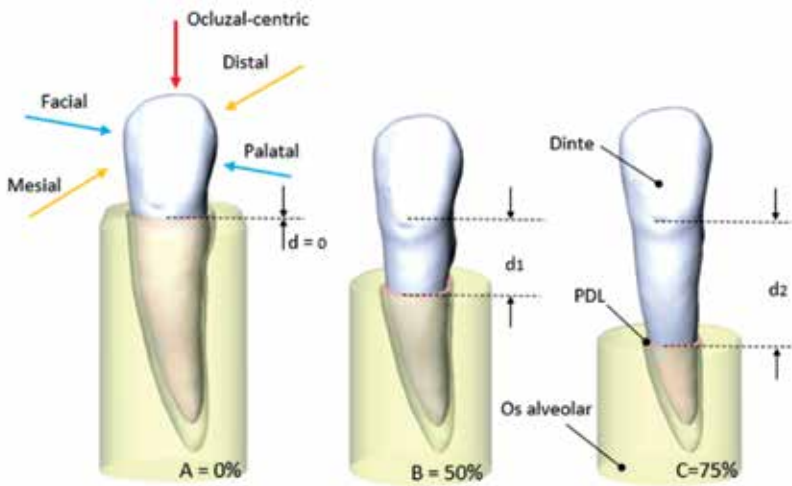


FIGURE 3. Analysis models: A-without bone retraction; B-with 50% bone retraction; C-with 75% bone retraction

The analysis was performed on 3 geometric models that materialize 3 levels of alveolar bone retraction. From the point of view of geometric construction, each model consists of 3 independent components: the alveolar bone, the (PL) and the tooth. The tooth is a geometric pattern reconstructed based on radiographic images to have a geometry as accurate as possible with the actual tissue. The PL ligament was constructed geometrically with the help of Boolean operations of addition and subtraction of reconstructed dental volume. This allowed a coincidental assembly between the tooth and the PL. The alveolar bone was constructed in the shape of a cylinder from which the volume corresponding to the ligament was extracted. Thus, 3 independent volumes 1, 2 and 3 (figure 2.) were obtained, which could be assembled by coincidence. In the same way we worked for all the components of the 3 analysis models: model A, B and C, the result being shown in figure 3. The differences in height of the alveolar bone and implicitly of the ligament were measured as a percentage from the top of it (gingival level). Also in figure 3 are shown the directions corresponding to the anatomy of the tooth, directions that will be referred to in the simulation.

The discretization of the structure was performed with automatic tetrahedral elements for the tooth and alveolar bone components. For PL, due to the special role that this component plays, the discretization was achieved with elements of constant size and value of 0.2 mm. This allows on the one hand a finer observation of the spectrum of stresses and strains by defining several control elements and nodes, and on the other hand facilitates the success of nonlinear analysis (figure 4).

RESULTS

PL ligament deformation is an indicator of how stress is transmitted from the tooth to the alveolar bone. The deformation is closely correlated with the tension in the ligament, respectively with the hydrostatic pressure that appears with the stress.

The results in **Figure 5** reveal the behavior of PL in case of lack of bone retraction for nonlinear material model. To see how the PL loading takes place, the results were presented according to the anatomical directions: distal, mesial, vestibular, and lingual.

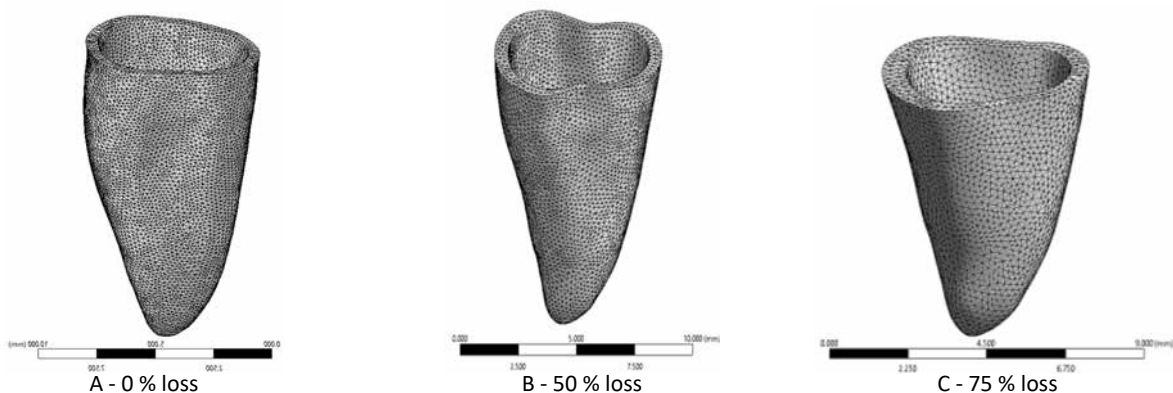


FIGURE 4. PDL discretization with constant tetrahedral elements of 0.2 mm

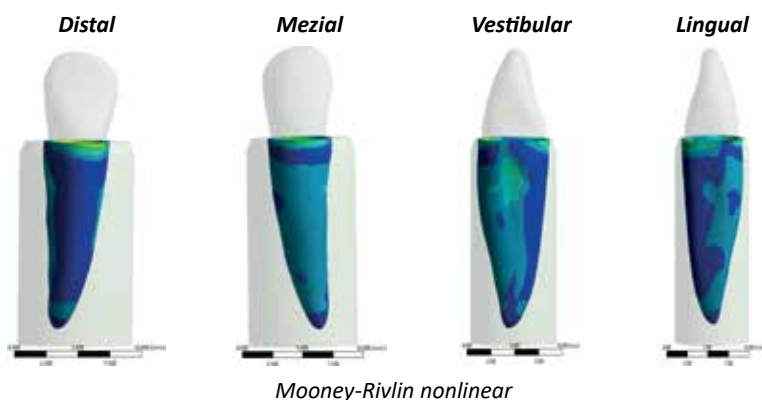


FIGURE 5. PL deformations model 0% loss

Due to the way in which the loading was applied, the results show a symmetry of the vestibular and oral surfaces, respectively an asymmetry of movement on the distal and mesial surfaces. Naturally, the most important displacement of the structure is in the area closest to the occlusal surface, which will dictate the position of the center of rotation of the tooth.

Figures 6 and 7 show the situations in which there is a retraction of the alveolar bone by 50 and 75% respectively from the complete anatomical position. This entails a corresponding retraction of the PL, i.e. a change in the distribution of stresses and displacements of the structure. Thus, considering the integral structure (0% loss) as a reference of the PL deformation, it can be seen that it increases greatly in the case of bone loss of 75%. The values obtained are significantly high and indicate a displacement of the tooth in relation to the alveolar bone, considered fixed. This displacement of the tooth is usually rotational. To determine the center of rotation of this movement it is necessary to identify the maximum and minimum values of stresses and deformations.

As a result of the change in the physical size of both the alveolar bone and the PL ligament, it is observed that the displacements of the structure are

increasing with the loss of bone (at the same load). This can be explained by a lever effect at which the fulcrum has changed. With this change, the value of the load together with the distance from the support point to the right support of the load produce a momentum of greater value, precisely due to the increase of this distance. Therefore, in order to have a mechanical balance at the level of the assembly, in the remaining contact points between the tooth/PL and the alveolar bone, reactions will be generated which in turn will produce a strong momentum of high value.

Figures 8, 9 and 10 show the results of sampling the voltage values in the nodes. For this, a number of 30 nodes were selected in the form of a curved line on the distal, mesial, vestibular and lingual surfaces of the PL ligament. The sampling line on each surface was drawn in such a way that it was always the median line of that surface. This was systematically repeated for the simulations of the 3 geometric models. Thus, the maximum and minimum aspects of the deformations can be observed, as well as the way in which the deformation transition is performed on the longitudinal direction of the tooth. It is found for the intact model (0% loss) how the minimum deformation is approximately in the middle of the median distance on any of the 4 surfaces stud-

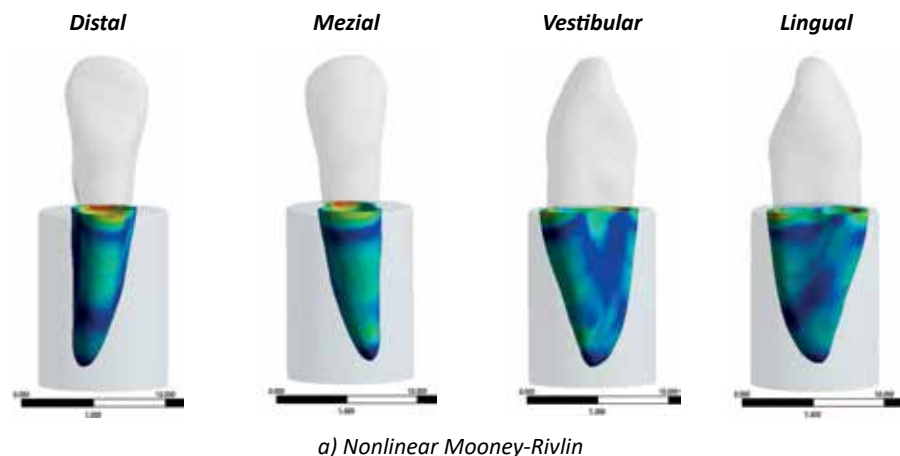
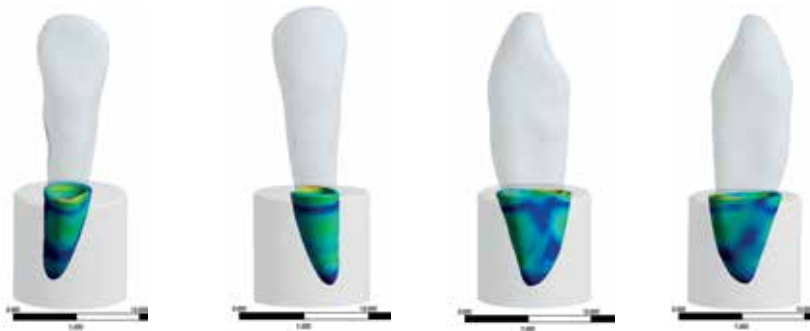


FIGURE 6. PL deformations model 50% loss



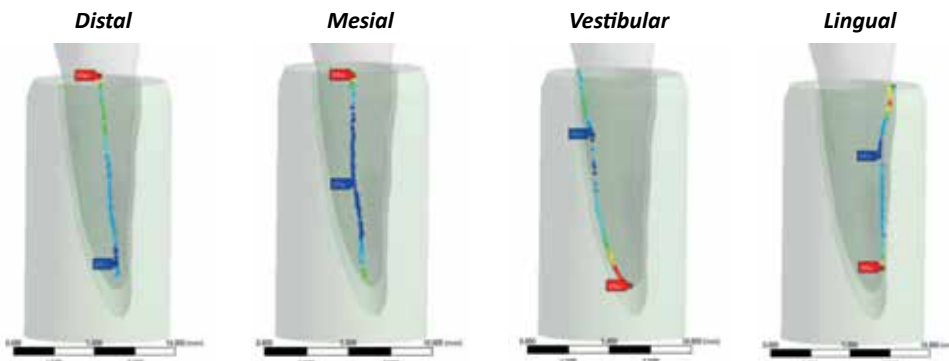
a) Nonlinear Mooney-Rivlin

FIGURE 7. PL deformations model 75% loss

ied. This directly indicates that the tooth is undergoing a rotational movement, with the center of rotation in the area of minimum displacement. Depending on the position of the center of rotation, there are also movements from the apex to the crown.

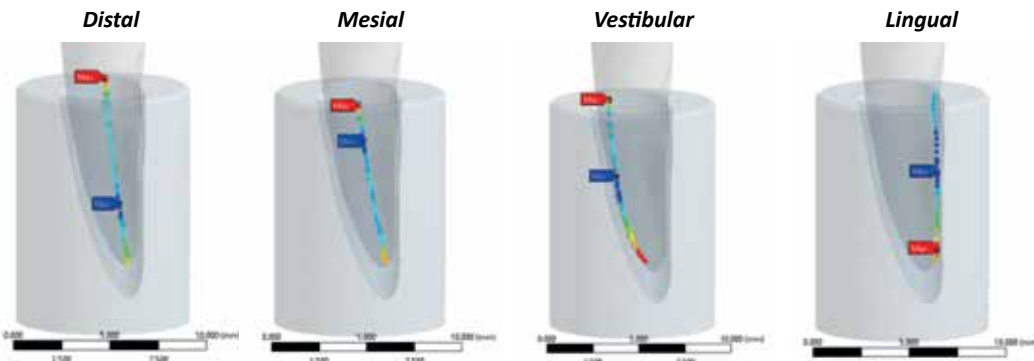
Maximum displacement is produced mainly in the area of the dental crown, on all 4 surfaces studied. The results appear consistently on the distal and mesial surfaces because they are planes of tension and compression in relation to the direction of stress. The other two surfaces, labial and lingual, are shear planes relative to the direction of loading.

Figures 11 (a,b,c) show the stress results at the upper extremity of the PL. Also, the shape and dimensions of these sections resulting from the conjugation of the alveolar bone with the tooth on the 3 levels explained above can be identified here. Considering the significantly higher dimensions of the upper surface of the PL in the case of the integral model compared to the dimensions of the models with bone loss, it can be understood why the value of stress and deformation increases with the bone loss, the stress being defined as the ratio between the value of the external stress and the area of the surface on which it is distributed.



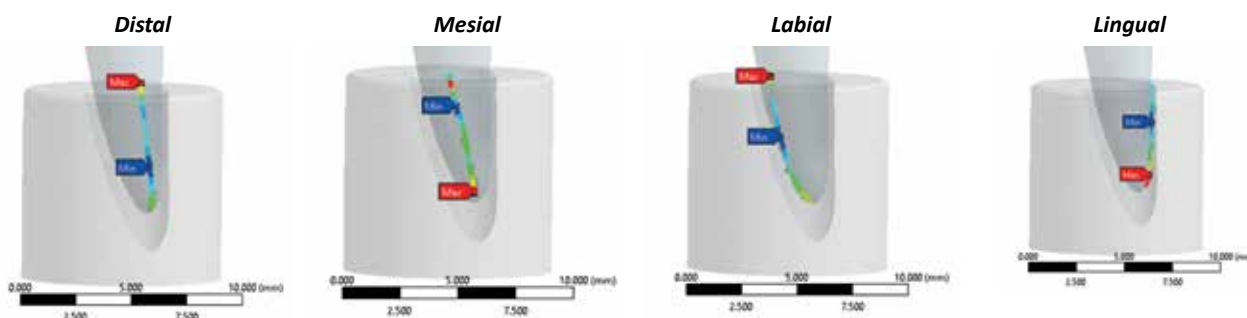
b) Mooney-Rivlin nonlinear

FIGURE 8. Sampling the displacement values from the PL for the 0% loss model



Mooney-Rivlin nonlinear

FIGURE 9. The displacement values from the PL for the 50% loss model



a) Mooney-Rivlin nonlinear

FIGURE 10. The displacement values from the PL for the 75% loss model

For the model *without bone loss*, the stresses appear almost evenly distributed on the upper surface of the PL. This is observed by the presence of the transition colors between the high voltage values (represented by warm colors: red-orange) and those of low value (represented here by cold colors: green-blue). The appearance of a sudden change from blue to orange is mainly found in the case of the 75% loss model and almost independent of the material model used. This sudden change in value and sign of tension occurs in the distal-mesial direction due to the fact that the stress has important components introduced in this direction (shear direction). Overall, the stresses in the PL surface adjacent to the dental crown are higher than those at the apex, due to the higher rotational displacement of this extremity.

According to the PL deformation analysis, the minimum and maximum values as well as their physical positions on the simulated tissue were extracted. The physical positioning of the maximum and minimum deformation is shown in Tables IV a,b,c. An important factor in the manifestation of these positions is the direction of the occlusal force. The decisive role in the occurrence of maximum and minimum in the simulation is played by the shape of the structure, which derives from both tooth anatomy and from simulating alveolar bone loss.

In the complete periodontium it is found that have maximum values in the mesial crown area, and the minimum values are located differently, at distal in 1/2 crown area.

In the case of a 50% gingival retraction, the changes are visible, the nonlinear model has the maximum distal crown and the minimum lingual crown.

Gingival recession of 75% shows a value of the maximal crown mesial and the minimum at the apex for the low elasticity model, the maximum distal-crown and the minimum mesial-crown for the linear model with high elasticity and the maximum crown-distal and the minimum at the apex for the nonlinear model.

The deformations inside the longitudinal sections of the PL are shown in Figures 12 a,b,c. Their appearance confirms the deformation distribution in the PL and also indicates the imbalances that occur with the stress and the change of geometry. Lower values of apex stresses and strains are recorded because the stress occurs in the form of rotation of the tooth in the support structure. Otherwise, at a pure compression occlusal stress, the apex of the structure will be most intensely stressed, due to its low volume compared to the rest of the model. In the present situation, the stress distribution to the area considered fixed is made towards the mesial surface and in the upper third of the model.

The center of rotation is in the area of minimum displacement (blue), being located in the lower half of the section.

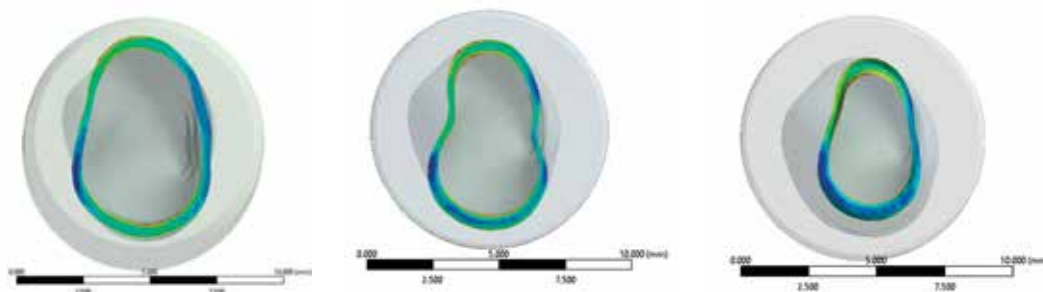


FIGURE 11. Ligament tensions in the border area of the dental crown, model with a. 0%; b. 50%; c 75%)

TABLE 2. The positions of min. and max. of equivalent deformation in PL a. 0% loss; b. 50% loss; c.75% loss

Value	a.		b.		c.	
	Nonlinear (Mooney-Rivlin)	Position	Nonlinear (Mooney-Rivlin)	Position	Nonlinear (Mooney-Rivlin)	Position
Max. [mm/mm]	$6.28 \cdot 10^{-2}$	mesial-crown	$1.19 \cdot 10^{-1}$	distal-vrown	$4.39 \cdot 10^{-1}$	distal-crown
Min. [mm/mm]	$1.32 \cdot 10^{-3}$	distal - ½ crown	$2.91 \cdot 10^{-3}$	lingual - ½ crown	$9.50 \cdot 10^{-3}$	apex

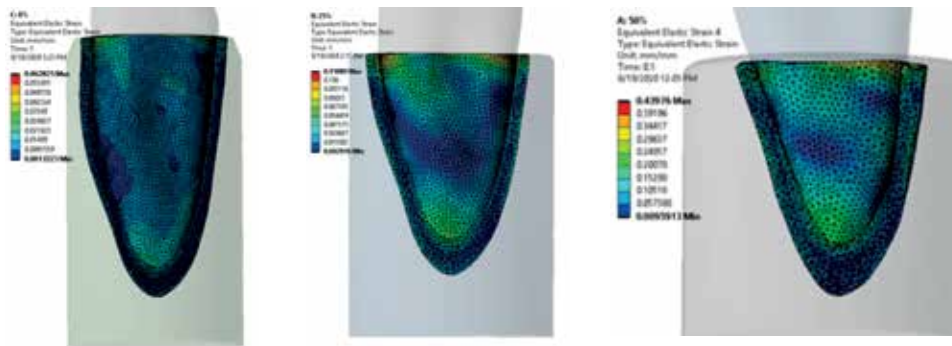


FIGURE 12. PDL ligament tensions in the border area of the crown, for the model with 0% bone loss; b. 50% bone loss;75% bone loss.

DISCUSSIONS

PL has an important role in biological functions through fibrous connections with cementum and alveolar bone and a mechanical role in absorbing occlusal forces and maintaining alveolar bone height. PL plays a key role in alveolar bone remodeling and resorption during tooth movements, which is why it is absolutely necessary for the practitioner to fully understand the predictability of tooth mobility under functional dental loads, i.e. the mechanical behavior of PL, which continues to raise questions from a biomimetic point of view [26, 4].

The morphological diversity of PL and its complex structure is a challenge when it comes to investigating its mechanical properties, as PL cannot be separated from the bone-PL-tooth complex without destroying the periodontal complex. That is why studies investigating the relationship between the mechanical properties of human PL and tooth position are limited [4].

These results have implications not only for understanding the stresses and strains that occur in PL under the action of forces generated by the dento-maxillary apparatus during the exercise of its functions, but also for providing complete information for practitioners and researchers on the role of material models used to address the mechanical behavior of PL and other components of the tooth.

The occlusal forces acting on the teeth depending on the somatic and psychic changes of each patient. Furthermore, the adaptive capacity of the periodontal complex, under the action of these forces, also varies significantly from individual to individual and even in the same individual in different periods of time.

One of the most used methods of theoretical analysis of biological structures is finite element analysis. Simulation of PL behavior by finite element analysis can be applied to calculate the stresses and quantitative deformations of PL under the action of static and dynamic forces but also to analyze the mechanical properties of the components implemented in the model [27].

The nonlinear model is much closer to reality because it simulates the reaction of the tissue to its immediate loading through a very low rigidity. After that, due to the hydrostatic pressure created in the interstitial fluid, the value of tissue stiffness increases exponentially.

The present study presents the results of the simulation with MEF of the behavior of the periodontal ligament in 3 clinical situations: for the integral biomechanical structure, in which the alveolar bone and the ligament have normal physiological dimensions and for the situations in which the bone undergoes a retraction by 50% and 75% of the initial value. These three variables of a geometric nature are added the type of material that simulates the PL, such as the non-linear model Mooney-Rivlin.

In this study we can see the deformity increases progressively with the loss of the supporting bone. Hemanth M et al. stated that the stresses found in the nonlinear analysis are higher than the stresses found in the linear analysis for the same magnitude of the force. Conversely, nonlinear analysis requires less force compared to linear analysis. [28].

PDL-induced apical stress increases as PDL thickness decreases. Alveolar bone stress decreases with increasing PDL thickness. Clinically, this distribution of stress may be useful in adult patients with

increased PDL thickness and the use of optimal forces to reduce the susceptibility to root resorption at the root tip [29].

The viscoelastic biomechanical behavior of PDL can better address the various tasks of PDL, i.e. anchoring the tooth to the bone, amortization while chewing, swallowing or squeezing, transmitting force to the bone, and transforming long-term orthodontic force into biomechanical signals for bone remodeling and orthodontic movement of the teeth (Keilig et al., 2016) [30].

Reducing alveolar bone height had little effect on the tooth and supporting tissue when 25% of the bone support was lost, however the stress increased dramatically when 50 and 75% of the bone support was lost. Also, the stress moved apically on the tooth, with increasing bone resorption, which is explained by the increase in the amount of force distributed in relation to the surface area, as tissue loss leads to a decrease in the surface, according to this study [31].

In our study, at the level of the whole periodontium it is found have maximum values in the mesial – crown area, and the minimum values are located and distal in $\frac{1}{2}$ crown.

In the case of a 50% gingival retraction, the changes are visible, so that in the nonlinear model shows the maximum distal-crown and the minimum crown-lingual.

Gingival retraction of 75% shows a value of the distal-crown maximum and the apex minimum for the nonlinear model.

Hemanth et al showed in their study that with the application of extrusive force, tensile stresses were observed at the apex, while the compression stress was distributed to the cervical margin. When the rotational force was applied, the maximum compression stress was distributed to the apex and the cervical third, while the tensile stress was distributed to the cervical third of the PL on the lingual surface [28].

Stress values were higher at the apex, regardless of the degree of bone loss. This may explain the appearance of the incipient periapical abscess, which appears as an apical radiolucency, in those teeth with primary occlusal trauma without periodontitis. Another result of the same study refers to the fact that the maximum stress values were recorded at the distal-vestibular level, near the enamel-cement junction, which may explain the occurrence of abrasion, due to excessive loads [31].

In most of the situations simulated in this study, minimal deformation occurs at or near the apex regardless of the material pattern used or the geometric pattern, which means that the effect of tooth rotation is lowest in that area. The maximum displacement appears physiologically on the mesial surface then the geometric model simulates the in-

tegral structure. As bone loss occurs and develops, the maximal point changes position toward the distal surface, but also in the coronary third. This indicates that the strain deformations become larger than the compression deformations, the distal face being the one that opposes the direction of stress, and which is consequently the surface subject to stretching. Also, in our study, the deformities on the lingual surface of the PL are very small compared to the other faces, and the average $\frac{1}{3}$ of each surface analyzed has smaller deformations.

The study by Reimann S. et al. who analyzed the role and mode of action of the occlusal forces on the PL found that when applying a normal force at the level of a healthy periodontal structure the minimum tensions were highlighted on the middle part of the PL, and the maximum tensions on the palatal side. Applying the same normal load on the PL resulted in the production of minimal stresses on the distal side in the present situation of 50% bone loss and maximum stresses are seen at the junction of the apical and middle third when the bone loss is 75% on the palatal side [32].

At a physiological height of the alveolar bone, at a load of 150N (normal function), in PL the minimum stresses were observed on the mesial surface at the level of the enamel-cementum junction, and the maximum ones at the level of the palatal surfaces. As the load increased to 290N, the voltages also increased by 90%. In cases with bone resorption, at a load of 150N (normal function), the minimum stresses were observed at the superficial surface in the middle third of the root with 50% bone loss and the maximum tensions at the palatal face at the junction of the apical third with the middle third of the root with 75% bone loss. As the load increased to 290N, the voltages increased even by 90% [18].

The data from previous research studies are useful both for understanding the periodontal complex and how it works and responds to different stimuli, and for the evolution of treatment methods in terms of its pathology. However, it is difficult to make clear comparisons between the results of studies conducted so far due to variations in methodology, but also the proposed objectives.

CONCLUSIONS

The nonlinear model is much closer to reality because it simulates the reaction of the tissue to its immediate loading by a very low rigidity after which, due to the hydrostatic pressure created in the interstitial fluid, the value of tissue stiffness increases exponentially.

The deformities on the lingual surface of the PL are very small compared to the other surfaces.

For the model without bone loss is found that maximum values are in the mesial crown area, and

the minimum values are located at distal in ½ crown area.

Bone loss is associated with a change in the position of the center of rotation due to the lack of supporting bone volume. Its migration occurs to the apical area.

In the case of a 50% gingival recession, the changes are visible, the nonlinear model has the maximum distal crown and the minimum lingual crown.

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REFERENCES

1. Abraha HM, Iriarte-Diaz J, Ross CF et al. The Mechanical Effect of the Periodontal Ligament on Bone Strain Regimes in a Validated Finite Element Model of a Macaque Mandible. *Frontiers in Bioengineering and Biotechnology* Volume 7, Article 269 October 2019 30;7:269.
2. Nanci A, Bosshardt DD. Structure of periodontal tissues in health and disease. *Periodontology* 2000. 2006; 40:11–28.
3. Minch L. Material properties of periodontal ligaments. *Postepy Hig Med Dosw.* 2013 Dec 11;(67):1261–4.
4. Wu B, Fu Y, Shi H et al. Tensile testing of the mechanical behavior of the human periodontal ligament. *BioMed Eng OnLine.* 2018 Nov 23;17(1):172.
5. Mori S, Sato T, Hara T et al. Effect of continuous pressure on histopathological changes in denture-supporting tissues. *J. Oral Rehabil.* 1997 Jan;24(1):37–46.
6. Imai Y, Sato T, Mori S, Okamoto M. A histomorphometric analysis on bone dynamics in denture supporting tissue under continuous pressure. *J. Oral Rehabil.* 2002 Jan;29(1):72–79.
7. Isobe A, Sato Y, Kitagawa N et al. The influence of denture supporting tissue properties on pressure–pain threshold: measurement in dentate subjects. *J. Prosthodont. Res.* 2013 Oct;57(4):275–83.
8. Atwood DA. Reduction of residual ridges: major oral disease entity. *J. Prosthet. Dent.* 1971 Sep;26(3):266–279.
9. Akazawa H, Sakurai K. Changes of blood flow in the mucosa underlying a mandibular denture following pressure assumed as a result of light clenching. *J. Oral Rehabil.* 2002 Apr;29(4):336–40.
10. Kumakura S, Sakurai K, Tahara Y, Nakagawa K. Relationship between buccal mucosa ridging and viscoelastic behaviour of oral mucosa. *J. Oral Rehabil.* 2011 Jun;38(6):429–33.
11. Leiderman R, Barbone PE, Oberai AA, Bamber JC. Coupling between elastic strain and interstitial fluid flow: ramifications for poroelastic imaging. *Phys. Med. Biol.* 2006 Dec; 21;51(24):6291–6313.
12. Stokes IAF, Laible JP, Gardner-Morse MG et al. Refinement of elastic, poroelastic, and osmotic tissue properties of intervertebral disks to analyze behaviour in compression. *Ann. Biomed. Eng.* 2011 Jan;39(1):122–131.
13. Yoshida N, Minagi S, Sato T et al. Effect of mechanical pressure on the blood flow in human palatal mucosa measured by temperature controlled thermoelectrical method. *J. Oral Rehabil.* 1991 Sep;19(5):527–533.
14. Chen J, Ahmad R, Li W et al. Biomechanics of oral mucosa. *J. R. Soc. Interface.* 2015 Aug 6;12(109):20150325.
15. Sato T, Hara T, Mori S et al. Threshold for bone resorption induced by continuous and intermittent pressure in the rat hard palate. *J. Dental Res.* 1998 Feb;77(2):387–92.
16. Lytle RB. 1962 Soft tissue displacement beneath removable partial and complete dentures. *J. Prosthet. Dent.* 12,34.
17. Cai, Y., Yang, X., He, B. et al. Finite element method analysis of the periodontal ligament in mandibular canine movement with transparent tooth correction treatment. *BMC Oral Health.* 2015 Sep 4;15:106.
18. Reddy RT, Vandana KL. Effect of hyperfunctional occlusal loads on periodontium: A three-dimensional finite element analysis. *J Indian Soc Periodontol.* 2018;22(5):395-400.
19. Poppe, M, Bourauel C, Jäger A. Determination of the elasticity parameters of the human periodontal ligament and the location of the center of resistance of single-rooted teeth a study of autopsy specimens and their conversion into finite element models. *J. Orofac. Orthop. Fortschr. Kieferorthopädie.* 2002 Dec; 63(5): 358–370.
20. Liu, TC, Chang CH, Wong TY, Liu JK. Finite element analysis of miniscrew implants used for orthodontic anchorage. *Am. J. Orthod. Dentofac. Orthop.* 2012 Apr;141(4): 468–76.
21. Cattaneo PM, Dalstra M, Melsen B. The finite element method: a tool to study orthodontic tooth movement. *J Dent Res.* 2005 May;84(5):428-433.
22. Tanne K, Yoshida S, Kawata T et al (1998). An evaluation of the biomechanical response of the tooth and periodontium to orthodontic forces in adolescent and adult subjects. *Br J Orthod.* 1998 May;25(2):109-115.
23. Vollmer D, Bourauel C, Maier K, Jager A (1999). Determination of the centre of resistance in an upper human canine and idealized tooth model. *Eur J Orthod.* 1998 Dec; 21(6):633-648.
24. Jones ML, Hickman J, Middleton J, Knox J, Volp C (2001). A validated finite element method study of orthodontic tooth movement in the human subject. *J Orthod.* 2001 Mar; 28(1):29-38.
25. Toms SR, Eberhardt AW (2003). A nonlinear finite element analysis of the periodontal ligament under orthodontic tooth loading. *Am J Orthod Dentofacial Orthop.* 2003 Jun; 123(6):657-65.
26. Pini M, Zysset P, Botsis J, Contro R. Tensile and compressive behaviour of the bovine periodontal ligament. *J Bio-mech.* 2004 Jan;37(1):111–9.
27. Karimi A, Razaghi R, Biglari H et al. Finite element modeling of the periodontal ligament under a realistic kinetic loading of the jaw system. *The Saudi Dental Journal.* 2020 Nov;32(7):349-356. ISSN 1013-9052.
28. Hemanth M, Raghuvveer HP, Rani MS et al. An Analysis of the Stress Induced in the Periodontal Ligament during Extrusion and Rotation Movements—Part II: A Comparison of vs Nonlinear FEM Linear Study. *J Contemp Dent Pract* 2015 Oct 1;16(10):819-823.
29. Gupta M, Madhok K, Kulshrestha R et al. Determination of stress distribution on periodontal ligament and alveolar bone by various tooth movements - A 3D FEM study. *Journal of Oral Biology and Craniofacial Research.* 2020 Oct-Dec;10(4):758-763.
30. Keilig L, Drolshagen M, Tran K, Hasan I, Reimann S, Deschner J, Brinkmann K, Krause R, Favino M, Bourauel C. In vivo measurements and numerical analysis of the biomechanical characteristics of the human PDL. *Anat.* 2016 Jul;206:80-88
31. Muneer S, Vandana KL. Effect of Different Occlusal Loads on Periodontium: A Three-dimensional Finite Element Analysis. *CODS J Dent.* 2016;8(2):78-90.
32. Reimann S. et al. Biomechanical analysis of mandibular frontal crowding in the presence of gingival recession defects. *8th World Congress of Biomechanics.* P4103, 8-12 July, Dublin, Ireland 2018.