Revista Odontológica Mexicana
Órgano Oficial de la Facultad
de Odontología UNAM

Original research

Simulation of Mini-Screw Fracture when Inserted into the Mandibular Body, through Finite Element Analysis

Camila María Mendoza-Escobar[1], Víctor Manuel Araujo-Monsalvo[2], Ricardo Ondarza-Rovira[3], Salvador García-López[4]

[1] Egresada de la Maestría en Ortodoncia. Universidad Intercontinental, Ciudad de México, México. https://orcid.org/0009-0003-8093-4515

[2] Instituto Nacional de Rehabilitación, Ciudad de México, México. https://orcid.org/0000-0002-6969-0965

[3] Profesor de estadística. Universidad Intercontinental, Ciudad de México, México. Instituto Nacional de Investigaciones Nucleares, Estado de México, México. https://orcid.org/0000-0002-0135-8501

[4] Director de tesis, Universidad Intercontinental. Profesor honorario Departamento de Ortodoncia. Hospital General Dr. Manuel Gea González. Universidad Autónoma Metropolitana. Ciudad de México, México. https://orcid.org/0000-0003-1391-738X

Correspondence author:
Salvador García López
E-mail: salvador.garcia@universidad-uic.edu.mx

Received: 30 January 2024
Accepted: 24 September 2024

Cite as:
Mendoza-Escobar CM, Araujo-Monsalvo VM, Ondarza-Rovira R, García-López S. Simulación de fractura de minitornillos al insertarlos en el cuerpo mandibular, a través de un análisis de elementos finitos [Simulation of Mini-screw Fracture when Inserted into the Mandibular Body, through Finite Element Analysis]. *Rev Odontol Mex*. 2025; 29(3): 48-56. DOI: 10.22201/fo.1870199xp.2025.29.3.94256

Abstract

Introduction: Finite element analysis helps to predict the stress, compression, and deformation of a body under load. **Objective:** To assess the fracture risk of a stainless steel mini-screw when inserted into the mandibular body using finite element analysis. **Materials and Methods:** A 3D-printed mandibular model was obtained from a CBCT of a patient with class III malocclusion. A 2 x 12 mm long stainless steel mini-screw was then inserted into the mandibular body of the model to simulate the same direction in the tomography; 150 slices were obtained from the volumetric tomography, which was imported into the InVesalius 3.1.1 program. The tension levels exerted at 10 N/cm^2 and 14 N/cm^2 were measured at 2, 4, 6, 8, and 10 mm depths, both in the mini-screw and in the bone of the insertion area of the mandibular body. **Results:** When the mini-screw was inserted at 2 mm depth, at 10 N/cm^2 and 14 N/cm^2, the mini-screw presented greater tension at the tip and the circumferential bone edge of the insertion. When inserted at 10 mm, the mini-screw received greater tension in the body's upper part and neck. In the images, we observed that the self-tapping mini-screw received greater tension at the neck and upper part of the body than the self-drilling mini-screws, which received tension at the tip. A greater statistically significant difference was shown at 14 N/cm^2 compared to 10 N/cm^2. **Conclusions:** It is suggested to use self-drilling stainless steel mini screws of 2 mm width and 12 mm length in the first instance, applying a force of 10 to 12 N/cm^2 to achieve optimal stability, decreasing the force when fully inserting the mini screw to avoid fracture of its head.

Keywords: mini-screws, stress, fracture, finite element analysis, Orthodontics.

INTRODUCTION

Surgical mini-screws for orthodontic anchorage were introduced in 1997 by Kanomi[1], and other devices have emerged over time. Their application in orthodontics has brought great benefits to different types of treatment. The main advantages of mini-screws include: biocompatibility; minimal need for local anesthesia; easy insertion and removal; immediate acceptance of orthodontic forces; and minimal patient discomfort. However, over-insertion of the mini-screw can cause torsional stress on the neck of the device, loosening it and even causing it to become embedded in the gingival tissue[2,3]. In turn, the failure rate of mini-screws is higher in the mandible, where the cortical bone is thicker and denser, usually due in part to the greater stress concentrated around the mini-screw insertion region[4].

Currently, the use of mini-screws in orthodontic treatment is more common, as they facilitate the mechanics of treatment. Today, there is a wide variety of these devices, with diameters ranging from 1.0 mm to 2.3 mm and lengths from 4 mm to 21 mm, as well as different materials. The most common are titanium and stainless steel, with stainless steel being the best option due to its high rigidity and tip quality[5]. Each mini screw is selected according to the insertion site and the amount of available bone. There are a variety of mini screws in terms of dimensions and composition, some of which have proven to be better than others or may even fracture due to their diameter and length. In patients with Class III malocclusion and a need for mandibular arch retraction to achieve slight skeletal and dental correction, without extractions or orthognathic surgery, mini-screws are indicated in the mandibular body[6]. Inaba[7] and Park *et al.*[8] suggest placing the mini-screws at an angle to the bone surface that increases bone contact. Mini-screws should be placed laterally to the roots of the molars and as perpendicular as possible to the occlusal plane so that they function as an anchorage attachment. During the placement of mini-screws, increasing torsional stress is generated, which can bend, fracture, or cause small fissures in the peri-implant bone and affect its stability[9,10].

There is still a need to learn more about the risk of fracture of a mini-screw when inserting it into the mandibular body, as well as complications during placement and after activation of the orthodontic mechanics, all of which are related to patient stability and safety. A thorough understanding of the correct insertion technique, bone density, hard and soft tissue around the insertion site, regional anatomical structures, and post-placement indications is necessary to ensure the success and safety of mini-screws. To avoid stress on the mini-screw, it should be inserted slowly, with minimal pressure to ensure maximum bone contact. When inserting into dense cortical bone, the clinician should consider unscrewing one or two turns as the insertion is made to reduce stress on the mini-screw and bone. In dense bone, using long mini-screws can generate significant torsional forces and result in mechanical failure, usually in the area underlying the mini-screw head[11]. In addition, screws can be self-drilling or self-tapping. Self-drilling screws drill through bone tissue without the need for prior drilling, while self-tapping screws do require prior drilling.

Finite element studies of mini-screws analyze mechanical behavior and help optimize their design, as they include the evaluation of stresses, deformations, and load capacity under different conditions. This allows the performance of mini-screws to be simulated in real-life situations, such as torsion or axial loading, in order to prevent failure and improve efficiency. Therefore, the purpose of this study was to analyze the risk of fracture of 2mm-thick mini-screws placed in the mandibular body through finite element analysis.

MATERIALS AND METHODS

A simulation and virtual model study was performed using a CBCT scan of a male patient with Class III skeletal growth, which required a single CBCT scan to generate slices with a 0.625 mm separation in DICOM format, serving as the model for this study.

Finite Element Analysis (FEA) is a software used to perform analyses in different areas of engineering, among them, biomechanics. It allows complex structures such as human organs and tissues to be studied with high precision. FEA is based on dividing a model (continuous medium) into a series of non-intersecting subdomains called finite elements. The creation of the FEA required two stages: Pre-processing, which consisted of describing the geometry of a model with finite elements, as well as determining the material conditions (homogeneous, non-homogeneous, isotropic or anisotropic, and load conditions); and Post-processing, the stage in which the model was simulated with the various conditions applied to obtain the solution to the study. Therefore, the FEA allows doctors and engineers to study the biomechanics of human body structures using virtual models based on tomographic or magnetic resonance studies.

The study was performed using a Carestream CS 9500 3D System (Carestream Health, Inc., Rochester, NY) for volumetric tomography of a patient with Class III malocclusion. The following computer programs were used: Mimics 17.0™ (Materialise Inc., Leuven, Belgium); 3-Matic 10.0™ (Materialise Inc., Leuven, Belgium); Autodesk Fusion 360; ANSYS 14.5™ (Ansys Inc., Canonsburg, PA, USA). Two stainless steel mini-screws with a diameter of 2 mm and a length of 12 mm were chosen, one self-tapping and the other self-drilling.

Subsequently, DICOM format slices were obtained and exported to the Mimics 17.0 program. This was followed by segmentation, which consisted of using the Hounsfield units of the images to delimitate the different structures present to allow the assignment and demarcation of boundaries between each tooth and its ligament with reference to the surrounding bone and to establish the thickness of the cortical bone. In this step, noise elements and artifacts unrelated to the desired anatomical structures were removed.

Multiple coordinates were established in the three planes of space based on the slices obtained in the two-dimensional DICOM format. The relationship between these coordinates allowed the three-dimensional structure to be assembled. This operation was achieved using the 3-Matic 10.0 program. The Autodesk Fusion 360 program was used to design the screws, taking each mini-screw as a reference. Subsequently, the model and the mini-screw were exported to the ANSYS software, which was used to achieve volumetric reconstruction and break down several operations performed through ANSYS itself.

The jaw model was printed in 3D (Ender 3 pro creality, Shenzhen Creality 3D Technology Co., Ltd, China) using stereolithography from the tomography used for this study, which had previously determined the area and technique for inserting the mini-screws. Once the exact insertion position to be evaluated had been selected, each mini screw was inserted, guided step by step by the insertion technique recommended by Inaba[7] and Park *et al.*,[8] who suggest placing the mini screws at a 30° angle to the bone surface to increase bone contact. Figure 1 shows the stereolithographic model with the presence of the mandibular body or shelf where the mini-screw insertion area was selected. Using the Boolean subtraction operations of the Autocad program, it was possible to remove the amount of alveolar bone necessary for the intimate adaptation of the mini-screw inserted into the bone as a method of dental anchorage.

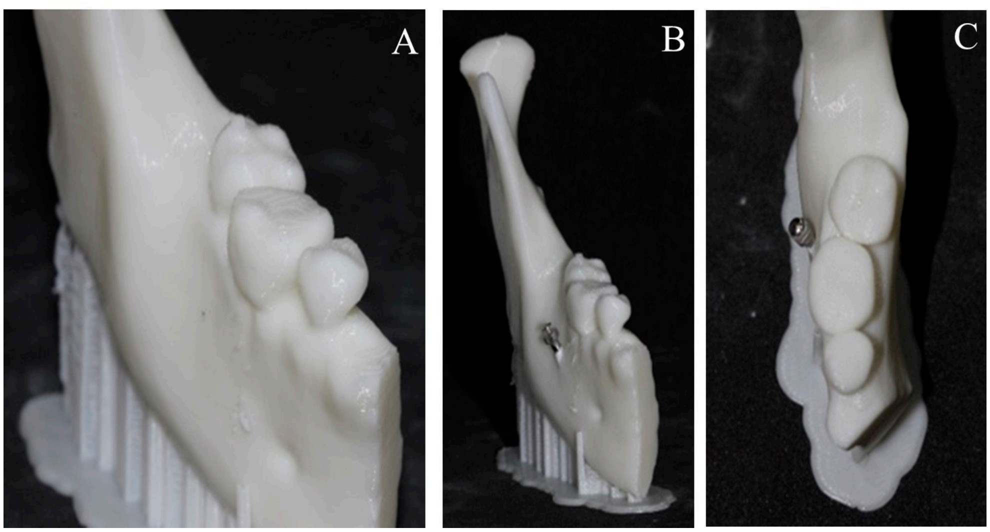

Figure 1. 3D-printed stereolithography showing the sequence of placement of a mini screw in the mandibular body. A. Initial model. B. Sagittal view with the inserted mini screw. C. Occlusal view.

Biomechanical properties were assigned to each of the structures that formed the model (Table 1), calculated by the program. The model was meshed by selecting elements consisting of ten-node tetrahedra. Boundary conditions were assigned to the distal area of the mandibular bone. The force vectors were simple load points starting from the mini-screw, and loads of 10 N/cm² and 14 N/cm² were used. Five scenarios were analyzed according to the stress generated at depths of 2, 4, 6, 8, and 10 mm, obtaining the Von Mises stress as well as the deformations in the mandibular bone. The tests were analyzed, the results were quantified, and they were entered into a Microsoft® Excel spreadsheet for statistical analysis. The data were expressed as mean and standard deviation (SD). The differences between the 10 N/cm^2 mini-screws compared to the 12 N/cm^2 mini-screws, both made of stainless steel, were determined using the Mann-Whitney U test with GraphPad Prism 10 software (GraphPad Software Inc., San Diego, CA, USA). The significance level was set at $p<0.05$.

Table 1. Elastic modulus and Poisson's ratio of each of the components used in the finite element model.

Material	Elastic Module (MPa)	Poisson ratio
Cortical bone	14.7	0.30
Tooth	20.7	0.30
Mini- screw	114	0.34
Periodontal ligament	6.89×10^{-5}	0.45
Titanium alloy	110	0.30
Stainless steel	230	0.30

Table footnote: The insertion of 2 mm x 12 mm self-tapping and self-drilling mini-screws was compared with the stress on the mandible of stainless steel mini-screws 2 to 10 mm deep at 10 and 14 N/cm^2 evaluated in megapascals (MPa).

RESULTS

Table 2 shows that the highest stress on the stainless steel self-tapping mini screw occurred at 2 mm of insertion. However, there was a statistically significant difference when comparing the stress of 10 N/cm^2 with that of 14 N/cm^2.

Table 2. Stress on the self-tapping stainless steel mini-screw.

Depth	Self-tapping stainless steel 10 N/cm^2	Self-tapping stainless steel 14 N/cm^2
2 mm	44.07	61.70
4 mm	28.14	39.39
6 mm	31.01	43.41
8 mm	27.46	38.44
10 mm	31.87	44.62
Average with standard deviation	32.51± 6.72	45.51± 9.42**

Table footnote: There was a statistically significant difference in the self-tapping mini screw at 14 N/cm^2 of force, compared to 10 N/cm^2 **$p<0.001$. The highest tension was at 2 mm (blue).

The tensile force of the self-drilling stainless steel mini screw at the time of insertion at 2, 4, 6, and 10 mm can be identified in Table 3, which shows that there was a statistically significantly greater difference in the self-drilling mini screw at 14 N/cm^2. Through Finite Element Analysis, the results are observed and expressed with images through Von Mises Stress, which expresses the physical magnitude proportional to the distortion energy, and was represented by colorimetry.

Table 3. Stress on the self-drilling stainless steel mini screw.

Depth	Self-drilling stainless steel 10 N/cm²	Self-drilling stainless steel 14 N/cm²
2 mm	51.12	71.57
4 mm	36.06	50.48
6 mm	39.17	54.84
8 mm	25.59	35.82
10 mm	26.94	37.71
Average with standard deviation	35.77 ±10.36	50.08 ±14.50**

Table footnote: There was a statistically significant difference in the self-drilling mini-screw at 14 N/cm² of force compared to 10 N/cm². The highest tension was at 2 mm (blue) ** $p<0.001$.

Figure 2 shows the results of the finite element analysis, where stainless steel mini-screws are inserted into the mandibular body, one at 10 N/cm² and the other at 14 N/cm². When inserted at 2 mm, the tips of both are shown in aqua, green, yellow, orange, and red, indicating higher stress in that area.

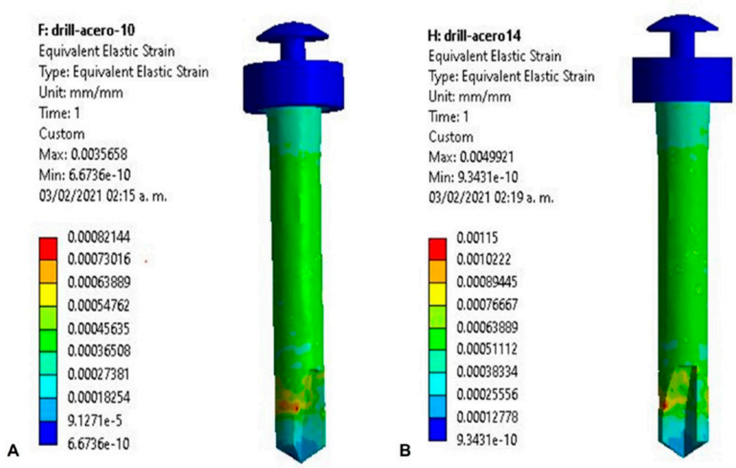

Figure 2. Colorimetric images of finite elements. A. 10 N/cm². B. 14 N/cm². The areas of highest stress at the tip of the mini-screw when inserted into the mandibular body are shown; the blue color indicates the highest stress at the tip.

On the other hand, Figure 3 shows the results of the finite element analysis comparing a self-tapping stainless steel mini screw and a self-drilling stainless steel mini screw when inserted 2 mm into the mandibular body. The self-tapping stainless steel mini screw shows green, yellow, and orange colors, indicating generalized pressure throughout the body of the mini screw. In the case of the self-drilling stainless steel mini screw, localized orange and red tones are shown, indicating greater stress at the tip. Finally, Figure 4 shows the results of the finite element analysis, which compares a self-tapping stainless steel mini screw with a self-drilling one when inserted 10 mm into the mandibular body. It can be observed that in both mini screws, the greatest stress occurs from the middle of the body towards the neck of the mini screws.

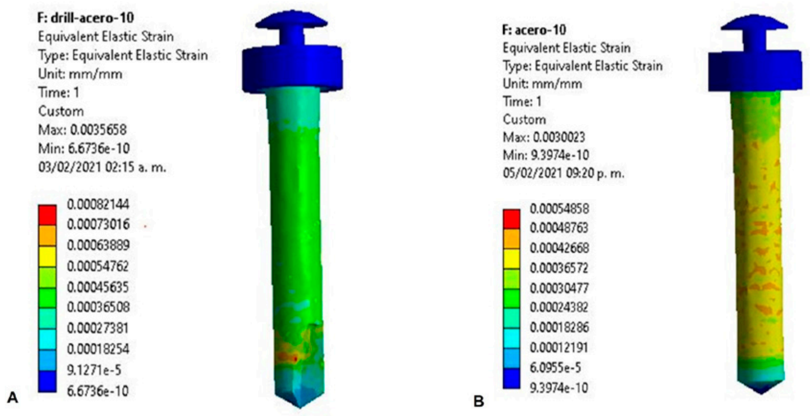

Figure 3. Comparison of colorimetric images of self-tapping and self-drilling mini-screws in the stress areas at 2 mm of insertion. A. Self-tapping screws receive greater overall stress in the body of the mini-screw (green). B. Self-drilling screws receive greater stress at the tip of the mini-screw (blue) and to a lesser degree in the body (yellow).

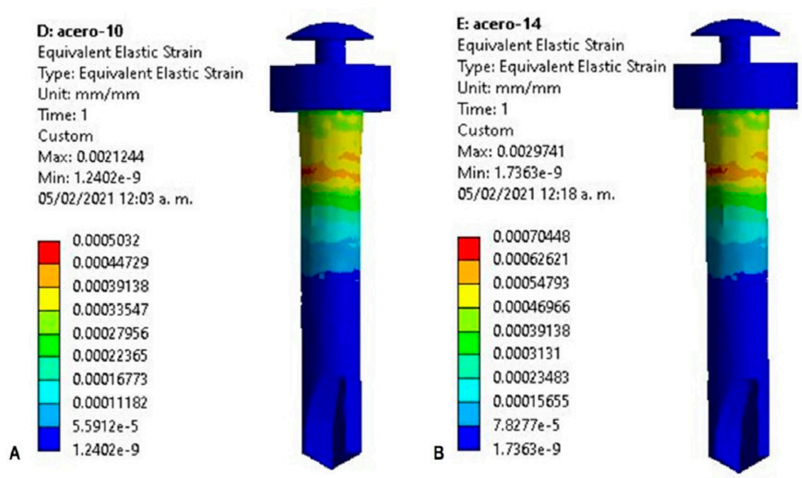

Figure 4. Colorimetric comparison of self-drilling mini-screws A. 10 N/cm^2. B. 14 N/cm^2. Both showed greater stress at the tip of the mini-screw (blue) inserted at 10 mm, with stress increasing in the neck area (green).

DISCUSSION

Mini-screws are a very useful alternative for performing different movements required in orthodontic practice, whether for anchorage or to exert a special mechanical force that is impossible to achieve with conventional orthodontics. On the other hand, placing a mini screw in the mandibular bone, which is considered one of the most compact bones in the skeleton, can lead to fracture of the mini screw during insertion, since the optimal torque recommended to avoid fracture of a mini screw is 5 to 10 N/cm^2 for a diameter of 1.5 mm[12].

Considering that the diameter of an orthodontic mini-screw to be inserted into the mandibular crest should be 2 mm[13], 2 mm diameter x 12 mm long stainless steel mini-screws were assessed to determine the risk of fracture through finite element analysis. This study showed that the stainless steel mini-screw, when inserted into the first 2 mm, as well as at 10 N/cm^2 and 14 N/cm^2, presented higher stress at the tip. This coincides with a study by Buschang *et al.*[14], which demonstrated the risk of fracture at the tip at the time of insertion. On the other hand, there was a greater statistically significant difference in stress at 14 N/cm^2 compared to 10 N/cm^2 in self-tapping mini-screws. Considering the above, it is recommended to use stainless steel mini-screws at 10 N/cm^2 of force for low-density bones, such as the maxilla, which ranges around 9.6 N/cm^2. However, the results found in the study apply to clinical needs, since the insertion torque for high-density bones, such as the mandible, ranges between 12.6 and 23.2 N/cm^2 [15-17], so an average force of 12 N/cm^2 can be used.

Studies conducted by Heidemann *et al.*[9], as well as those by Phillips and Rahn[10], demonstrated that during the placement of mini-screws, increasing torsional stress is generated, which can cause the mini-screw to bend, fracture, or produce small fractures in the peri-implant bone and affect the stability of the mini-screw. However, in the results obtained in this study, as the self-tapping mini-screw was inserted, an increase in tension was observed at different distances, with increases at 2, 6, and 10 mm and decreases at 4 and 8 mm. These variations demonstrated that the generated torsional stress does not occur gradually, contrary to what was proposed by Heidemann *et al.* [9]. When reviewing the results of the present study, it was demonstrated that when the self-tapping or self-drilling steel mini-screw is inserted at 10 mm, either at 10 N/cm^2 or 14 N/cm^2, the stress approached the neck area, which may cause stress and loss of stability of the mini screw. This coincides with the study conducted by Kravitz[2], who demonstrated that over-inserting the mini screw can cause stress on the neck of the device by generating torsional stress, which leads to the screw loosening and invaginating into the gingival tissue. During the evaluation of the tension of the stainless steel mini-screw, both self-drilling and self-tapping, the tension received in the insertion area in the mandibular body was shown. It should be noted that stainless steel mini-screws exert greater pressure around the insertion area[4]. This has an important clinical aspect, since fracture of the tip during insertion is part of a complication that involves surgical removal. In addition, it should be considered that when the mini screw is fully inserted, the stress on the body and neck of the screw increases, which can also cause fracture of the mini screw head.

CONCLUSIONS

Considering that the mandibular body has compact bone, it is suggested to use self-drilling stainless steel mini screws 2 mm wide and 12 mm long in the first instance for practical reasons, applying a force of 10 to 12 N/cm^2 to achieve optimal stability, reducing the force when fully inserting the mini screw to avoid fracturing its head.

Self-tapping mini screws 2 mm wide and 12 mm long exert greater tension on the body and head of the mini screw, so it is also recommended to use the same force suggested for the self-drilling mini screw.

Conflicts of interest: The authors declare no conflict of interest related to this work.

BIBLIOGRAPHIC REFERENCES

1. Kanomi R. Mini-implant for orthodontic anchorage. *J Clin Orthod*. 1997; 31(11): 763-767. PMID: 9511584
2. Kravitz ND, Kusnoto B. Risks and complications of orthodontic miniscrews. *Am J Orthod Dentofacial Orthop*. 2007; 131(4, Suppl): S43-S51. DOI: 10.1016/j.ajodo.2006.04.027
3. Cheng SJ, Tseng IY, Lee JJ, Kok SH. A prospective study of the risk factors associated with failure of mini-implants used for orthodontic anchorage. *Int J Oral Maxillofac Implants*. 2004; 19(1): 100-106. PMID: 14982362
4. Lin TS, Tsai FD, Chen CY, Lin LW. Factorial analysis of variables affecting bone stress adjacent to the orthodontic anchorage mini-implant with finite element analysis. *Am J Orthod Dentofacial Orthop*. 2013; 143(2): 182-189. DOI: 10.1016/j.ajodo.2012.09.012
5. Chen CH, Chang CS, Hsieh CH, Tseng YC, Shen YS, Huang IY, *et al*. The use of microimplants in orthodontic anchorage. *J Oral Maxillofac Surg*. 2006; 64(8): 1209-1213. DOI: 10.1016/j.joms.2006.04.016
6. Lin JJJ. The most effective and simplest ways of treating severe class III, without extraction or surgery. *IJOI. Int J Orthod Implantol*. 2014; 33: 4–18. Disponible en: https://iaoi.pro/asset/files/ijoi_33_pdf_article/004_018_new.pdf
7. Inaba M. Evaluation of primary stability of inclined orthodontic mini-implants. *J Oral Sci*. 2009; 51(3): 347-353. DOI: 10.2334/josnusd.51.347
8. Park HS, Jeong SH, Kwon OW. Factors affecting the clinical success of screw implants used as orthodontic anchorage. *Am J Orthod Dentofacial Orthop*. 2006; 130(1): 18-25. DOI: 10.1016/j.ajodo.2004.11.032
9. Heidemann W, Gerlach KL, Gröbel KH, Köllner HG. Influence of different pilot hole sizes on torque measurements and pullout analysis of osteosynthesis screws. *J Craniomaxillofac Surg*. 1998; 26(1): 50-55. DOI: 10.1016/s1010-5182(98)80035-8
10. Phillips JH, Rahn BA. Comparison of compression and torque measurements of self-tapping and pretapped screws. *Plast Reconstr Surg*. 1989; 83(3): 447-456. DOI: 10.1097/00006534-198903000-00007
11. Campbell JH. Titanium screw failure: a case report. *J Oral Maxillofac Surg*. 1993; 51(5): 603-605. DOI: 10.1016/s0278-2391(10)80526-9
12. Motoyoshi M, Hirabayashi M, Uemura M, Shimizu N. Recommended placement torque when tightening an orthodontic mini-implant. *Clin Oral Implants Res*. 2006; 17(1): 109-114. DOI: 10.1111/j.1600-0501.2005.01211.x
13. Tseng LLY, Chang CH, Roberts WE. Diagnosis and conservative treatment of skeletal class III malocclusion with anterior crossbite and asymmetric maxillary crowding. *Am J Orthod Dentofacial Orthop*. 2016; 149(4): 555-566. DOI: 10.1016/j.ajodo.2015.04.042
14. Buschang PH, Carrillo R, Ozenbaugh B, Rossouw PE. 2008 survey of AAO members on miniscrew usage. *J Clin Orthod*. 2008; 42(9): 513-518. PMID: 18974458
15. Cattaneo PM, Dalstra M, Melsen B. Moment-to-force ratio, center of rotation, and force level: a finite element study predicting their interdependency for simulated orthodontic loading regimens. *Am J Orthod Dentofacial Orthop*. 2008; 133(5): 681-689. DOI: 10.1016/j.ajodo.2006.05.038
16. Farah JW, Craig RG, Sikarskie DL. Photoelastic and finite element stress analysis of a restored axisymmetric first molar. *J Biomech*. 1973; 6(5): 511-520. DOI: 10.1016/0021-9290(73)90009-2
17. Brown RN, Sexton BE, Chu TMG, Katona TR, Stewart KT, Kyung HM *et al*. Comparison of stainless steel and titanium alloy orthodontic miniscrew implants: a mechanical and histologic analysis. *Am J Orthod Dentofacial Orthop*. 2014; 145(4): 496-504. DOI: 10.1016/j.ajodo.2013.12.022