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Insertion angle of orthodontic mini-implants and their biomechanical performance: finite element analysis

Ângulo de inserção de mini-implantes ortodônticos e seu desempenho biomecânico: análise de elementos finitos

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Resumo

Objetivo: O objetivo deste estudo foi avaliar as tensões e deformações de duas marcas comerciais de mini-implantes ortodônticos geradas após a aplicação de dois tipos de forças (de tração de 200 gf e torção de 20 N.cm) inseridos em duas angulações (45° e 90° em relação ao osso cortical). **Material e método**: Modelos tridimensionais das duas marcas de mini-implantes (SIN - Sao Paulo, Brasil, e RMO – Coréia do Sul) foram construídos e analisados por análise de elementos finitos (FEA). As análises foram realizadas em simulações no osso cortical, osso esponjoso e no parafuso. **Resultado**: A análise FEA mostrou que os mini-implantes da marca RMO apresentaram maior deformação elástica quando submetidos à tração e as forças de torção quando comparado aos mini-implantes da marca SIN. Em ambas as marcas testadas, e para os diferentes ângulos de inserção, houve uma maior deformação do osso cortical, com maior tensão localizado no mini-implante. A tensão no mini-implante foi localizado na região do perfil transmucoso. **Conclusão**: Ao comparar as análises de elementos finitos das duas marcas comerciais de mini-implantes, concluiu-se que um maior número de roscas e maior inclinação resultam em menor resistência à deformação e induzem uma maior tensão no osso cortical quando submetidos à forças de torção e tração, especialmente quando inserido em um ângulo de 45º com o osso cortical.

Descritores: Procedimentos de ancoragem ortodôntica; ortodontia; análise de elementos finitos.

Abstract

Objective: The aim of this study was to assess the stresses and strains generated after the application of two types of forces (traction of 200 gf and torsion of 20 N.cm) in two types of orthodontic mini-implants inserted at different (45° and 90° to the cortical bone) angles. **Material and method**: three-dimensional models of two brands of mini-implant (SIN – Sao Paulo, Brazil, and RMO – South Korea) were exported and analyzed by finite element analysis (FEA). Analyses were performed on simulations of cortical bone, cancellous bone and the screw. **Result**: FEA analysis showed that RMO mini-implants had greater elastic deformation when subjected to tensile and torsional forces when compared with SIN mini-implants. For both trademarks and insertion angles tested, there was greater cortical bone deformation, but with the greatest strain located on the mini-implant. Tension on the mini-implant was located in its transmucosal profile region. **Conclusion**: When comparing the two brands of mini-implants by FEA, it is fair to conclude that that the larger number of threads and their greater angle of inclination resulted in less resistance to deformation and induced a higher level of tension in the mini-implant and cortical bone when subjected to forces, especially when inserted at an angle of 45º to the cortical bone.

Descriptors: Orthodontic anchorage procedures; orthodontic; finite element analysis.

INTRODUCTION

The use of orthodontic mini-implants (OMI) has changed many concepts in orthodontics[1,2]. The possibility of moving teeth without causing unwanted side-effects has allowed the treatment of more complex cases, and added predictability of the outcome of treatment plans[1,2]. To reduce the failure rate of OMIs (10-30%), several biological and biomechanical factors associated with their performance have been studied[3-5]. From the literature, it can be speculated that the design of the OMI, as well as the inclination angle in which it was inserted, are factors of major relevance[3-5].

One of the methodologies that can be used to assess the biomechanical impact of possible changes in the design and in the insertion angle of OMIs is the finite element analysis (FEA). FEA is a methodology that relies on the "discretization" of large structures to smaller elements of known size, which are net-connected and capable of being mathematically interpreted[6-8]. The main advantage of using FEA to evaluate the biological performance of an OMI is that this methodology is based on a virtual environment, saves time, costs, and avoids the use of animals[6-9]. The literature has shown a positive correlation between the biomechanical characteristics of OMIs and their *in vivo* performance[10].

Previous studies using FEA have shown the quality of trabecular bone in the region of OMI insertion is not critical for its stability, however, a layer of cortical bone at least 1 mm thick is[11]. On the other hand, the thinner the cortical bone, the larger the tension the OMI will cause to the bone, increasing the risk of implant failure[12-14]. It has been shown that, Even the thick cortical bone layer of the palate region has been shown to be incapable of supporting forces greater than 480 gf[13]. The shape and geometry (macrostructure) of an OMI are directly connected with the amount of tension it will induce into the bone[15-17]. The region in which the highest tensions to the cortical bone are detected is the first thread of the OMI[18,19]. Furthermore, the angle at which the OMI is inserted might alter this parameter[3].

To provide a better understanding of the parameters related to OMI success, the aim of this study was to assess the tensions and deformations (stresses and strains) generated after the application of two types of forces (traction and torsion) in two types of OMIs inserted at different angles.

MATERIAL AND METHOD

Two brands of OMIs were tested: SIN (Implant System, São Paulo/SP, Brazil) and RMO (Rocky Mountain Orthodontics, Seoul, South Korea). They were both self-drilling, measured 1.6 mm in diameter and 6 mm long. Further details of implants tested are shown in Table 1.

Tridimensional models of the implants were constructed using dedicated software (Autodesk Inc., San Raphael, CA - USA), based on reverse engineering: the implants were measured under a stereomicroscope, and the measurements established were used to construct their 3D models. Two simplified bone blocks were also constructed using the same software, and they simulated of two layers of bone: one of cortical bone (1.5 mm thick), and one of trabecular bone (18.5 mm thick). The interface between the two layers of bone was considered a continuous line. Tridimensional models were then exported and analyzed by a single trained operator using FEA dedicated software (ANSYS Workbench 14.5, Swanson Analysis Systems, Canonsburg, PA - USA). All materials were considered as homogeneous and isotropic, linearly elastic, and the biomechanical properties attributed to them are shown in Table 2[20-24]. Assessments were made considering the implant already inserted and integrated into the bone[12,13,16,18,25].

OMIs were tested under traction (200 gf) and torsion (20 N.cm) forces (commonly used for orthodontic movement and insertion of the mini-implant respectively), in addition to considering insertion in two directions: at angles of 90° and 45° to the cortical bone. Traction forces were applied in the opposite direction to OMI inclination. Assessments were made at the level of cortical bone, trabecular bone, and OMI, separately.

RESULT

Tables 3 and 4 show the tensions and deformations found for the conditions tested. For all tested setups, the highest tension was seen in the OMI, and the largest deformation was seen in the cortical bone (as shown in Figures 1-4). The worst results were observed when the inclination of 45 degrees was tested. Deformation was usually larger for RMO implants. Tension was higher for SIN implants after the use of traction forces, while it was higher for RMO implants after the use of torsion forces.

DISCUSSION

The aim of this study was to assess the tensions and deformations (stresses and strains) generated after the application of two types of forces (traction and torsion) in two types of OMIs inserted at different angles, by means of FEA. Not only should the limitations of the selected method of analysis (the homogeneity of the considered bone blocks and the absence of biological-triggered response being

Table 1. Implant characteristics

	SIN	RMO
Number of threads	12	13
Distance between threads (mm)	0.7	0.3
Thread inclination (degrees)	132.7	162

Table 2. Tested materials properties

	Elasticity modulus	Poisson Coefficient
Cortical bone[26,27]	13.7 MPa	0.3
Trabecular bone[26,27]	1.3 MPa	0.3
Ti_6Al_4[26,27]	113.8 GPa	0.3

the most relevant) be borne in mind, but also its well-known importance in precisely identifying and quantifying stresses and strains in pre-defined (and validated) models[15,24]. The loading protocol was identical for the two brands of OMI tested, in order to minimize the risk of bias. Furthermore, it could be speculated that the differences found are mainly related to the macroscopic features (design) of the tested OMIs, since both were fabricated of the same metal alloy (Ti_6Al_4Va).

Table 3. Maximum tension (MPa) and deformation ($x10^{-5}$, mm/mm) values assessed following the use of traction forces

	Tension	**Deformation**						
	SIN 45^0	**SIN 90^0**	**RMO 45^0**	**RMO 90^0**	**SIN 45^0**	**SIN 90^0**	**RMO 45^0**	**RMO 90^0**
Cortical bone	0.91	0.54	1.67	0.57	8.11	4.21	0.13	4.30
Trabecular bone	0.0048	0.0048	0.0037	0.0025	5.24	3.61	2.87	1.90
OMI	3.51	2.16	2.83	1.84	3.16	1.94	5.20	1.90

Table 4. Maximum tension (MPa) and deformation ($x10^{-3}$, mm/mm) values assessed following the use of torsion forces

	Tension	**Deformation**						
	SIN 45^0	**SIN 90^0**	**RMO 45^0**	**RMO 90^0**	**SIN 45^0**	**SIN 90^0**	**RMO 45^0**	**RMO 90^0**
Cortical bone	245.65	93.69	296.05	132.59	18.50	6.94	21.86	9.85
Trabecular bone	4.69	2.92	4.70	3.14	3.43	2.24	3.44	2.39
OMI	542.44	186.68	769.08	381.42	4.89	1.81	7.20	3.71

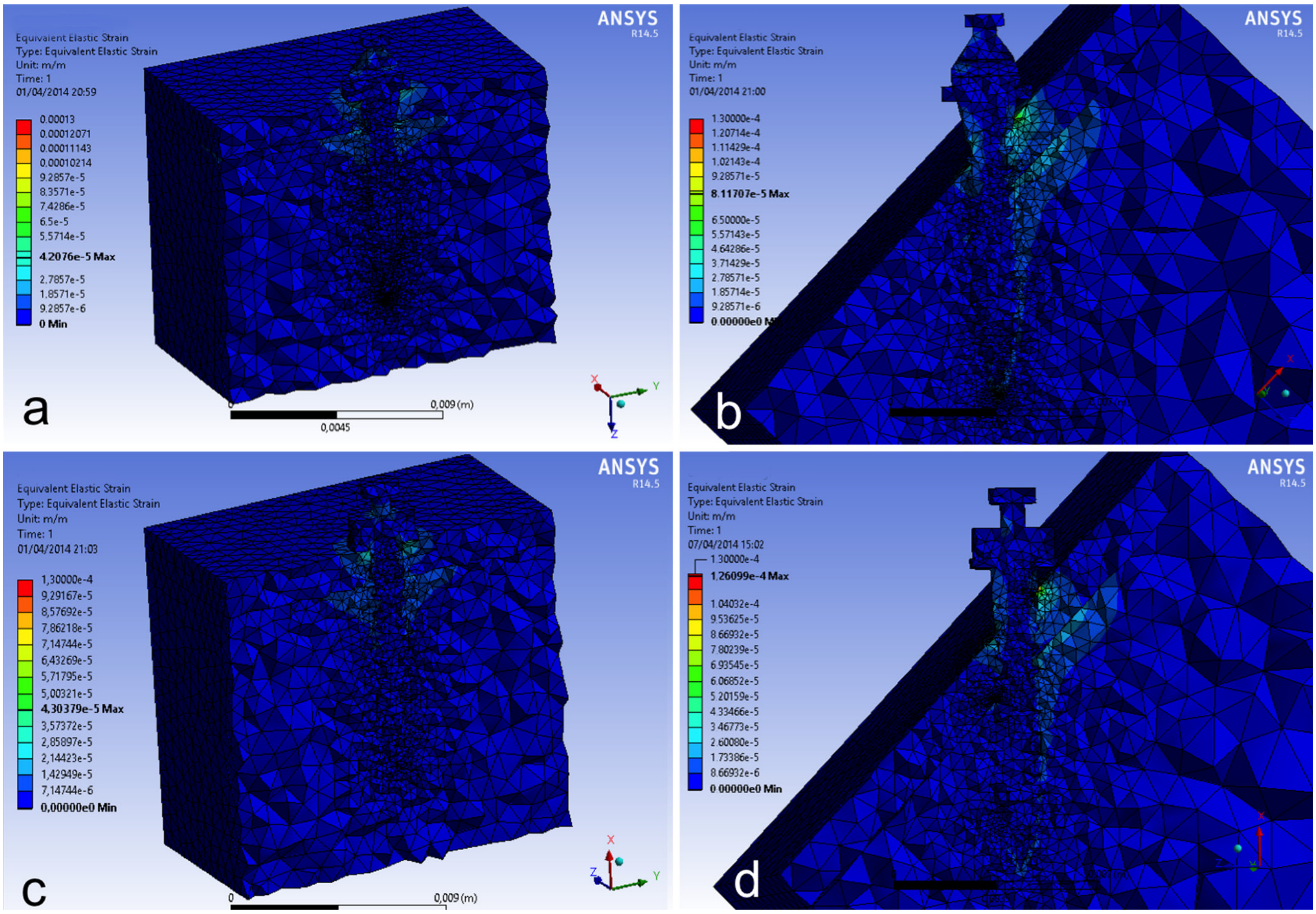

Figure 1. Maximum deformation (mm/mm) of the OMI when inserted in the bone block, under a 200 gf traction force. a SIN 900, b SIN 450, c RMO 900, d RMO 450.

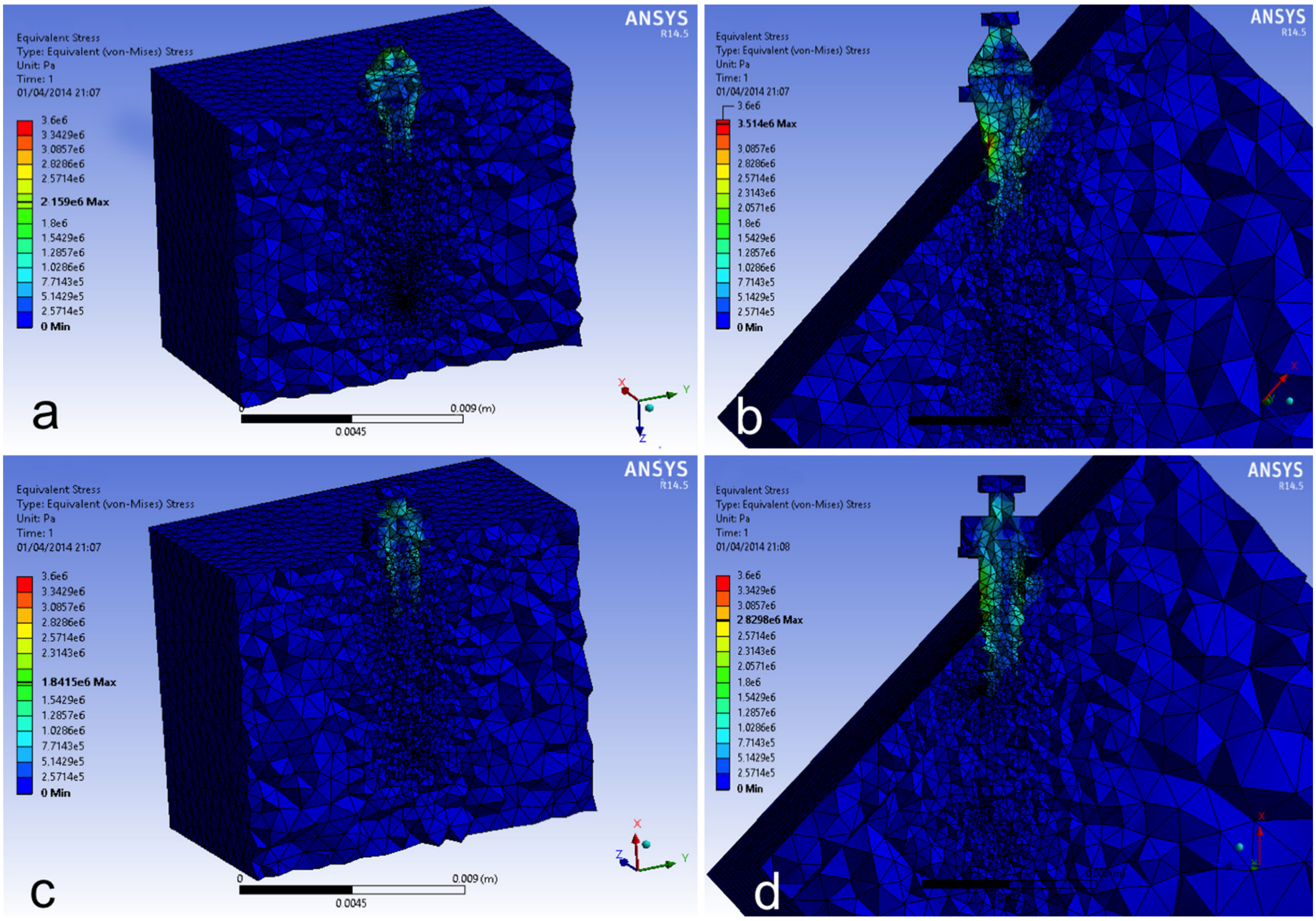

Figure 2. Maximum tension (MPa) induced on the OMI when inserted in the bone block, under a 200 gf traction force. a SIN 900, b SIN 450, c RMO 900, d RMO 450.

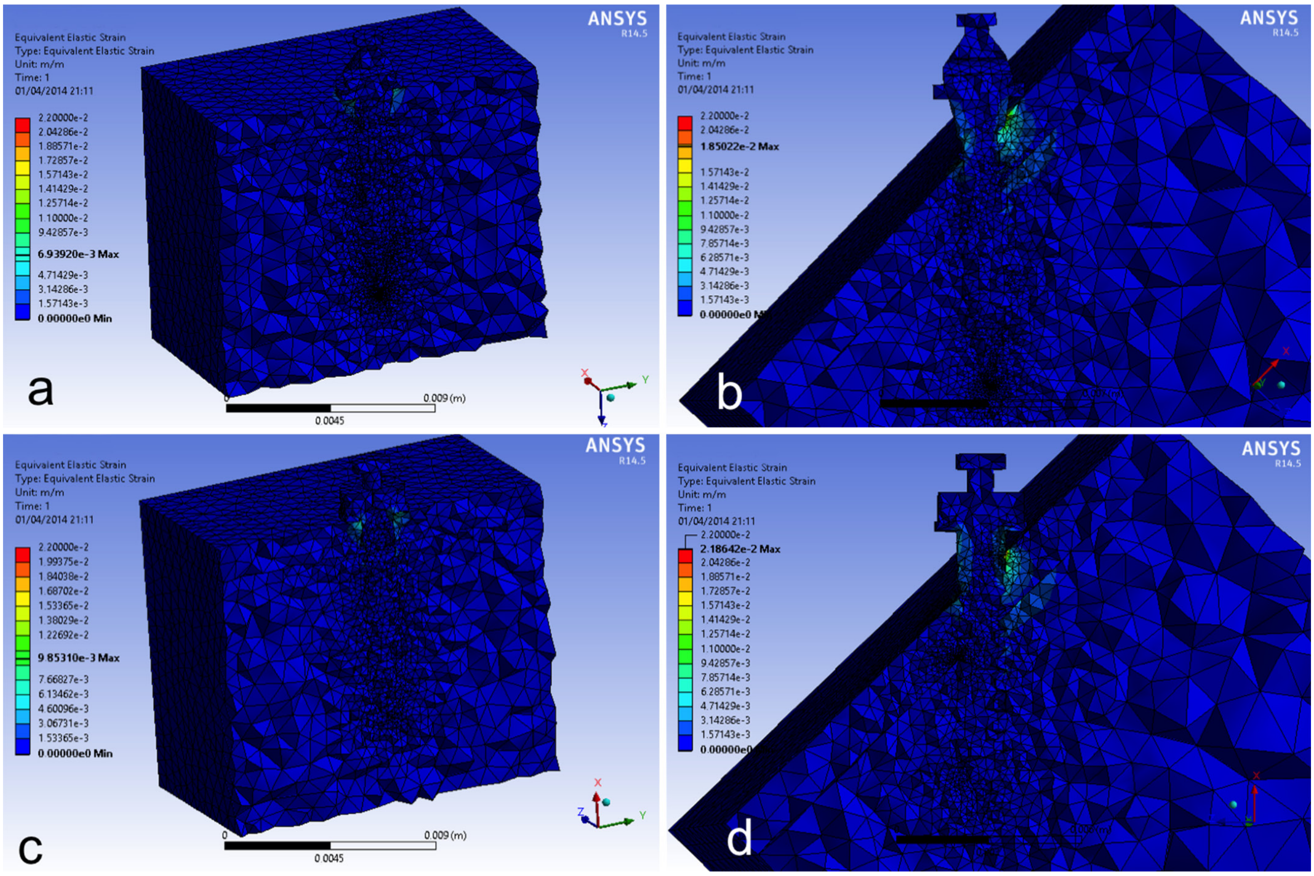

Figure 3. Maximum deformation (mm/mm) of the OMI when inserted in the bone block, under a 20 N.cm torsion force. a SIN 900, b SIN 450, c RMO 900, d RMO 450.

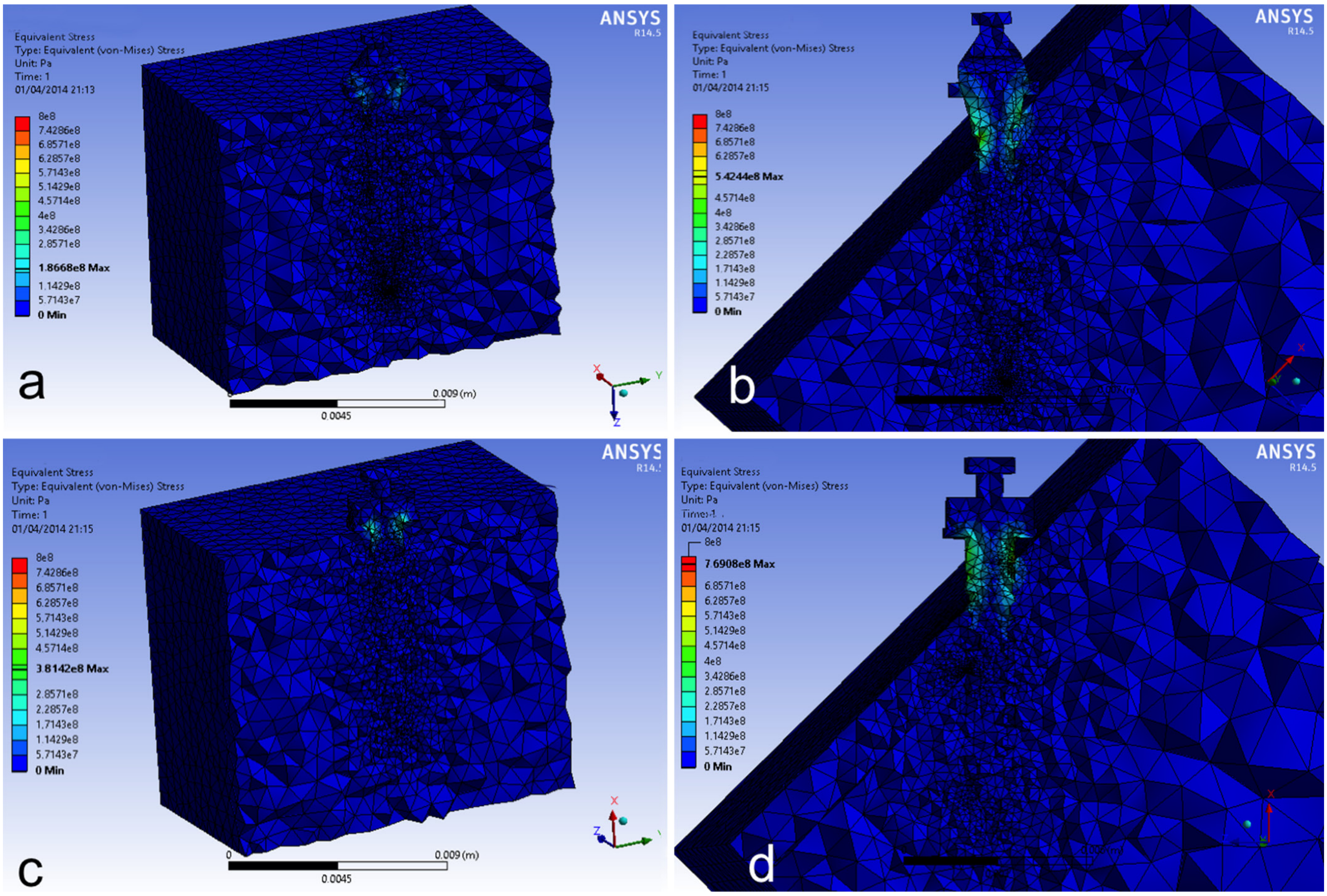

Figure 4. Maximum tension (MPa) induced on the OMI when inserted in the bone block, under a 20 N.cm torsion force. a SIN 900, b SIN 450, c RMO 900, d RMO 450.

When the brands tested were compared, RMO was shown to undergo greater deformation than SIN, even when subjected to tension forces. Therefore, it seems reasonable to assume that a smaller distance between threads, rather than their number, was responsible for the reduced resistance to elastic deformation of the screw. This speculation is based on the fact - according to the manufacturer's description – that the two implants tested were made of the same alloy, and were equally evaluated (e.g. size and conditions). These findings corroborate those previously published on the importance of the shape of the screw in the effectiveness of the technique[18].

With regard to stress distribution on the bone, tension in the cortical bone was far higher than that found in cancellous bone, showing the cortical bone to be responsible for the actual interlocking with the bone and the OMI. These data have been confirmed in literature, in which the quality of cortical bone has been shown to influence the primary stability of the mini-implant[1]. Therefore, the smaller the cortical bone thickness, the higher the tension induced into it, which may lead to resorption (and consequently failure) of the OMI[12-16]. Although there was an increased amount of deformation in the cortical bone in all the situations tested, there was a greater distribution of elastic deformation when cancellous bone tensile strength was considered. It was evident that When the OMI was subjected to orthodontic forces, there was evident elastic deformation in the bone around the entire screw, which was more concentrated in the cervical region. SIN showed higher strain values in cancellous bone, and the greater distribution in the body and apex of the screw could be a result of the type of thread tip used in this brand.

The results of this study showed that irrespective of force (tension or torsion), the OMI brand, or the inclination of the screw, there was always a greater strain on cortical bone and increased tension in the OMI. It could be speculated that the OMI should be stronger, especially in the transmucosal profile, the region that showed the highest level of stress concentration. Manufacturers would be prudent if they were to increase the resistance of the screw in this region, thus avoiding the occurrence of fractures during insertion. Other FEA-based studies have also observed an accumulation of stress in the same region of OMIs[15,17].

Insertion at an angle of 45 degrees increased the tension and bone strain for both brands of OMI tested, and in both loading protocols tested. This was evidently caused by the larger contact area between the cortical bone and the OMI. This is similar to findings previously shown in the literature with regard to increase in the number of exposed threads of the OMI (as caused by the angled OMI) resulting in increased bone tension around the OMI[25]. The authors also suggested that the traction force applied to an angled implant produced greater tension due to the increased lever arm which is formed, since a larger portion of the screw is outside of the bone[25].

A few studies in literature have suggested that OMIs should not be inserted perpendicular to the cortical bone[28,29]. This would avoid them coming into contact with the roots of teeth, in addition to increasing the insertion torque and primary stability of the OMI[28,29]. Nevertheless, a previous study has shown no significant difference in the success rate of OMIs inserted at an angle to the bone, when compared with OMIs inserted perpendicular to the bone[25]. The major problem associated with angulated insertion is that it does not allow complete insertion of the OMI threads into the bone, leading to a larger lever arm formed off-the-bone, negatively contributing to primary and secondary stability, as well as increasing the possibility of forming niches where food may accumulate[28]. Based on FEA analyses of human hemi-mandibles (*ex vivo*) it has been shown that the greater the insertion angle of OMIs, the greater the tension induced in the cortical bone, and the lower the tensile strength when subjected to forces parallel to the occlusal plane[28]. The distance between the cortical bone and the application of force varied from 2 to 4 mm, for insertion angles of 90 and 30 degrees, respectively. These findings corroborate those previously reported in literature showing that that the greater the distance between the bone and the application of force, the higher the OMI failure rate[29]. A recent study also showed that best results for OMI were achieved when an insertion angle of 90 degrees was used[30].

Divergences among the results of this and previous studies in the literature suggest the need of further *in vitro* and *in vivo* studies, to allow the formulation of a definitive rationale on the topic. However, from the mechanical point of view, a higher level of tension between the bone and the OMI would lead to a higher insertion torque (and therefore to better primary stability); the definition of a threshold, from which bone-tissue injuries would be induced, leading to implant failure should be considered[26,27].

When the FEA of the two brands of mini-implants were compared, it is fair to conclude that the larger number of threads and greater inclination of the threads resulted in less resistance to deformation and induced a higher level of tension in the mini-implant and cortical bone when subject to forces, especially when mini-implants were inserted at an angle of 45° to the cortical bone.

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CONFLICTS OF INTERESTS

The authors declare no conflicts of interest.

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