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Research Article

Evaluation of Stress in Upper Jaw Caused By Orthodontic Mini-Screws (Oms) Produced From Different Biomaterials Using 3 Dimensional Finite Element Analyse

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Abstract

Objective: The purpose of this study was to evaluate the stress in upper jaw caused by orthodontic mini - screw produced from different biomaterials using 3 dimensional finite element analyse.

Materials and Method: For this purpose, there were constructed 4 different scenarios (stainless steel, Ti6Al4V, TiZr and Zr orthodontic mini screw biomaterials) without changing the treatment mechanic and orthodontic screw dimensions. The scenario was constructed on a premolar extraction case, in upper jaw; the canine tooth retraction was performed on a continuous arch (0,016 x 0,022 stainless steel) using 0.0018'' slot brackets and the orthodontic mini screw (OMS) was placed between second premolar and the first molar tooth. The canine retraction has been realized by hanging a closed coil from the hook of the brace to OMS, which applied a force of 1 N.

Results and Conclusion: According to the results; maximum principal stress (Pmax) values recorded at the cortical bone, starting from the smallest, were respectively stainless steel, Zr TiZr, and Ti6Al4V; maximum principal stress (Pmax) values recorded at the cancellous bone, from the smaller, respectively is found as following: stainless steel, Zr, TiZr, and Ti6Al4V.

Keywords: Biomaterial; Mini Screw; Orthodontics; 3 Dimensional Finite Element Analyse

Introduction

In order to achieve optimal orthodontic results, the orthodontist has to balance many factors. Some of those factors depend on clinician and some others to patients. Correct diagnosis and the treatment planning, the implementation of the appropriate devices and installation of the appropriate treatment mechanics are some of the factors that are depended from the clinicians. On the other hand, providing an optimal oral health, respecting doctor's advices, diet limitations and regularly attending appointments are some of the factors, which have impact on stabile, functional and aesthetical results and are depended on the patients [1]. Since patient compliance/cooperation is one the most difficult aspect of orthodontics, clinicians historically have given importance to develop treatment mechanics that will require less cooperation [2]. Nowadays-orthodontic screws with different designs, size and shapes used in orthodontic practice have been produced from Titanium (Ti) alloys.

The fact that this alloy contains Aluminum (Al) and Vanadium (V) ions and causes local tissue reaction and immunological reactions by releasing harmful ions to the body has led to the doubts about its biocompatibility properties. Although their biocompatibility properties have been investigated in details, there are not many studies on the effects of Al and V ions. Al ions affect the proliferation, differentiation and metabolic activities of osteoblasts. Although V is a necessary element, it is considered to be a toxic element because there is a very fine line between the required dose and the toxic dose. The cytotoxic effects of V have been scientifically investigated and proven. V acts on macrophages and fibroblasts, binds to proteins and causes them to proliferate and accumulate in certain parts of the body [3].

Aiming to eliminate these side effects, some materials used in the production of implants were revised and Ti-Zr alloy based dental implants were produced.

It is known that Ti-Zr alloy based implants have higher tensile strength and are successful even in small diameters without compromising more compatible mechanical properties than other titanium alloys [4]. The purpose of this study was to evaluate stress in upper jaw caused by orthodontic mini - screw produced from 4 different biomaterials (stainless steel, Ti6Al4V, TiZr and Zr alloy) using 3 dimensional finite elements analyse.

Material and Methods

This study was designed to simulate the left side of upper jaw and to assess with 3 dimensional element analyses the stress release from the OMS, were produced from 4 different alloys: stainless steel (SS), Ti6Al4V, TiZr and Zr alloy. The treatment and dimensions of the OMS used in the scenarios has not been changed and 4 different models have been constructed.

The first premolar tooth was extracted for the orthodontic treatment. The canine retraction was performed on a continuous dental arch (16 x 22 stainless steel) using 0.0018'' slot brackets and to maintain anchorage OMS has been installed between second premolar and the first molar.

For the retraction, a closed spring was hanged from the hook of the canine bracket towards the OMS, applying a force of 1 N (Figure 1). This study has been conducted between Gazi University Department of Orthodontics and Parsim Engineering Laboratory.

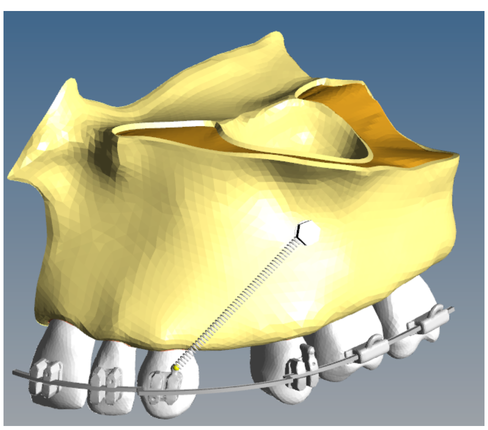


Figure 1: Vestibule view of the study model.

Regarding the determined purpose, in order to create and edit 3 dimensional mesh and making the working model more homogeneous, it was used a computer equipped with Intel Pentium® D CPU 3.00 GHz processor, 2TB Hard disk, 48 GB RAM and Windows 7, a laser scanner of NextEngine (NextEngine Inc, California, USA), CATIA (McNeel Inc, Seattle, WA, USA) and ADINA (Autodesk Inc, Pittsburgh, PA, USA).

In order to establish the working model, the upper jaw was modeled by using cone beam computer-assisted tomography of an adult without systemic disease and/or craniofacial anomaly. Computer-assisted CT images were obtained with ILUMA (3M Imtec, Oklahoma, USA) 3D scanning device with 120 kvp, 3.8 mA and 40 seconds scanning. Afterwards, the volumetric data was reconstructed with 0.2 mm thickness and transferred to DICOM 3.0 format. Imported recordings were imported into MIMICS (Able Software Corp, Massachusetts, USA). In the final stage, the upper jawbone was modeled on the MIMICS computer software, taking into account the 'interactive segmentation' working philosophy and Hounsfield values.

Periodontal ligament were obtained by giving an offset of 0.2 mm in the parts of the obtained tooth models within the cortical bone. The miniscrew with 1.6 mm diameter, 8 mm length and 0.8 mm thread distance used in the study (Figure 2), the periodontal ligament and anatomical teeth were modeled in CATIA (McNeel Inc, Seattle, WA, USA) three dimensional modeling software. (Figure 2).

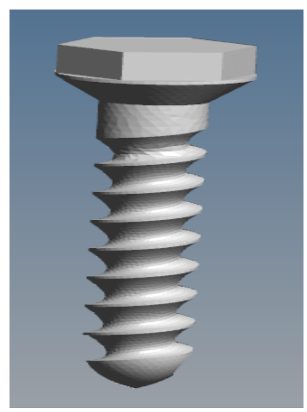


Figure 2: Image of modeled mini screw.

All models used in this study were considered linear elastic, homogeneous, and isotropic; Young's modulus and Poison ratio used are given in Table 1, the number of nodes and elements of the models are given in Table 2.

	Young Modul [5]	Poisson Ratio
Cortical bone	13600	0.3
Cancellous bone	1360	0.3
Tooth	18600	0.31
PDL	0.69	0.45
Ti6Al4V	110000	0.34
TiZr	125000	0.3
Zr	205000	0.23
Stainless Steel (SS)	210000	0.3

Table 1: Poisson ratio and Young's modulus values used in the study.

	Number of nodes	Number of elements
Upper jaw model	117 940	609 464
Mini Screw	10 893	47 444

Table 2: Node and element counts.

Determination of scenarios: Stainless steel alloy-based OMS was chosen for the treatment mechanics modeled in scenario 1. Ti6Al4V alloy based OMS was applied in scenario 2, TiZr based OMS was applied in scenario 3 and Zr alloyed based OMS was chosen in scenario 4. In all scenarios, a force of 1 N was applied to retract the canine, and all factors, except the OMS alloy, were kept constant.

Results

In Scenario 1 (Stainless Steel), it was observed that the maximum principal stress value (Pmax) obtained from tension was 1.27 MPa in the cortical bone, opposite to the force vector applied to the OMS (Figure 3a).

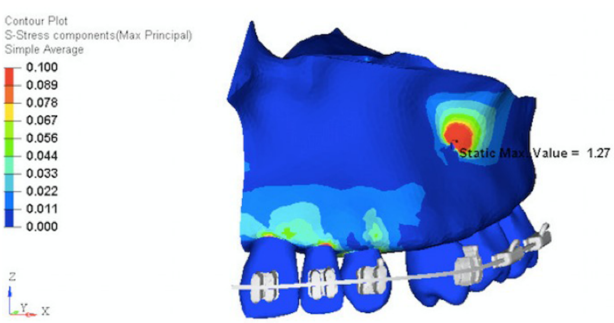


Figure 3a: Maximum principal stress value observed in cortical bone, in Scenario 1.

In scenario 1, the minimum principal stresses (Pmin) in the cortical bone were determined to occur on the side of the force vector applied to the OMS and were calculated as -1,135 MPa. When the cancellous bone in scenario 1 was evaluated, the maximum principal stress (Pmax) was found to be in the opposite direction of the force vector applied to the OMS and its value was 0.037 MPa (Figure 3b).

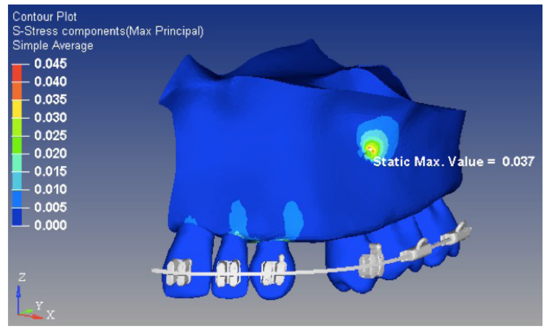


Figure 3b: Maximum principal stress value observed in cancellous bone, in Scenario 1.

In scenario 1, the minimum principal stress (Pmin) in the cancellous bone was found to be -0.044 MPa in the direction of the force applied to the OMS. In scenario 2 (Ti6Al4V), the maximum principal stress value (Pmax) obtained in terms of tension was calculated as 1.49 MPa in the cortical bone, opposite to the force vector applied to the OMS (Figure 4a).

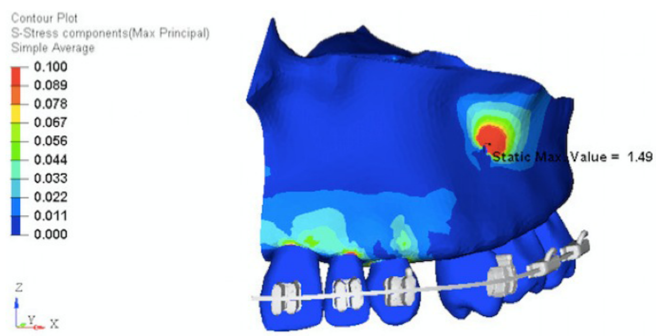


Figure 4a: Maximum principal stress value observed in cortical bone, in Scenario 2.

In scenario 2, the minimum principal stresses (Pmin) in the cortical bone occurred on the side of the force vector applied to the OMS and were calculated as -1.339 MPa. Considering the cancellous bone values in scenario 2, the maximum principal stress (Pmax) was calculated as 0.045 MPa in the opposite direction of the force vector applied to the OMS (Figure 4b).

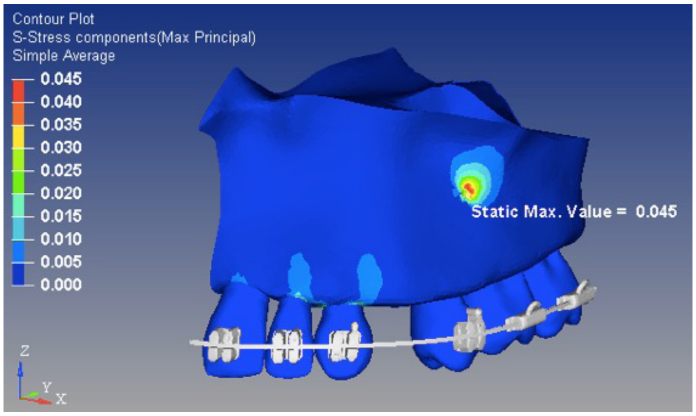


Figure 4b: Maximum principal stress value observed in cancellous bone, in Scenario 2.

When the cancellous bone in scenario 2 was evaluated, the minimum principal stress (Pmin) was found to be --0.054 MPa. In scenario 3 (TiZr), the maximum principal stress value (Pmax) obtained in terms of tension in the cortical bone, opposite to the force vector applied to the OMS, was calculated to be 1.45 MPa (Figure 5a).

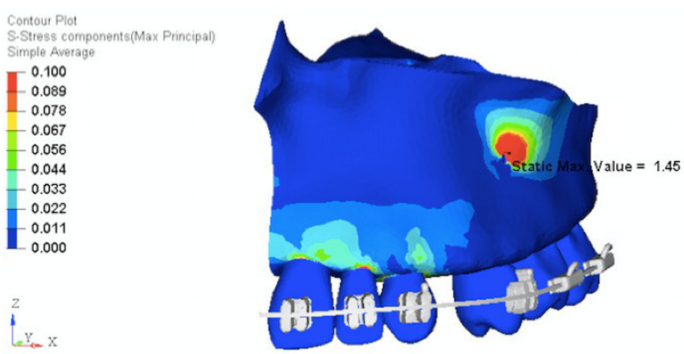


Figure 5a: Maximum principal stress value observed in cortical bone, in Scenario 3.

In scenario 3, the minimum principal stresses (Pmin) in the cortical bone occurred on the side of the force vector applied to the OMS and were calculated as -1,299 MPa. Considering the cancellous bone values in scenario 3, the maximum principal stress (Pmax) was calculated as 0.043 MPa in the opposite direction of the force vector applied to the OMS (Figure 5b).

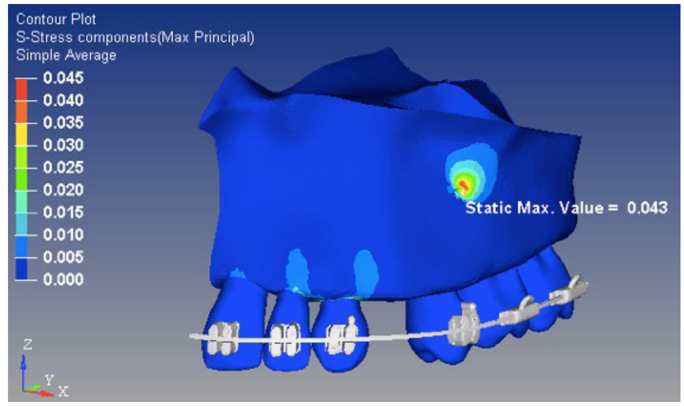


Figure 5b: Maximum principal stress value observed in cancellous bone, in Scenario 3.

When the cancellous bone in scenario 3 was evaluated, the minimum principal stress (Pmin) was found to be -0.052 MPa. In scenario 4 (Zr-based OMs), the maximum principal stress value (Pmax) obtained in terms of tension in the cortical bone, opposite to the force vector applied to the OMS, was calculated as 1.28 MPa (Figure 6a).

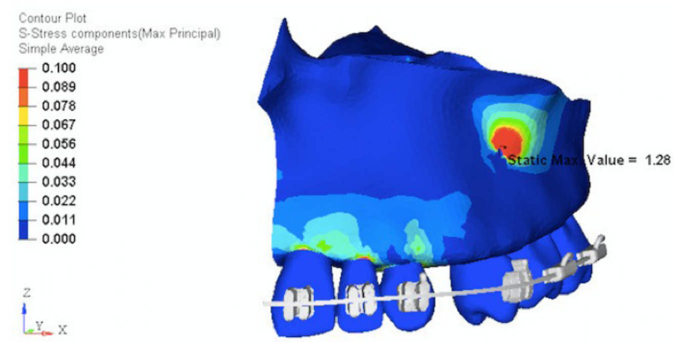


Figure 6a: Maximum principal stress value observed in cortical bone, in Scenario 4

In scenario 4, the minimum principal stresses (Pmin) in the cortical bone occurred on the side of the force vector applied to the OMS and were calculated as -1,145 MPa. Considering the cancellous bone values in scenario 4, the maximum principal stress (Pmax) was calculated as 0.037 MPa in the opposite direction of the force vector applied to the OMS (Figure 6b).

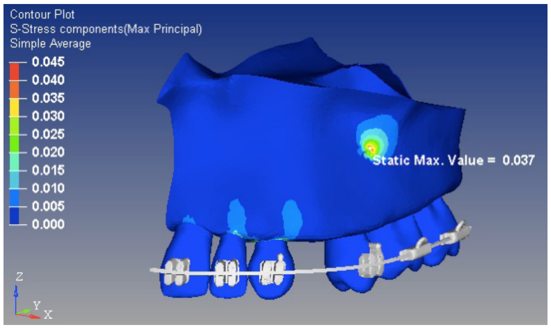


Figure 6b: Maximum principal stress value observed in cancellous bone, in Scenario 4.

The minimum principal stress (Pmin) in the cancellous bone in scenario 4 was found to be -0.044 MPa. Comparison of the findings obtained in cortical and cancellous bone is given in Table 3 and Table 4 and the von Misses results are shown in Table 5 and Graphic 1.

Principal stress in cortical bone	**Scenario 1 (Stainless Steel - 316L)**	**Scenario 4 (Zr)**	**Scenario 3 (TiZr)**	**Scenario 2 (Ti6Al4V)**
Maximum (Pmax)	1,27 MPa	1,28 MPa	1,45 MPa	1,49 MPa
Minimum (Pmin)	-1,135 MPa	-1,145 MPa	-1,299 MPa	-1,339 MPa

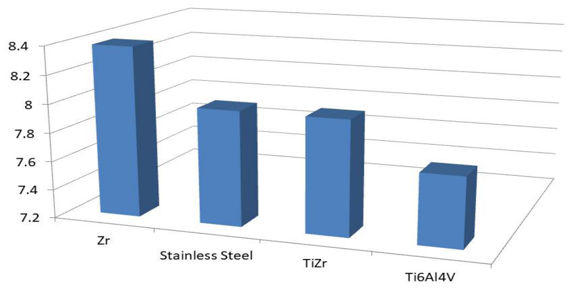
Table 3 Maximum and minimum principal stress values in cortical bone and comparison between scenarios.

Principal stress in cancellous bone	**Scenario 1 (Stainless Steel - 316L)**	**Scenario 4 (Zr)**	**Scenario 3 (TiZr)**	**Scenario 2 (Ti6Al4V)**
Maximum (Pmax)	0,037 MPa	0,037 MPa	0,043 MPa	0,045 MPa
Minimum (Pmin)	-0,044 MPa	-0,044 MPa	-0,052 MPa	-0,054 MPa

Table 4 Maximum and minimum principal stress values in cancellous bone and comparison between scenarios.

	Von Mises
Zr	8.390 MPa
Stainless Steel	8.002 MPa
TiZr	7.998 MPa
Ti6Al4V	7.687 MPa

Table 5 von Mises results of OMS.



Graphic 1: von Mises results of OMS.

Discussion

In this study, 3D finite element analysis method was used to evaluate and compare the selected scenarios and models with each other and to measure the stress in the related models in the most precise and mathematical way. Since different orthodontic anchorage methods require patient cooperation, it has been said that the most difficult part of the treatment plan is to provide appropriate anchorage with traditional methods in orthodontic and maxillofacial orthopedics treatments [6].

Controlling and/or providing anchorage is known to prevent unwanted tooth movement from occurring. However, it has been argued that the slightest change in anchorage control may cause undesirable effects and therefore, absolute/maximum anchorage should be provided [7].

OMS-supported skeletal anchor units, introduced in recent years, have been described as one of the greatest developments in the orthodontic literature. It has been reported that OMS have features such as expanding the range of motion of the teeth and not needing patient cooperation orthodontically [8].

OMS has been observed to be widely used in providing skeletal anchorage [9]. When looking at the reasons for the widespread use of OMSs, it is stated that they are generally simple to implement, low cost and easy to remove [10].

In today's orthodontic practice, OMS are available in different sizes (diameter and length), depending on the area to be applied, from 1.2 mm to 2.0 mm in diameter, and from 6.0 mm to 12 mm in length. reported to have changed [11].

In the orthodontic literature, there is only one study evaluating different OMS alloys with FEA. Singh, et al. [4] In their research published in 2012, evaluated the stresses arising around these OMS by applying horizontal and torsional loading to stainless steel and Ti6Al4V based OMS. For this purpose, it was reported that OMS with a length of 10.62 mm and a diameter of 2.48 mm were selected; They said that they applied a force of 350 g in the horizontal direction and 400 g in the torsional direction. Contrary to the findings of this study, the related authors reported that they detected a high stress around the stainless steel OMS (19.56 MPa), compared to the Ti6Al4 OMS (11.35 MPa). It is thought that this difference may be due to the chosen scenario, the force vector and amount applied to the applied OMS, the number of elements and nodes of the model. It has been determined that the stresses occurring in OMS occur in the neck part, similar to our study.

Kuroda, et al. [12] in their study, evaluated the stresses occurring in the mini-screw and alveolar bone during orthodontic treatment with the finite element method. The hypothesis of this study was that if the size of the OMS outside the alveolar bone is reduced, it will be more successful against orthodontic loads. For this purpose, it was seen that OMS of 4 different sizes (12 mm, 8 mm, 10 mm and 6 mm) were applied vertically to the geometric figures obtained. Similar to this study, it was reported that tensions were detected in the cortical bone and neck of the OMS.

Alrbata, et al. [13] In their finite element study, determined the optimal limits of the loads on orthodontic micro-implants. For this purpose, they stated that they applied 0.5, 1.0, 1.5, 2.0, 2.5, 3.0, 3.5 and 4.0 N forces to the micro-implants in the horizontal direction. The authors proved that the amount of force that can be applied without damaging the micro-implants should be between 3.75 and 4.5 N at most. The 1 N force chosen in this study both stimulated the retraction mechanics of the selected clinical canine tooth and was observed to be within optimal limits.

Popa, et al. [14] In their in vitro and in vivo studies, evaluated the primary stability of orthodontic mini-implants. For this, cortical bone thickness and the application angle of the orthodontic mini-implant were examined. Mini-implants with application angles of 30°, 60°, 90° and 120° were applied to bone models with cortical bone widths of 1, 1.5 and 2 mm and it was investigated which angle caused the least tension. In cases where the appropriate amount of cortical bone (2 mm) is available, a 90° application angle is recommended to prevent necrosis of the orthodontic mini-implant, micro-crack in the cortical bone and failure of the mini-implant. In this study, all OMS were applied with 90°.

At different times, Kobayashi, et al. [15] and Grandin, et al. [16] investigated TiZr alloys and reported that they provide mechanically superior performance compared to pure titanium. In 2009, Bernhard, et al. [17] compared TiZr implants with titanium implants and found that TiZr alloys showed 40% more strength in terms of fatigue strength and tensile stresses.

Ikarashi, et al. [18] stated that they found superiority over titanium in their studies investigating the biocompatibility of TiZr alloy.

It has been reported that the Ti-6Al-4V alloy, which is used in different fields, is biocorroded *in vitro* [19]. Although orthodontic mini screws are used for a shorter time than joint prostheses, Ti-6Al-4V alloy used in the production of mini screws is also a corrosion sensitive alloy. Mouhyi, et al. [20] reported that titanium ions released as a result of corrosion triggered peri-implantitis and affected the success of the mini-screw.

According to the Pearson correlation analysis of the findings of this study, a strong and negative correlation was found between Young's modulus and the amount of tension (in cortical and cancellous bone) ($r=-0.996$, $r=-0.998$ $p<0.01$, respectively). In other words; It was found that while the Young's modulus value increased, the tension in the bone decreased from a mathematical / statistical point of view, regardless of the reasons.

Ho, et al. [21] investigated the structure, mechanical properties and machinability of TiZr alloy applied in the field of dentistry in their study in 2008. As a result of the their study, they found that the TiZr alloy was 5.5 times more machinable than the commercial pure titanium alloy.

Brizuela-Velasco, et al. [22], in their study examining the mechanical properties and biomechanical behaviors of TiZr alloy compared to Ti6Al4V, found no difference between the two alloys in terms of stress and deformation around the implant, but found, in parallel with our study, that the stress value was less in TiZr alloy.

Altuna, et al. [23], on the other hand, focused on the clinical evidence of TiZr dental implants in their meta-analysis study published in 2016 and stated that TiZr-based narrow implants were 95% more successful compared to other narrow implants.

In the orthodontic literature, there are different opinions about the forces required for distalization of the canine; Lee, et al. [24] suggested a force of 150 to 260 g, while Retain, et al. [25] suggested a force of 250 g. In their study published in 2000, Iwasaki, et al. [26] stated that forces between 18 and 60 g would be sufficient without causing any side effects in the distalization of the upper canine.

Conclusion

The results of the finite element study carried out to determine the tension in the upper jaw caused by orthodontic mini screws based on different biomaterials are as follows:

The maximum principal stress values (Pmax) obtained in the cortical bone were 1.27 MPa (scenario 1 - stainless steel), 1.28 MPa (scenario 4 - Zr), 1.45 MPa (scenario 3 - TiZr), and 1.49 MPa (scenario 2 – Ti6Al4v), respectively, from smallest to largest.

The minimum principal stress values (Pmin) determined in the cortical bone, from smallest to largest, were -1.135 MPa (scenario 1 - stainless steel), -1.145 MPa (scenario 4 - Zr), -1.299 (scenario 3 - TiZr) and -1,339 MPa (scenario 2 - Ti6Al4V);

The highest maximum prime (Pmax) stress values obtained in cancellous bone were 0.037 MPa (scenario 1 - stainless steel), 0.037 MPa (scenario 4 - Zr), 0.043 MPa (scenario 3 - TiZr), and 0.045 MPa (scenario 2), respectively, from smallest to largest. - Ti6Al4V).

The highest minimum prime (Pmin) stress values obtained in cancellous bone, from smallest to largest, were -0.044 MPa (scenario 1 - stainless steel), -0.044 MPa (scenario 4 - Zr), -0.052 MPa (scenario 3 - TiZr), and -0.054. MPa (scenario 2 - Ti6Al4V).

In terms of von Mises findings of OMS obtained from different biomaterials included in the study, were calculated respectively from largest to smallest: 8,390 MPa in Scenario 4 (Zirconium), 8,002 MPa in Scenario 1 (stainless steel); 7,998 MPa in Scenario 3 (TiZr); 7,687 MPa in scenario 2 (Ti6Al4V).

Considering both the negative biocompatibility properties of Ti6Al4V alloy and TiZr biological and mechanical superior properties, TiZr alloy might as well be considered as an alternative for OMV production.

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