ISSN 2176-5480

22nd International Congress of Mechanical Engineering (COBEM 2013)
November 3-7, 2013, Ribeirão Preto, SP, Brazil


# STRAIN LEVEL AT MIDPALATAL SUTURE - CORRELATION WITH MECHANOBIOLOGICAL CONCEPTS.

**Larissa Carvalho Trojan-Serpe**
Universidade Federal de Minas Gerais, Belo Horizonte/ MG - Brasil
E-mail : laritrojan@gmail.com

**Libardo Andrés González-Torres**
Universidade Federal do Vales do Jequitinhona e Mucuri, Diamantina/ MG – Brasil
E-mail: l.gonzales@ict.ufvjm.eu.br

**Leandro Seles Dorneles**
Universidade Federal de Minas Gerais, Belo Horizonte/ MG - Brasil

**Ana Claudia Moreira Melo Toyofuku**
Instituto Latino Americano de Ensino e Pesquisa Odontológico, Curitiba/PR – Brasil

**Estevam Barbosa de Las Casas**
Universidade Federal de Minas Gerais, Belo Horizonte/ MG - Brasil
E-mail : estevam@dees.ufmg.br

***Abstract.*** *Rapid Maxillary Expansion (RME) is an orthodontic technique widely used in dental clinics to induce the separation of the midpalatal suture (MPS) and increase the transverse dimension of the maxilla. The purpose of this computer simulations study using finite element method was analyzes the stresses and strains distribution generated in the maxillary bone, mainly in the MPS. The objective of this work is to estimate the strains and displacements experienced along the suture and to study the correlation between mechanical stimulus and sutural growth, based on mechanobiological theories. For this two stimuli present during expansion were considered: the displacement applied during the RME treatment and the magnitude of chewing force, they both affect the mechanical environment in sutures. The strain level promoted by expansion displacement was in the range of 6% to 20% on anterior inferior region of MPS, in the top (anterior to posterior) were observed strains lower than 1%. Considering only the masticatory loads, the highest strains occurred in the anterior inferior region (0.35% to 0.7%). The presented results indicate that numerical simulation may predicted correctly the shape of maxilla opening and bone growth during RME and also that magnitude of the chewing force can be a relevant factor affecting mechanical environment of sutures.*

***Keywords:*** *Mechanobiology; Midpalatal Suture; Rapid Maxillary Expansion; Bioengineering; Orthodontic.*

## 1. INTRODUCTION

Biomechanical studies on dental structures using finite element method (FEM) are common in orthodontics (Holberg and Rudzki-Janson, 2006; Maruo, 2011; Araugio, 2009). FEM is an important tool for stress and strain analysis, for example those that are generated in the craniofacial complex due to rapid maxillary expansion (RME) (Provatidis, *et al.,* 2008). This orthodontic treatment consists basically in the separation of the midpalatal suture (MPS) in order to increase the transverse dimension of the maxilla, which is needed, for example, to correct the atresic palate format.

The MPS, as well as other facial sutures, is a network of connective tissue between mineralized bones. MPS forms a kind of hinge joining the two jaws (right and left parts of maxilla) and extends throughout the hard palate. The distance from a bone end to another is between 161.16 and 211.20 µm (Knaup, *et al.*, 2004). The MPS ossification process occurs lately, and gets a sinuous and imbricate morphological aspect in the bone edges, generating a space filled by a dense fibrous connective tissue organized in several cell layers (Consolaro and Consolaro, 2008). Some studies indicate that skeletal maturation of MPS may limit the RME (Haas, 1973; Haas, 1970; Wertz, 1970).

The technique used for this purpose - Rapid Maxillary Expansion - consists in the use of a palatal expander, which is an appliance attached to the posterior teeth and may have a relative support in the palate soft tissue. This device applies forces to the lingual-buccal direction (from palate to cheek) for each side.

Although widely used in dental clinic, RME still presents some common problems: increasing of the vestibular (region between lip/cheek and teeth) tipping of the posterior teeth (Sun *et al.*, 2011); limited skeletal displacement (Shapiro and Kokich, 1988); dental root resorption (Erverdi, 1994) and lack of anchorage to maintain long-term suture expansion (Parr et al. 1997). The orthopedic effect of RME can also show some degree of relapse (Bishara and Stanley, 1987), but the dental effect is more unstable after expansion (Haas, 1973; Timms, 1968). Therefore, the greater the skeletal effect and the lower the tooth movement, better the prognosis in terms of stability.

ISSN 2176-5480



This study supported in computer simulations for the structural analysis of maxilla uses consistent numerical techniques to study the behavior of MPS in order to estimate the strains and displacements experienced along the suture in the presence of external and internal mechanical forces and also to correlate the mechanical stimulus and the sutural growth based on mechanobiological theories. This could explain the ossification of the suture observed in clinical studies after expansion.

## 2. MECHANOBIOLOGIC ASPECTS

Based on the mechanobiological aspects, it is important to note that in bone and other tissues that hold an important structural function, such as those affected by RME, it has been observed the importance of mechanical factors in its structure, performance and maintenance (Robling, *et al*., 2006; Mao, 2002).

Several theoretical models [Pauwels, 1960; Perren, 1979; Perren and Cordey, 1980; Prendergast, *et al.,* 1997] have been formulated to predict how the differentiation of tissues occurs in relation to the mechanical stress (or strains) they undergo. The differentiation of mesenchymal stem cells present in the beginning at fracture area may occur into osteoblasts, chondroblasts or fibroblasts, depending on the mechanical and biological stimuli. These cells work together to synthesize new tissues: fibrous tissue, cartilage or bone, generating a *differentiation* of the initial tissue to particular specialized ones.

The model proposed by Prendergast, *et al.* (1997) contains most of the biologically relevant processes such as cell division and differentiation, tissue regeneration and ideas of cell motility, incorporating the influence of mechanical properties and stimuli on the biological processes (Fig. 1). This mechanoregulatory model for tissue differentiation is based on the poroelastic (biphasic) tissues behavior. The combined effect of octahedral shear deformation (variable quantifying the distortion of the shape in a particular point) and maximum fluid velocity relative (variable quantifying the velocity of interstitial fluid inside tissues) constitute the stimulus that controls the differentiation process. High values of deformation and fluid velocity favors fibrous tissue formation, while intermediates values leading to cartilaginous tissue formation. Bone can be formed only if the values are sufficiently low (Fig. 1).

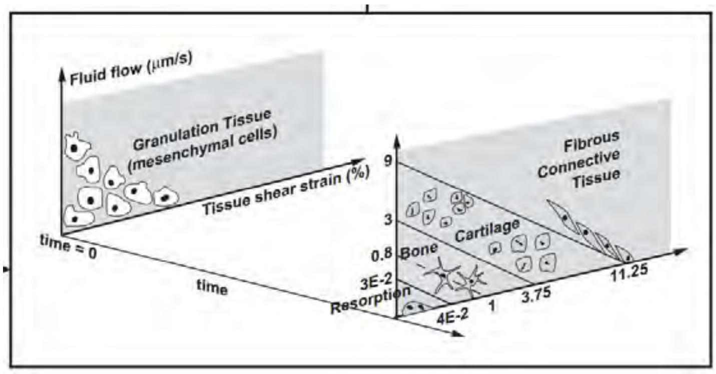


Figure 1. Mechanoregulatory model proposed by Prendergast, *et al.* (1997).

Given the observed influence of mechanical factors on tissue differentiation, as represented by the model of Predergast, et al. (1997), it is reasonable to think that these biological processes also occur for tissues in maxilla and are affected by mechanical stimuli. Thus, came the need of a structural study of the maxilla to evaluate the role of mechanics in the mechanism of tissue formation. In particular, in this work we study structurally the MPS, to evaluate qualitatively the mechanical environment inside it during the treatment for the RME, and study its influence on tissue formation. This study also intends to improve the understanding of the role of cranial sutures connected to maxilla during RME and of the elements that generate resistance to expansion.

## 3. MATERIALS AND METHODS

In order to analyze the mechanical behavior of the MPS when subjected to treatment with palatal expander, a three-dimensional model of the maxilla was developed.

The three-dimensional model of the maxilla was initially extracted from computed tomography (CT) images of a 25 years old patient. These images, in turn, were saved in DICOM format and imported into the software for computer-aided design, Simpleware® (Innovation Centre, Exerter, UK). Each one of the sections in DICOM images was segmented according to the gray scale level of different structures in images. Thus, the different levels of grays were associated to different materials to identify the most important structures in maxilla.

The tools for automatic segmentation of Simpleware® were not enough for the satisfactory definition of all structures such as enamel and periodontal ligament. Therefore, the software SolidWorks 19.4 (Concord, Massachusetts,

ISSN 2176-5480



USA, 1993) was used to refine the model. On this computer aided design (CAD) software, the geometric models were modified to better define all structures and to represent more accurately the real geometry. The outer surfaces of teeth and bone were partitioned manually to the correct definition of the structures, from the refined geometry already obtained. The other components of the maxilla geometric model (bone, enamel and periodontal ligament) were then defined using other software features as: shells, offsets, splines and lofts. The dental pulp was taken as an empty space due to its low modulus of elasticity. To represent the MPS a solid was added in the maxilla mid region, representing a SPM thickness of 1 mm. Altogether, the model was composed of the following structures: bone, enamel, dentin, pulp, periodontal ligament, MPS and orthodontic bands, where the appliance was bonded on posterior teeth, as shown in Fig. 2. Aiming to reduce the computational costs, four anterior teeth were removed from the model, since they have little significant influence on the structures of interest in this study.

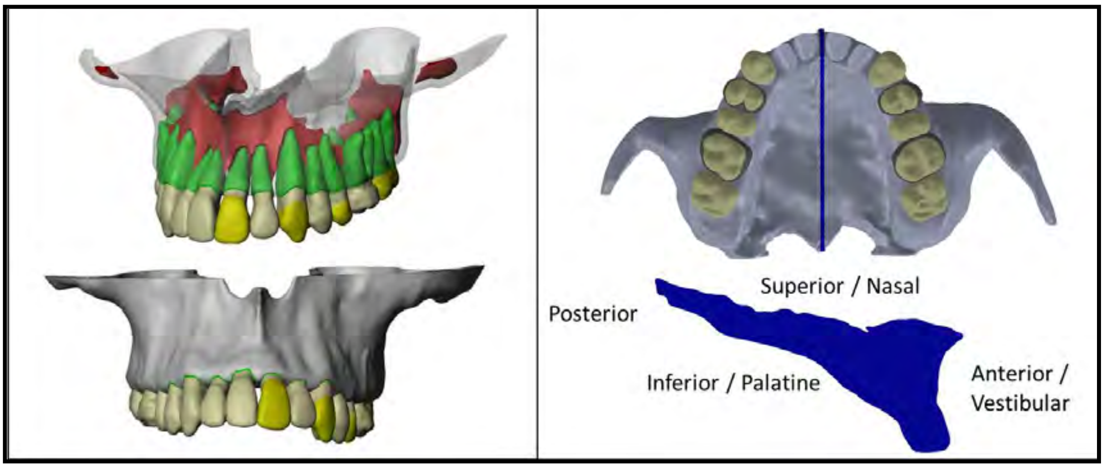


Figure 2. Maxillary geometric model. (A- Complete model; B- Geometric model used with orthodontic band and MPS added. The MPs is showed in a closer view with proper regions)

The discretization of the model was performed in ANSYS (version 12, Belcan Engineering Group, Downers Grove, Ill), generating a mesh of 9 450 114 tetrahedral elements and 1 653 710 nodes. The mesh was then exported to the finite element program Abaqus 6.12 (Providence, Rhode Island, USA, 2012), a software for computer-aided engineering (CAE), that was used to estimate the mechanical environment in maxilla, considering the mechanical properties of each material, external loads and boundary conditions (Fig. 3).

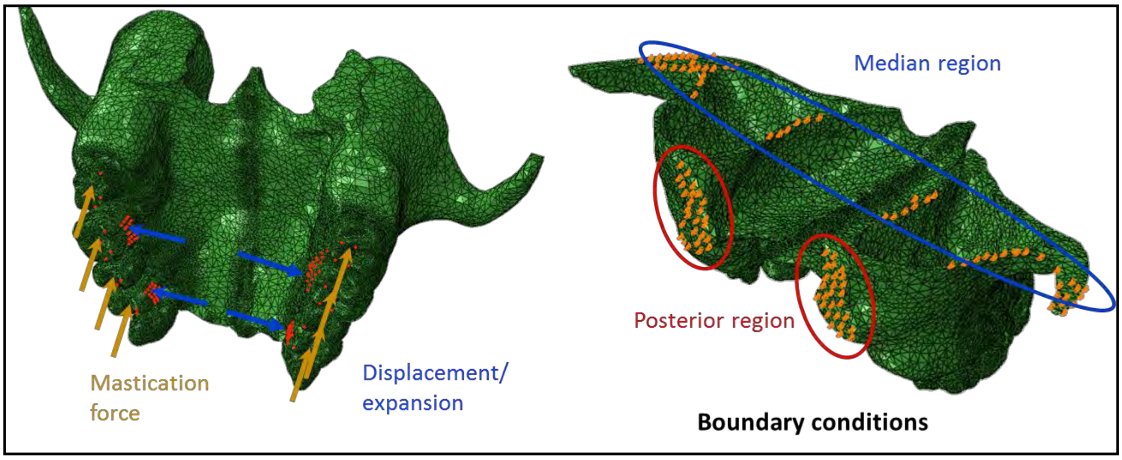


Figure 3. Finite Element Model with load, displacement and boundary condition regions represented.

All structures were considered as isotropic linear elastic materials according to properties reported in the literature. The values used for the elastic modulus (E) and Poisson's ratio (υ) were as follows: bone 10 000 MPa and 0.3 – intermediate properties considering cortical bone, 13 700 MPa and 0.3 and trabecular bone, 1 370 MPa and 0.3 (Yu, *et al.*, 2007) - enamel, 84 100 MPa and 0.2 (Jones, *et al.*, 2001); dentin, 18.600MPa and 0.31 (Jones, *et al.*, 2001); periodontal ligament, 0.69 MPa and 0.49 (Yoshida, *et al*., 2001) and stainless steel, 200 GPa and 0.33 (Hibbeler, 2004). For MPS two different elastic moduli were assumed: one equal to 1 MPa (as proposed by Provatidis, *et al.* (2008)) and

ISSN 2176-5480



another one equal to 1 Pa, suggesting absence of an organized tissue. Regarding boundary conditions, they have been defined as fixed conditions on the posterior regions of the model, considering the rigidity established by the base of the skull. It has been also considered for models 2 and 5 (in Table 1.), fixed conditions on medium region just in one direction (limiting upper displacement) to represent the lower constraints in this region when compared to the posterior region.

The simulation performed in this study aims to study the mechanical environment of MPS promoted by the forces generated by the expander appliances and by mastication, forces that stimulate the suture region. And, as previously related, mechanical environment generated by natural forces may affect the remodeling of tissues during and after RME.

Six finite element models (FEM) were implemented and solved in this work. In two of them the palatal expander action was represented by an applied displacement of 0.25 mm (Model 4 and 5) on lingual surfaces of orthodontic bands, in lingual/buccal direction, on first pre molar and first molar. This displacement occurs for each appliance's activation, usually done twice a day. Models 1 and 2 (Table 1.) consider the total displacement obtained at the end of RME, which depends on patient's needs, for simulations an opening of 7.5 mm was considered, as done by Provatidis, *et al. (2008*). For models regarding mastication force, it was applied a force of 70 N distributed on oclusal face of posterior teeth on models 3 and 6. As masticatory force has great amplitude, an intermediated value of lower forces (12 to 150 N) which commonly appeared in all masticatory cycles was selected (Widmark, *et al*., 1995).

Mastication forces are important because in the period between activations and after finished RME, only these natural stimuli are present. Mastication forces are cyclic, nevertheless for simplicity in this work we just considered the maximum value of this force, represented here as a constant load. Models including mastication forces (Models 3 and 6) are different only for MPS' elastic modulus. Model 3 (E = 1 Pa) assumed as a very low elastic modulus for MPS, representing the period before the connective tissue is remodeled, it occurs after RME or between activations. After this, when the connective tissue in suture is remodeled, the elastic module of suture increased. Model 6 considers this situation with a higher elastic modulus (E = 1MPa) reported in literature (Gonzalez-Torres, 2011).

Table 1. Finite Element Models simulated.

| FEM | Elastic Modulus - MPS | Loading | Boundary Conditions |
|---|---|---|---|
| **1** | 1 Pa | Displacement (7.5mm ) | Posterior clamped |
| **2** | 1 Pa | Displacement (7.5mm ) | Posterior and Medium (Y-direction) clamped |
| **3** | 1 Pa | Mastication force (70 N) | Posterior clamped |
| **4** | 1 MPa | Displacement (0.25mm ) | Posterior clamped |
| **5** | 1 MPa | Displacement (0.25mm ) | Posterior and Medium (Y-direction) clamped |
| **6** | 1 MPa | Mastication force (70 N) | Posterior clamped |

## 4. RESULTS

The total displacement derived from RME on models 1 and 2 are represented in Tab. 2. The values were taken from three MPS regions: anterior and inferior region of MPS, close to central incisors (UI); anterior and superior region, close to nasal anterior spine (US); on level of canines (UC) and first molars (UM); and at posterior and inferior region (UP). Both models showed MPS "V" shaped opening with fulcrum in the posterior region as shown in Fig. 4.

Table 2. Displacement Finite Element Models simulated (mm).

| FEM | UI | US | UC | UM | UP |
|---|---|---|---|---|---|
| **1** | 5.69 | 3.97 | 4.64 | 2.26 | 0.17 |
| **2** | 4.2 | 2.1 | 3.7 | 1.67 | 0.07 |

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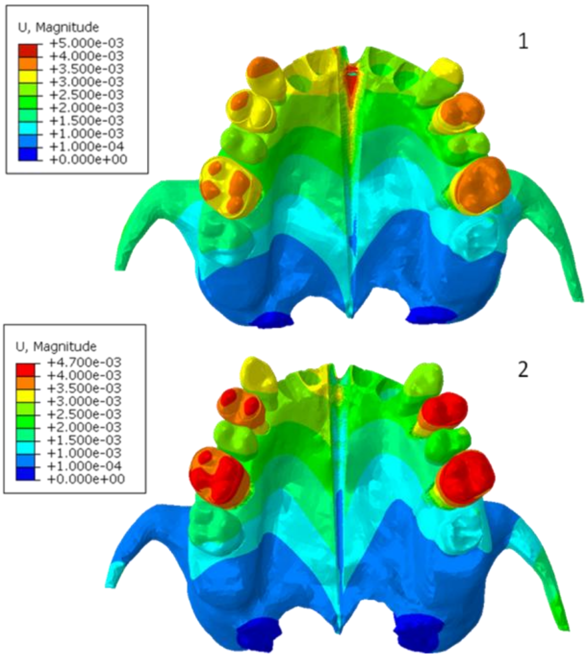


Figure 4. Total displacement after RME (Models 1 and 2).

The influence of boundary conditions on FEM was also evaluated. It was applied a displacement representing a single activation (in models 4 and 5). The stress distribution was very similar for both models, either on palatal region as in the vestibular region of alveolar bone (Fig. 5).

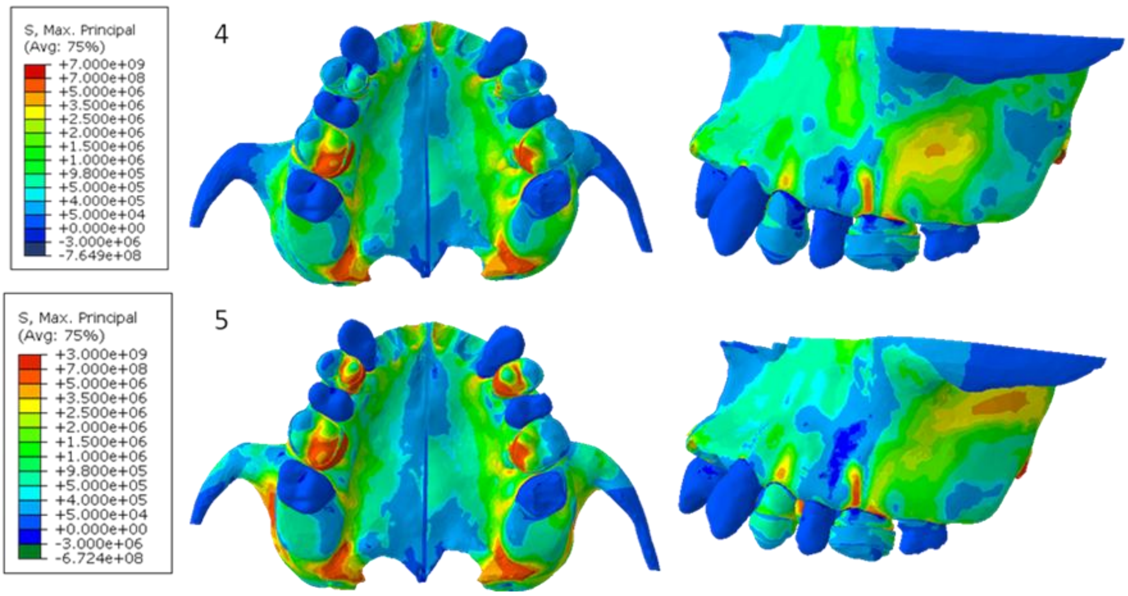


Figure 5. Maximum principal stress distribution due single appliance activation (0.25mm).

From models 3 to 6 were examined the strain levels estimated for MPS (Fig. 6) and some regions were selected to evaluate quantitatively this aspect, as showed in Tab. 3. The strains generated on MPS from models 4 and 5 show how the suture is affected by a single activation of palatal expanders, reminding that more rigid boundary conditions were used for model 5, representing a higher stiffness of craniofacial sutures than in model 4. As occurred for stresses distribution, deformation levels did not show great variance for both models. The highest strain was observed on anterior and inferior region, 6% up to 20%. And these levels were decreasing to upper regions, achieving the nasal extremity close to null deformation.

ISSN 2176-5480



The increased MPS stiffness used for model 6 causes a decreasing in strain levels in relation to model 3.

Table 3. Simulated strains on Finite Element Models (%).

| FEM | Anterior and inferior | Anterior and superior | Posterior |
|---|---|---|---|
| **3** | 1.5% - 2.5% | 1.5% - 2% | 0.1% - 0.3% |
| **4** | 6% - 20% | 0.9% -1% | 0.01%-0.5% |
| **5** | 6% - 20% | 0.9% -1% | 0.01%-0.5% |
| **6** | 0.35% - 0.7% | 0.2% - 0.3% | 0 – 0.1% |

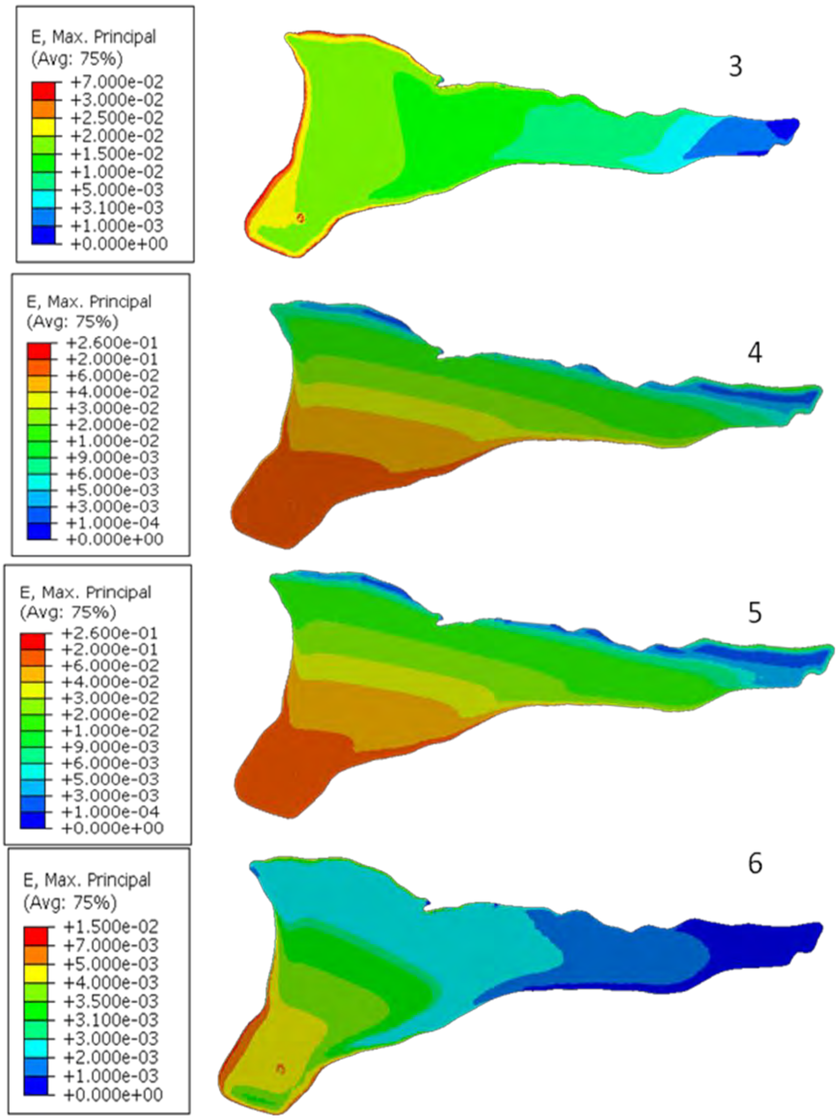


Figure 6. Maximum Principal Strain due single appliance activation (Model 4 and 5)
and mastication force (Model 3 and 6).

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## 5. DISCUSSION

RME has been largely studied in orthodontics since mid-1960s. According to clinical studies, the suture opening was dependent of skeletal maturation, resulting in a "V" shape with fulcrum on posterior region (Bishara and Stanley, 1987; Weissheimer, 2011). The displacements estimated with models 1 and 2 show decreasing values from incisor to posterior region. The values from model 2 - displacement of 4.2 mm on MPS near to incisor teeth, 3.7mm on canine region and 1.67mm on molar region - were similar to a previous work (Ghoneima *et al.*, 2011) in which the authors obtained displacements up to 4.8 mm for MPS region between incisor teeth, up to 4.5 mm for canine region and up to 2.1mm to molar region. For model 1, assumed boundary conditions were less rigid than the ones considered in model 2; for this reason, it showed larger displacements than model 2. Both results (for models 1 and 2) were lower than displacements found by Provatidis *et al.* (2007) who applied a displacement of 7.5mm on the FEM of full cranial bone; however, the displacements of our models were more similar to results obtained from experimental studies (Ghoneima *et al.*, 2011). In general terms, it was observed that restriction applied on a FEM on median superior region can modulate MPS opening, and that the stress distribution on maxillary bone and strain at MPS is not significantly affected, as showed for models 4 and 5.

Although RME is a well-known orthodontic approach, the correlation between mechanobiological aspects with remodeling process that occurs on midpalatal region after MPS opening is still unclear. Some radiographic and tomographic studies (Silva Filho *et al.*, 2007, Ballanti *et al.,* 2010) show that mineralization process to reestablish the suture can take approximately 6 to 9 months and during this period some retention appliance should be used to avoid relapse of palatal expansion. More controlled studies are necessary to understand the regeneration process and to control the relapse tendency.

Considering computational studies, several works have been performed obtaining consistent results with the tissue differentiation observed during tissue regeneration and remodeling (Isaksson, *et al.* 2007, Lacroix and Prendergast 2002, Gomez-Benito, *et al.* 2005). These studies correlated the tissue that appeared with different levels of mechanical stimuli during the regeneration process. One variable used in various computational works to determine the differentiation of tissues in the local of regeneration is the magnitude of distortional strain. Thus, in order to study the differentiation of tissues in the MPS, we computed maximum strain (a variable that in this case has a value very similar to the distortional strain), to related the magnitude of it with the tissues appearing inside the MPS.

According to the level of strains computed for the different studied cases, we can discuss on which tissues should appear in each local according to mechanical stimuli. Considering the differentiation rule of Prendergast (Lacroix and Prendergast, 2002) and low magnitudes of velocities of fluid flow, it is possible to specify some ranges of strains at which fibrous tissue, cartilage and bone are promoted (Fig. 2, at low magnitudes of velocity). These ranges are: bone appears with strains between 0.04% and 3.75%, cartilage with strains between 3.75% and 11.25% and fibrous tissue with strains higher than 11.25%. Considering these limits and the results obtained in this work, it is possible to conclude that the stimulus generated during expansion (cases 4 and 5, Fig. 6) are high (between 6% and 26%, Fig. 6) in the inferior zone of MPS, promoting differentiation to cartilage and fibrous tissue. In the nasal zone and almost in half of the MPS, the level of strains is low enough (0.04% and 4%, Fig. 6) to cause the differentiation to bone.

Considering the MPS during mastication, in the case with a suture of E = 1Pa (Fig. 6), it is observed that almost all the MPS is predicted to be differentiated to bone (strains between 0.04% and 4%), just a small zone on the anterior perimeter of the MPS is predicted to be differentiated to cartilage. In MPS with E=1MPa, results were similar, but in this case cartilage is not promoted and a resorption zone appear on the posterior part of MPS.

Thus, according to the magnitude of strains in the different cases studied, some important points are observed. First, during expansion syntheses of bone is difficult in the whole MPS because of the high level of strains caused by expansion forces, for this reason fibrous tissue is promoted in a large zone of MPS. Second, likely the period in which more new bone is synthesized occurs when the expansion forces are suppressed, period in which just mastication forces are present and therefore low levels of strains are generated. Thirdly, MPS with low Young modulus (with E = 1 Pa) generate mechanical stimulus that promote the syntheses of bone more than the stiffer MPS (with E = 1 MPa), this suggest that a less rigid suture would have to be present during the period of more syntheses of bone.

In addition to the magnitude of stimuli is the frequency of the mechanical stimuli, which directly affects the interstitial fluid flow in tissues, and can induce the movement of waste, nutrients or growth factors, as well as stimulating the cellular proliferation and differentiation (Gonzalez-Torres, 2011). Based on these assumptions it is possible to consider that the mastication force as a cyclic stimulus could have an important role on regeneration processes of MPS.

Previous discussion agrees with some authors (Katsaros, *et al.*, 1994, Behrents, *et al.*, 1978) who observed that the suture appears to be exceptionally sensitive in its response to mechanotransduction. As little as a change in the amount of solid food consumed or incisor extractions in animal studies leads to significant changes in morphology and metabolism suture. In addition, relations between form and function of the cranial sutures and masticatory muscles were confirmed by animal study (Katsaros, *et al.*, 2006). The authors of this study refer to their results in animal experiments on the reduction of strain on the sutures caused by mechanical forces exerted by the masticatory muscles. This was the reason to computationally simulate the RME in this study to estimate strains in the suture and relate them to tissues observed in MPS.

## 6. CONCLUSION

An important characteristic of finite element analysis is the ability to predict events without needing exposure to biological material testing. Currently, this method is limited to the available time and computing power that require simplifications of complex structures such as the skull bones.

According to the literature review presented in this work on regeneration of bone tissue, primarily focused on long bones, one can expect some correlation between strain levels and the suture expanded regeneration. The presented results indicate that:

- Numerical simulation predicted correctly the shape of bone growth during RME;
- The magnitude of the chewing force can be a relevant factor in determining mechanical environment of the sutures. For this reason, both mastication and orthopedic forces have to be considered for studying the SPM during the MRE;
- Mechanobiologic theories of differentiation may be important to understand the syntheses of tissues in MPS during and after the RME treatment.

## 7. ACKNOWLEDGEMENTS

The authors want to thank to the CAPES for the support to the development of the present work.

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