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ORIGINAL RESEARCH

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Miniscrew Composition, Transmucosal Profile, and Cortical Bone Thickness: A Three-dimensional Finite-element Analysis of Stress Fields

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ABSTRACT

Aim: The aim of this study was to use the finite-element method (FEM) to analyze the stress fields generated in miniscrews (MSs) and surrounding bone on applying a force perpendicular to the MS according to variations in the cortical bone thickness and changes in the transmucosal profile length and MS composition.

Materials and methods: Miniscrews with stainless steel (SS) and titanium alloy mechanical properties with a 1 to 2 mm transmucosal profile inserted in bone blocks with cortical bone of varying thickness (1 and 2 mm) were three-dimensionally modeled using computer-aided design (CAD) and examined using FEM. A 3.5 N force perpendicular to the long axis of the MS was applied in the four mechanical tests: EM1: SS MS and a 1 mm transmucosal profile; EM2: titanium MS and a 1 mm transmucosal profile; EM3: SS MS and a 2 mm transmucosal profile; and EM4: titanium MS and a 2 mm transmucosal profile.

Results: The stress distributions in all mechanical tests were highest at the MS, especially at the MS–cortical bone interface. A greater stress concentration occurred in cortical bone measuring 1 mm thick than in the cortical bone measuring 2 mm thick. The MSs with a 2 mm transmucosal profile showed higher stress than those with a 1 mm transmucosal profile.

Conclusion: The titanium alloy MSs showed higher stress fields and deflection voltages than the SS MSs at the same cortical bone thickness and with the same transmucosal profile.

Clinical significance: From a mechanical perspective, this study showed the stress field generated in MSs with SS and titanium alloy (Ti) mechanical properties and surrounding bone. The stress distribution was concentrated at the MS, mainly at the interface with the cortical bone, and the difference between the stress values for the Ti and SS MSs was very small. Under this condition, the two types of MSs are suitable for orthodontic applications because their yield limits are much higher.

Keywords: Dental implants, Finite-element method, Orthodontic anchorage procedures.

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INTRODUCTION

The MSs have different sizes, shapes, diameters, and transmucosal profiles, and these variations may affect their performance. The MS form must provide mechanical locking into bone, which directly influences the implant stability. It should also ensure that the load distribution is not detrimental to the bone physiology and should limit trauma beyond that occurring during insertion.[1] The amount and quality of bone in which MSs are installed is another influential factor, as the implant can fail to osseointegrate.[2]

The primary stability of a MS is affected mostly by factors related to the screw and the patient.[3,4] Morphological characteristics, such as the screw diameter and length, are among the factors related to the MS.[5] The quality and quantity of the bone in which the MS is to be inserted are among the patient factors. The thickness of the cortical bone is a patient factor that can interfere with the primary stability of the MS.[5] Areas with thicker cortical bone are considered more stable for MS placement.

This bone density and thickness can vary between individuals and MS regions.[4]

Various clinical scenarios can be simulated using the FEM to evaluate the stress distribution in the areas surrounding the bone. Overload on the peri-implant bone region can result from high concentrations of stress, and some studies[4] suggest that the deformation-related fields stimulate bone resorption in the region, thereby undermining the effectiveness of the implant.[6] The assessment of the stress field distribution in the bone enables investigation of the effectiveness of endosseous implants and reveals the risk of implant failure.[7] Some studies have reported the stress field distribution in the bone surrounding the MS.[6,8]

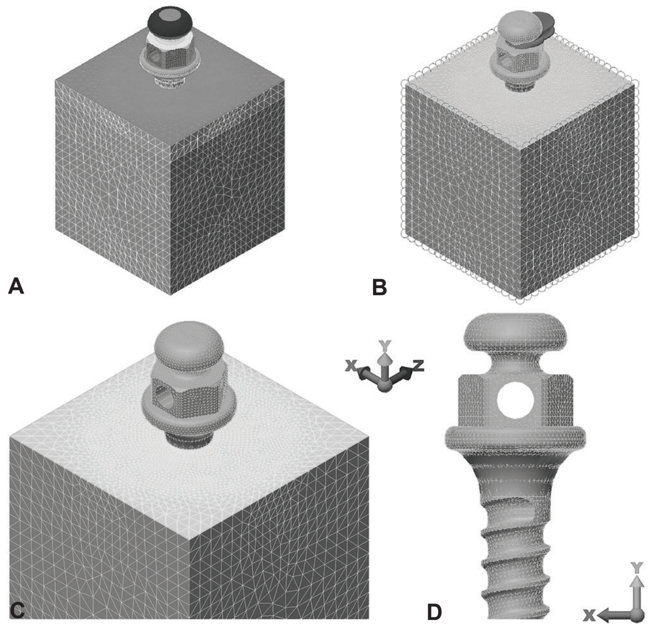
The objective of this study is to analyze, using the FEM, the stress fields that are generated when a perpendicular force is applied to the MS and surrounding bone, according to the changes in the transmucosal profile length (1 and 2 mm), variations in the cortical bone thickness (1 and 2 mm), and MS composition (titanium alloy and SS F138).

MATERIALS AND METHODS

Miniscrews (Morelli, SP) measuring 1.5 mm in diameter and 8 mm in length with a 1 and 2 mm transmucosal profile (Morelli 37.10.102 and 37.10.202) were geometrically simulated with a commercial software package for CAD, SolidWorks® 2015, academic version. Solid models of the MSs with dimensions provided by manufacturer were created using the software. The MSs used in this study were made of a titanium alloy (Ti_6Al_4V) (ASTM F136) with 6% aluminum (Ti) and a SS alloy (ASTM F138) with 18% chromium + 14% nickel + 2.5% molybdenum (SS).

All computational MS models were inserted into a cubic block (9 mm in height and 8 mm width and depth) and were modeled three-dimensionally using the same software. The blocks were built in two layers that represent the cortical and the trabecular bone (Fig. 1A). The cortical bone layer had a fixed thickness of 1 mm. The interface between the bone and MS was considered a perfect contact. After the CAD step, the solid model was transferred to a software program that used FE called Autodesk Simulation Mechanical® 2015. An FE mesh with linear tetrahedral elements was generated in this step. The edge lengths of the elements varied between 0.018 and 0.62 mm (Figs 1B to D). These values were defined after a convergence analysis of the stress field. The convergence analysis ensures that the results of the FEM analyses are accurate.[9]

The models displayed higher mesh refinement in the 3 mm bone region surrounding the MS. Refinement is used to correctly represent the MS thread profile. The mesh properties of the FE models are shown in Table 1. The translations were restricted in the x, y, and z directions as boundary conditions on the side faces of each cubic block. A 3.5 N load was applied to the MS head directed along the z-axis, which is represented by green lines (Fig. 1B). This force magnitude is realistic and applied clinically.[10] In addition, the mechanical properties of all materials involved in the FEM analysis and all other materials (cortical bone, cancellous bone, Ti MS, and SS MS) were assumed to have homogeneous isotropic and linear elastic behavior with a specific Young's modulus (E)[11-13] and Poisson's ratio (ν) as shown in Table 2.



Figs 1A to D: Finite-element models

The stress fields over the FE models were assessed according to the maximum distortion energy theory,[14] and have been used previously in orthodontics by other researchers.[15-18] The stress field in the cortical bone was assessed and compared among the models. The simulations were divided into two length groups (1 and 2 mm) and subjected to various MS and transmucosal profile material combinations (Table 1).

Table 1: Geometric model and FE model

Load case	*MS material*	*MS transmucosal profile length (mm)*	*Cortical bone thickness (mm)*	*Number of elements*	*Number of nodes*
L1	SS	1	1	1,932,406	357,555
L2	Ti	1	1	1,932,406	357,555
L3	SS	2	1	2,647,395	481,500
L4	Ti	2	1	2,647,395	481,500
L5	SS	1	2	1,932,406	357,555
L6	Ti	1	2	1,932,406	357,555

Table 2: Material properties used in FEM

Material	*E = Young's modulus (MPa)*	*v = Poisson's ratio*
Titanium alloy	110,000a	0.33a
Stainless steel alloy	205,000b	0.29b
Cortical bone	13,800c	0.26c
Cancellous bone	345c	0.31c

[a]Suzuki et al[13]; [b]Kojima and Fukui[12]; [c]Jones et al[11]

RESULTS

The stress distribution in the bone and in the MSs is shown according to a color scale. The cool color regions indicate low stresses, and hot color regions indicate high stresses. Figure 2 shows the results of the stress field for the six assessments. Higher stresses emerged at the contact surfaces between the cortical bone and the MS. The highest values of von Mises stress were found in L1 (35.47 MPa), L2 (30.22 MPa), L3 (51.87 MPa), L4 (42.61 MPa), L5 (37.60 MPa), and L6 (30.62 MPa). For all results, the stress concentration was located within the cortical bone only and did not extend into the trabecular bone. Figure 3 shows the stress field amplified in the cortical bone region without the MS. In this figure, the first eight threads of the MS are represented by a thin black line.

To show the stresses field on cortical and trabecular bone in a more accurate manner, an additional analysis was performed. In this analysis, the principal stress tensor in z-direction was assessed in the same direction of the 3.5 N load (Fig. 4). In these results, the blue and red regions represent the compressive and tensile stresses respectively. The average value of the higher stresses (compressive; tensile) found for the six analyses were L1 (–18.55 MPa; +21.23 MPa), L2 (–22.49 MPa; +24.08 MPa), L3 (–22.50 MPa; +24.57 MPa), L4 (–26.00 MPa; +29.60 MPa), L5 (–17.57 MPa; +20.64 MPa), and L6 (–21.85 MPa; +25.29 MPa).

In all test results, we noted that stresses were increased only in cortical bone due to the bending load applied to the z-axis. This occurred because the flexural stiffness of the cancellous bone is 40 times smaller than that of cortical bone and therefore, does not produce resistance

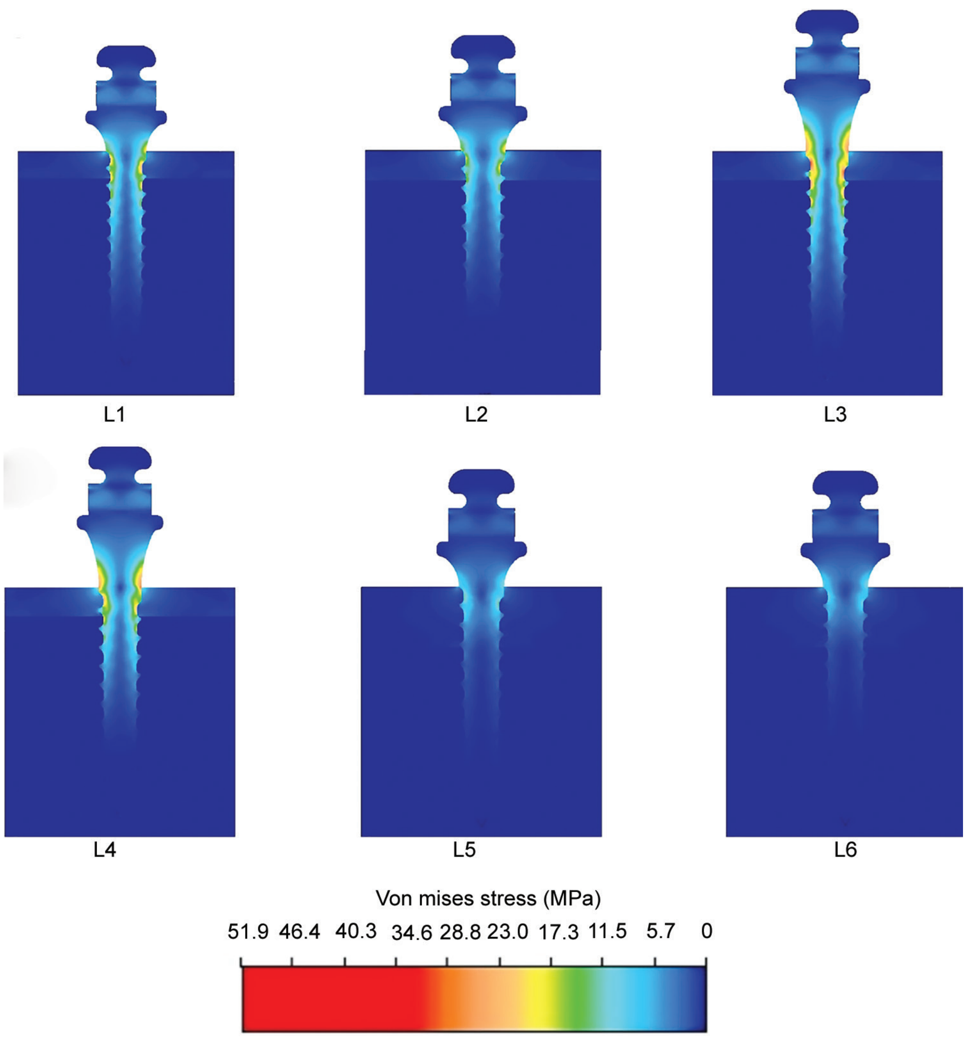


Fig. 2: von Mises stress field determined by FEM analyses

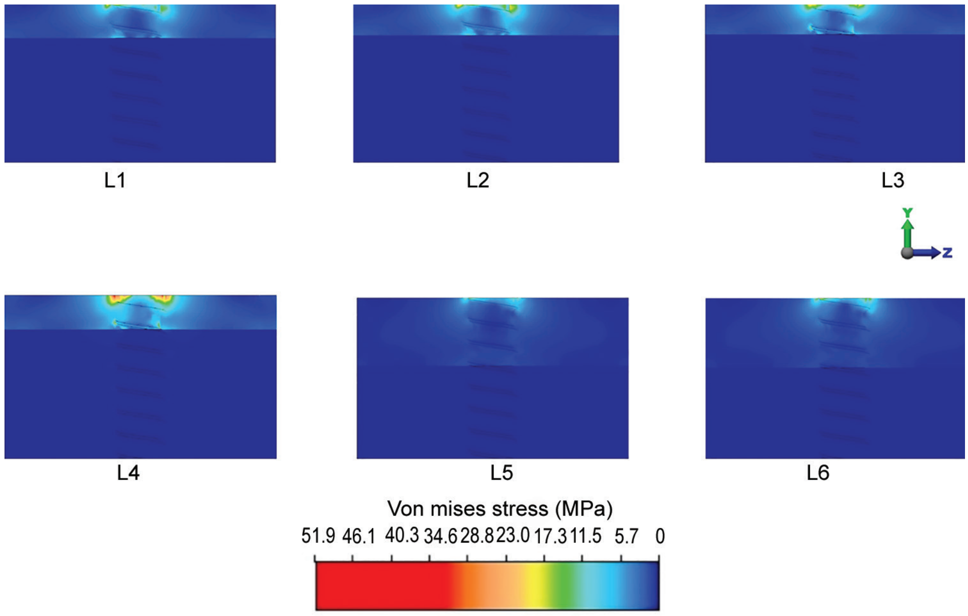


Fig. 3: von Mises stress field amplified in the cortical bone region

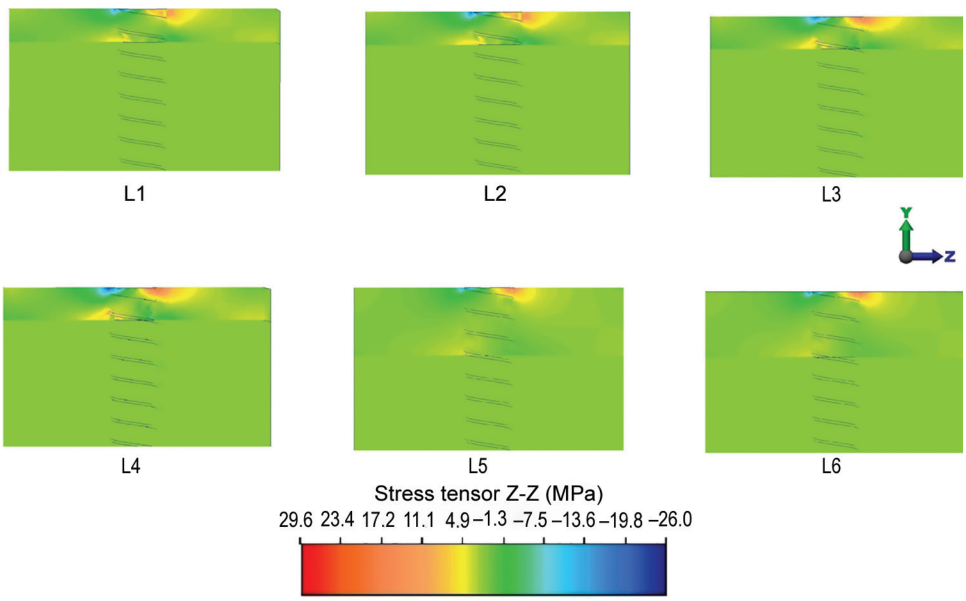


Fig. 4: Principal stress tensor z-z in the cortical bone region

to the applied load. Consistent with classical solids mechanics, the flexural stiffness is directly proportional to the longitudinal elastic modulus of the materials, i.e., Young's modulus.

As expected, the highest von Mises stresses on the cortical bone were observed in L3 and L4 (Fig. 3). These high stresses occurred because the flexural moment is directly proportional to the distance between the application of the bending load on the MS and the trabecular bone. In these load cases, the transmucosal profile of the MS had a length of 2 mm.

In all analyses, higher stresses were produced on cortical bone when the Ti MS was used. These results occurred because the Ti MS has a lower stiffness value than the SS MS; therefore, Ti MSs are subjected to greater strain (Fig. 5). The strain tensor is observed only in the z-direction for the MS and cortical bone. The deflection of the two components in the figure is purposely increased to show the strain field.

The strain is transmitted to the cortical bone, which has a lower stiffness value, and due to the generalized Hooke's law, this effect results in relatively higher stresses [Eq. (1)].[19] In addition, Equation (1) assumes that the materials are homogeneous and isotropic, i.e., they have the same properties in all directions as stated in the previous section.

$$\sigma_{zz} = \frac{E}{(1+v)(1-2v)}\left[v\epsilon_{XX} + v\epsilon_{YY} + (1-V)\,\epsilon_{ZZ}\right] \quad (1)$$

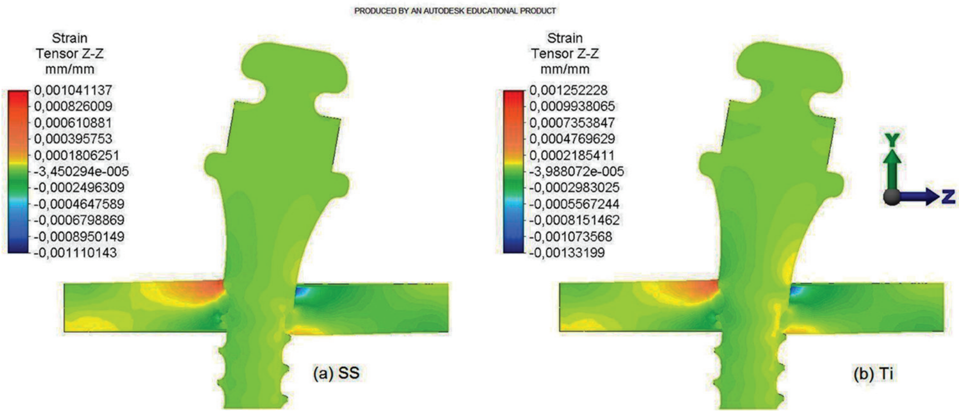


Fig. 5: Strain tensor z-z in the cortical bone region

where σ_{zz} is the stress in the z-direction, and ϵ_{xx}, ϵ_{yy}, and ϵ_{zz} are the strains in the x, y, and z directions respectively.

DISCUSSION

This study evaluated the stress distribution and magnitude in the bone and MS induced by a load applied to an orthodontic MS. These results contribute to the understanding of the biological reactions resulting from the system of loads applied to an MS.

Some FEM studies have shown that applying a lateral load to an MS causes most of the stress to concentrate in cortical bone areas.[5] The present study found similar results, as the highest stress concentrations were observed in the cortical bone–MS interface. Previous studies seeking to improve the primary stability of MSs have focused on the bony anatomy, namely the thickness of the cortical bone, and not on the quality of the trabecular bone.[20]

The transmucosal profile is designed to maintain the health of peri-implant tissues, especially in areas with minimal attached gingiva, because inflammation is a contributing factor to MS failure.[21] The results showed that the maximum von Mises stress was significantly higher in MSs with a 2-mm transmucosal profile than in those with a 1 mm transmucosal profile.

No studies have compared MSs according to the transmucosal profile thickness. However, the results reflect the principle of a bending moment caused by a load. The load moment is the effect produced on a body by a load applied at a relatively distant point on the line of action of this force. This load generates a trend of rotational movement on the transversal axis of the body.[22] Following this principle, a load applied further away will result in a higher moment and, consequently, increased stress.

Thus, an MS with a 2 mm transmucosal profile has a longer lever arm than an MS with a 1 mm transmucosal profile. Application of the load in the z-direction generated a greater moment in the MS with a 2-mm transmucosal profile and, consequently, higher bending stress. In another study performed to assess different MS lengths (8, 10, and 12 mm), Lin et al[8] concluded that a longer MS length was associated with higher stress on the cortical bone. Moreover, Nova et al[23] compared two MS brands with and without a transmucosal profile and concluded that the presence or absence of the transmucosal profile did not affect the insertion or removal torques because only one device (NEODENT) exhibited significant torque, while the torque in the other device (SIN) did not vary.

The present results have shown higher stress in the numerical analyses using a 1 or 2 mm cortical bone thickness, independent of the MS material. In the assays, where the cortical thickness was 1 mm, stress was concentrated 1 mm above the cortical bone and 1 mm below in the trabecular bone. In the 2 mm thick cortical bone, this stress concentration occurred along the entirety of the cortical bone and did not extend into the trabecular bone.

The magnitude of the von Mises stresses was higher in cortical bone with a 1-mm thickness, as shown in Figure 3. This occurred because the stresses are distributed over a smaller surface contact area between the MS and the cortical bone than in cortical bone with a 2 mm thickness.

To adapt the MS to different insertion sites and achieve greater primary stability, dentists use various MS sizes and types. Deformation and MS fractures can be avoided through a better understanding of the influence of these factors on the mechanical properties of the implant.

The force application axis used on most MSs in orthodontic mechanics is the axis perpendicular to the device. Therefore, MS deformation should be evaluated by applying force in this direction, as performed in the present study.[24] The largest deformation was observed in the MS head, and SS screws had higher offset values than Ti devices. The greatest strain occurred in the FE analysis using an MS with a 2 mm transmucosal profile compared with the 1 mm transmucosal profile. In this study, the MS strain was minimal.

Two different materials were selected for the MSs in this study: SS and titanium alloy. The SS used for implants in human tissues, especially in the oral cavity, must resist corrosion caused by exposure to body fluids because this type of corrosion can harm the patient and promote device fracture and other forms of treatment failure.[25] Titanium alloy implants have several advantages, including high strength and a low modulus of elasticity, compared with SS alloys, as well as approximately 30% more resistance to fatigue.[25] The present study revealed higher stresses and strain in the Ti than in the SS MSs. The magnitude of the stress field was higher in Ti MSs because their rigidity is approximately 47% less than that of SS MSs. As stated in the previous section, the bending stiffness is directly correlated with the magnitude of the Young's modulus.[19] The present results showed that the maximum von Mises stress occurred in the Ti MS. However, the difference between the stress values for the Ti and SS MSs was very small. Under this condition, the two types of MSs are suitable for orthodontic applications because their yield limits are much higher. The yield limits for the Ti and SS MS are 795 and 434 MPa respectively.

Although an anchoring MS is inserted into both the bone and soft tissue, the bone portion maintains the necessary force against orthodontic forces. Studies aided by computer engineering using the FEM allow visualization of the stress distribution in mechanical simulations, thereby allowing clinicians to apply orthodontic mechanics with greater certainty and predictability.

CONCLUSION

- The stress distribution was concentrated at the MS, mainly at the interface with the cortical bone.
- A greater stress concentration occurred in cortical bone measuring 1 mm in thickness than in cortical bone measuring 2 mm in thickness.
- Miniscrews with a 2 mm transmucosal profile displayed higher stress than those with a 1 mm transmucosal profile.

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