

Article

Maxillary Skeletal Expansion with Monocortical and Bicortical Miniscrew Anchorage: A 3D Finite Element Study

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Abstract: The aim of the present study is to use a 3D finite element analysis to investigate and compare the transverse displacement and stress distribution between stainless steel miniscrews and titanium alloy miniscrews used for monocortical and bicortical anchorage during miniscrew-assisted rapid maxillary expansions. Skull models were constructed to depict expansion after and before midpalatal suture opening at varying miniscrew insertion depths in four clinical scenarios: monocortical, monocortical deepening, bicortical, and bicortical deepening. Finite element analyses of miniscrew properties, including transverse displacement and von Mises stress distribution, were performed for each clinical scenario. Peri-implant stress was lesser in both bicortical anchorage models compared to both monocortical models. Transverse displacement in the coronal and axial planes was also greater and more parallel in both bicortical models compared to both monocortical models. Transverse displacement and peri-implant stress did not significantly differ between monocortical and monocortical deepening models or between bicortical and bicortical deepening models. From a biomechanical perspective, the bicortical deepening miniscrew anchorage is preferable to monocortical and monocortical deepening anchorage, because bicortical anchorage induces less stress on the peri-implant bone. Consequently, bicortical deepening anchorage should be considered the preferred option in challenging clinical scenarios in which strong anchorage is required for maxillary skeletal expansion.

Keywords: bicortical anchorage; finite element analysis; monocortical anchorage; maxillary skeletal expansion; stainless steel miniscrew; titanium alloy miniscrew

1. Introduction

The prevalence of maxillary transverse deficiency is highly present in all age populations [1–3]. During primary and mixed dentition and early permanent dentition, rapid maxillary expansion (RME) is a simple procedure with high success rates. It is a stable procedure in the short and long term, regardless of the type of expander used [4,5]. Tooth-

supported RME is not used in skeletally mature patients with a fused palatal suture because of the side effects and complications involved, such as buccal crown tipping, alveolar bone bending, alveolar bone dehiscence, gingival recession, reduction in buccal bone thickness, and marginal bone loss [6,7]. In adults, the surgically assisted RME (SARME) is the main alternative for the treatment of maxillary transverse discrepancies [8,9].

Recently, nonsurgical RME using miniscrew anchorage has been identified as another possibility for the treatment of adult maxillary transverse deficiency [10,11]. This novel bone-borne maxillary expansion technique is known as miniscrew-assisted RME (MARME). Whereas conventional rapid maxillary expanders have questionable effects on the maxillary basal bone, MARME avoids this issue by using a rigid element to deliver an expansion force directly to the basal bone.

Compared to a bicortical miniscrew placement, a monocortical placement provides lower force resistance and stability, but higher cortical bone stress [12]. Clinically relevant dimensions of bone available for a palatal mini-implant anchorage include both cortical layers, i.e., the outer cortical layer of the oral hard palate and the nasal floor. Clinicians should consider bicortical anchorage when a high orthodontic load is expected. In addition to the bicortical anchorage consideration, the effects of the number, position, and length of miniscrews in the palatal bone for expansion treatment should be deliberated [13]. The effects of the monocortical and the bicortical miniscrew placement were investigated by Sermboonsang et al. in 2020 [14]. The results showed that the bicortical placement technique increases the stability of miniscrews. The bicortical insertion of the miniscrews decrease the risk of screw deformation. Moreover, the bicortical placement techniques should be performed, producing the good clinical outcome.

The midpalatal suture opening was influenced by treatment protocol. The reference indicates that the bicortical miniscrew placement could promote a more favorable parallel displacement of the maxillopalatal structure, which obviously represents different expansions in shape between bicortical and monocortical placements [14]. The evidence has confirmed the contribution of the bicortical miniscrew placement, if the parallel displacement of a maxillary suture is needed. The midpalatal suture opening was discovered to have two types of parallels, and wedge widening using different treatment conditions of the rapid maxillary expansion, resulting in the midpalatal suture opening being displayed in the study by Habersack et al. [15]. The material effect of the miniscrews also influences the midpalatal suture opening. The previous literature indicates that miniscrews in stainless steel have significantly higher values of fracture torque and flexural strength than in titanium alloy [16]. Hence, the toughness in the stainless steel miniscrews is expressively greater than in titanium alloy miniscrews. The stiffer miniscrews seem to provide a better outcome in regard to the midpalatal suture opening when using rapid maxillary expansion.

The finite element analysis (FEA) is a useful resource for performing a virtual biomechanical assessment of possible clinical outcomes. The three-dimensional (3D) FEA is used to simulate maxillary expansion and to evaluate the stress distribution and displacement of the maxillofacial complex [17–19]. However, FEA is rarely used to evaluate the role of bone-borne expanders requiring heavy anchorage in maxillary orthopedic expansion. Most of the mini-implants in current use are composed of titanium alloy (Ti-6Al-4V; ASTM F136); in traumatology, however, mini-implants composed of implant-grade stainless steel (ASTM 316L) are still used [20]. Consequently, the main purpose of the present study is to use 3D FEA to analyze and compare how monocortical and bicortical anchorage using stainless steel and titanium alloy miniscrews affects stress distribution and displacement during MARME.

2. Materials and Methods

The FEA solid model of a 3D skull was reconstructed from CBCT images of a male subject by using medical image reconstruction of the Avizo software (version 7.0; Visualization Sciences Group, Burlington, MA, USA). The boundary of cortical and cancellous

bone of the maxillopalatal arch was determined and selected from each CT section to build the 3D model for preparing a standard maxillopalatal model in this study. The 3D maxillopalatal model was a homogenized FEA model, which was assumed to have linear elastic, homogenous, and isotropic properties. The total number of elements and nodes in the FEA model was 153,954 and 237,842, respectively. The ranges of element size, with edges from 0.3 to 1.5 mm, were considered to mesh in the FEA model according to investigating importance of the maxillopalatine and the expansion devices.

Figure 1 is the solid model of the skull, which included the cortical shell, cancellous bone, and midpalatal suture. The virtual skull model excluded other sutures, such as pterygomaxillary, zygomaticomaxillary, and zygomaticotemporal sutures. In this stage, teeth displacement was disregarded when analyzing effects of MARME on midpalatal suture opening. Hence, maxillary dentition, which was assumed to have the same material properties as cortical bone, was not isolated when tooth models were reconstructed from CBCT images. The skull model had a cortical bone thickness and a midpalatal suture width of 2 mm and 1.5 mm, respectively.

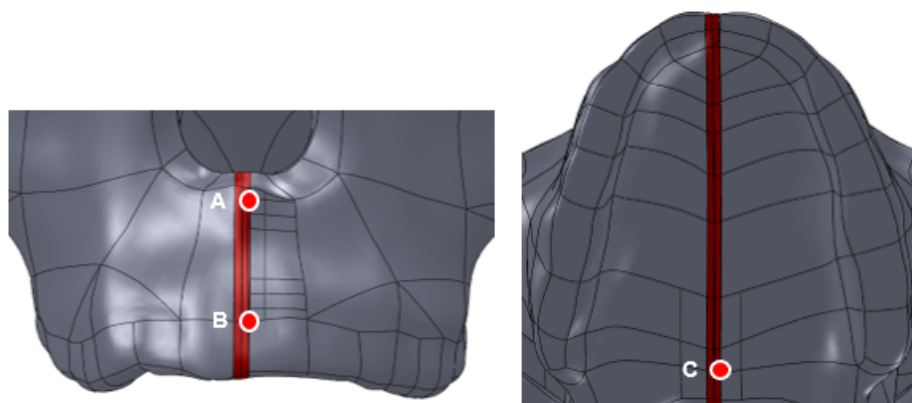


Figure 1. The 3D skull FEA model with midpalatal suture. Point A, the most anterior point of nasal floor; point B, the most anterior inferior point on the maxillary alveolar process; point C, the most posterior point at the posterior margin of the hard palate.

The mini-implants (2.0 mm in diameter; 6, 8, 10, 12, 14, and 16 mm in length; A1 Series; Bio-Ray, Biotech Instrument Co., Taipei, Taiwan) and a specified bone-borne maxillary expander design (skeletal maxillary expander, SME; Bio-Ray, Biotech Instrument Co.) were constructed with CAD software (SolidWorks 2010, Solidworks Corp, Waltham, MA, USA) according to design specifications provided by the manufacturer. The CAD model of the SME included the expansion screw (diameter, 2 mm), transverse rods (diameter, 1.5 mm), expansion sliders, support wires (diameter, 1 mm) (Figure 2), and fixed bands. Six expansion miniscrews of varying lengths were selected to investigate the effect of transverse displacement of palatal bone under application of expansion force caused by increased mechanical resistance in the midpalatal suture. Varying combinations of two different lengths of miniscrews inserted into the palatal bone were modeled to study the mechanical effect in clinical treatment.

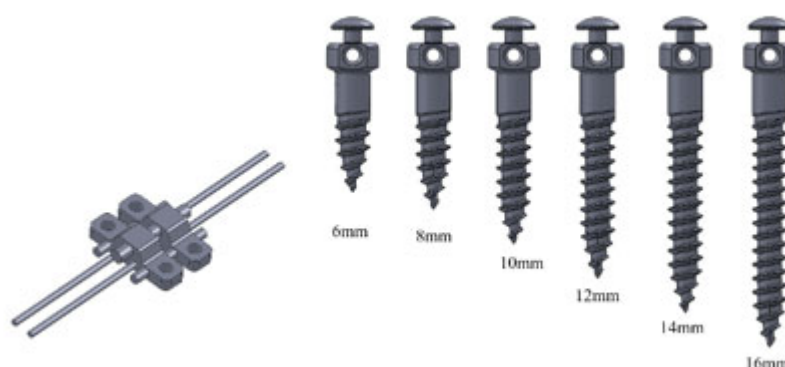


Figure 2. SME expansion screw and miniscrews of the FEA model.

Figure 3 shows the FEA models with different miniscrews in lengths in this study. The miniscrews were placed on both sides in the palatal aspect. Model A of monocortical anchorage was modeled using monocortical miniscrews 6 mm in length located at the posterior and anterior positions; model B of monocortical deepening anchorage was modeled using monocortical miniscrews 8 mm in length located at the posterior and anterior position; model C of bicortical anchorage was modeled using bicortical miniscrews 10 mm in length located at the posterior position and 12 mm in length located at the anterior position; model D of bicortical deepening anchorage was modeled using bicortical miniscrews 14 mm in length located at the posterior position and 16 mm in length located at the anterior position. Two different miniscrew compositions, stainless steel and titanium alloy, were also investigated. The mechanical properties of the cortical bone, cancellous bone, suture, titanium alloy, and stainless steel in the model were defined according to previous studies, as shown in Table 1 [21–24].

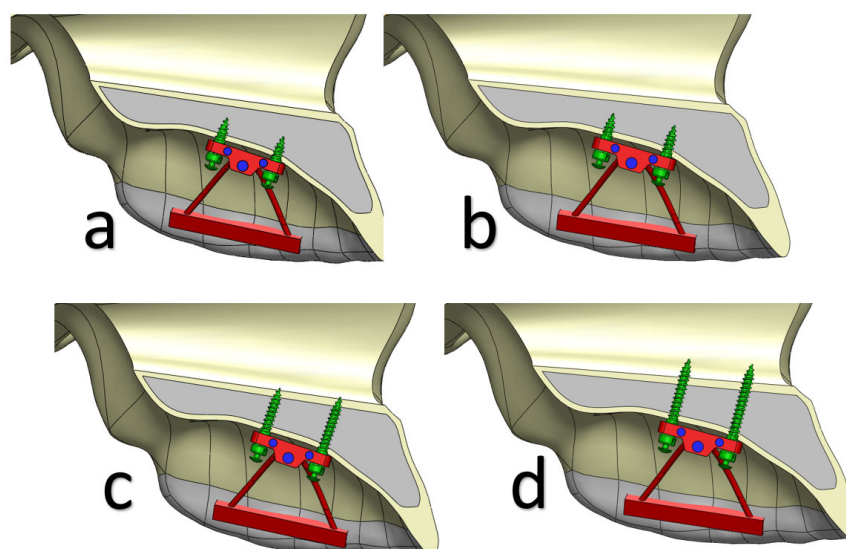


Figure 3. Four insertion types of miniscrews with varying lengths. (a) Monocortical: posterior and anterior miniscrews 6 mm in length both sides; (b) monocortical deepening: posterior and anterior miniscrews 8 mm in length both sides; (c) bicortical: posterior and anterior miniscrews 10 mm and 12 mm in length both sides; (d) bicortical deepening: posterior and anterior miniscrews 14 mm and 16 mm in length both sides.

Table 1. Mechanical properties of the materials assigned in FEA.

	Young's Modulus (MPa)	Poisson's Ratio	Reference
Cortical bone	13,700	0.30	17
Cancellous bone	1370	0.30	18
Suture	10	0.49	20
Stainless steel	210,000	0.30	19
Titanium alloy	113,000	0.33	20

The boundary condition of the FEA model was fixed at the whole surface of cutting plane in the skull to restrict degree of freedom in the translation and rotation. In Figure 4, blue triangles were only used to indicate where the boundary region of the FEA model was applied; hence, the boundary condition was applied at the cutting plane of the FEA model. The loading in the FEA model was applied in 20 steps of expansion displacement to reflect the expansion treatment protocol. The segmental load steps could provide slight changes of the suture expansion on the maxillopalatal bone by the use of rapid maxillary expansion.

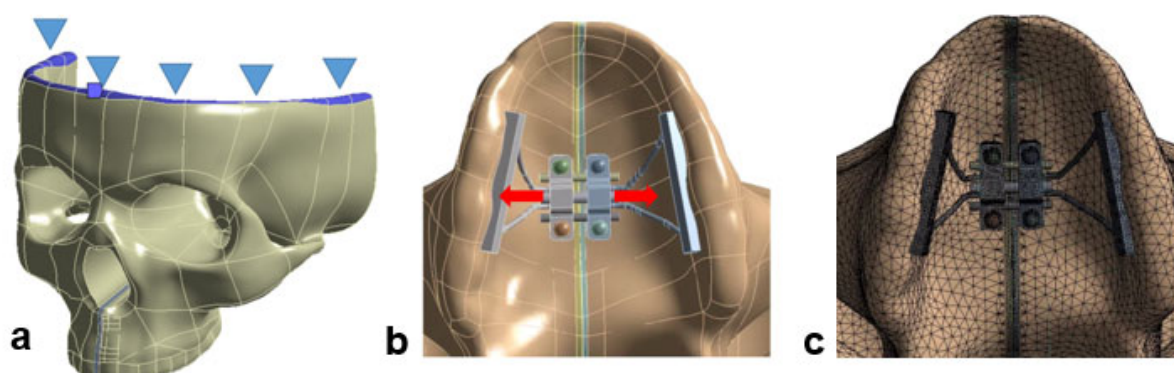


Figure 4. (a) The cross-section of cranium was constrained as boundary condition (triangle arrow-heads); (b) palatal bone-borne force application was indicated with red arrows; 0.25mm displacement of each driving step was applied at the expansion screw of the SME to simulate expansion treatment; (c) the mesh model of the skull with midpalatal suture treated by SME expansion system.

3. Results

The von Mises stress is only dependent on the difference between the three principal stresses. The definition of the von Mises stress (σ_v) is shown below, where σ_1 , σ_2 , and σ_3 are the principal stresses. The definition in all of the stress difference percentages between two models was calculated as $[(A - B)/(A + B)/2] \times 100\%$.

$$\sigma_v = \sqrt{\frac{(\sigma_1 - \sigma_2)^2 + (\sigma_2 - \sigma_3)^2 + (\sigma_3 - \sigma_1)^2}{2}}$$

The mean value of von Mises stress was represented to stand for an average of the von Mises stresses of four miniscrews (or peri-implant bone) in the FEA model. The mean von Mises stress was considered to evaluate the effects of the stress distribution of the miniscrews (or peri-implant bone) under different inserted depths of the miniscrews when each expansion step was performed.

3.1. Von Mises Stress in Peri-Implant Bone

The von Mises stress in the peri-implant bone was measured in the skull model with the midpalatal suture (Table 2). The difference between the bicortical model (2213.15 MPa)

and the monocortical model (2787.58 MPa) was 22.97%, the difference between the bicortical deepening model (2105.60 MPa) and the monocortical model (2787.58 MPa) was 27.87%, the difference between the bicortical model (2213.15 MPa) and the monocortical deepening model (2646.48 MPa) was 17.83%, and that between the bicortical deepening model (2105.60 MPa) and the monocortical deepening model (2646.48 MPa) was 22.76%. The difference between the monocortical deepening model (2646.48 MPa) and the monocortical model (2787.58 MPa) was 5.19%, and that between the bicortical deepening model (2105.60 MPa) and the bicortical model (2213.15 MPa) was 4.98%.

Table 2. Mean von Mises stress (MPa) of the peri-implant bones for the skull. Model with midpalatal suture for all 4 anchorage models.

Miniscrew Implant	Monocortical	Monocortical Deeping	Bicortical	Bicortical Deeping
Stainless steel	2787.58	2646.48	2213.15	2105.60
Titanium alloy	2398.63	2205.90	1967.08	1756.59

The mean von Mises stress at the implant–bone interface of titanium alloy miniscrews was also calculated for each anchorage model. The difference between the bicortical model (1967.08 MPa) and the monocortical model (2398.63 MPa) was 19.77%, the difference between the bicortical deepening model (1756.59 MPa) and the monocortical model (2398.63 MPa) was 30.90%, the difference between the bicortical model (1967.08 MPa) and the monocortical deepening model (2205.90 MPa) was 11.45%, and that between the bicortical deepening model (1756.59 MPa) and the monocortical deepening model (2205.90 MPa) was 22.68%. The difference between the monocortical deepening model (2205.90 MPa) and the monocortical model (2398.63 MPa) was 8.37%, and that between the bicortical deepening model (1756.59 MPa) and the bicortical model (1967.08 MPa) was 7.31%.

The von Mises stress was obviously lower in the two bicortical anchorage models compared to the two monocortical anchorage models. The implant–bone interface of stainless steel miniscrews had greater stress compared to that of titanium alloy miniscrews. In all models, the von Mises stress was localized at the implant–bone interface surrounding the initial cortical bone layers (Figure 5a–h).

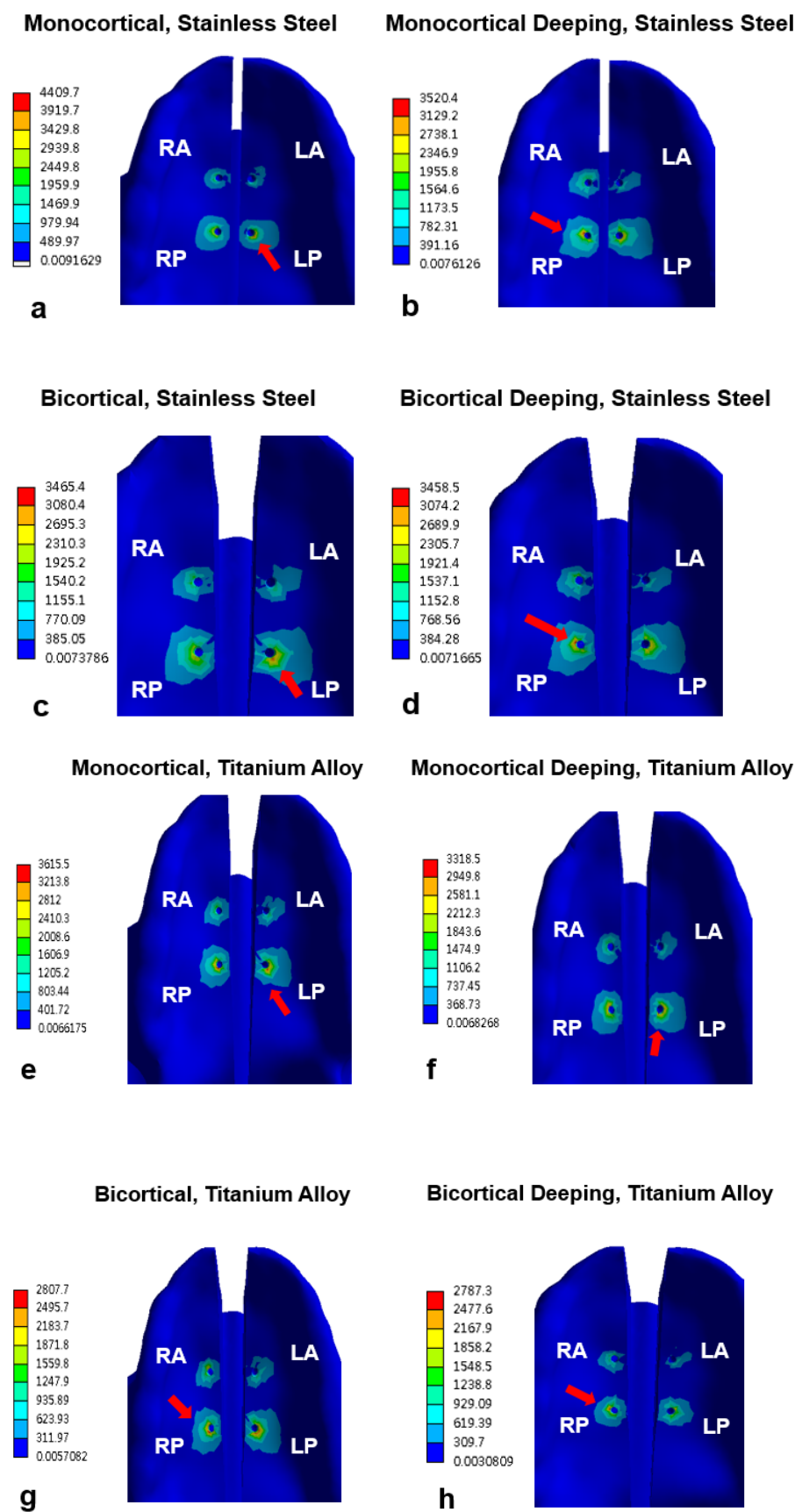


Figure 5. (a–d) Mean von Mises stress (MPa) of the peri-implant sites of stainless steel for the skull model with midpalatal suture for all 4 anchorage models. RA, right anterior; RP, right posterior; LA,

left anterior; LP, left posterior. Red arrow represents the location of the peak stress in the FE model. (e–h) Mean von Mises stress (MPa) of the peri-implant sites of titanium alloy for the skull model with midpalatal suture for all 4 anchorage models. RA, right anterior; RP, right posterior; LA, left anterior; LP, left posterior. Red arrow represents the location of the peak stress in the FE model.

3.2. Von Mises Stress in Miniscrews

The von Mises stress of the miniscrew implants was measured in the skull model with the midpalatal suture (Table 3). The difference between the bicortical model (23,506.00 MPa) and the monocortical model (36,768.88 MPa) was 44.01%, the difference between the bicortical deepening model (20,774.25 MPa) and the monocortical model (36,768.88 MPa) was 55.59%, the difference between the bicortical model (23,506.00 MPa) and the monocortical deepening model (34,987.75 MPa) was 39.26%, and that between the bicortical deepening model (20,774.25 MPa) and the monocortical deepening model (34,987.75 MPa) was 50.98%. The difference between the monocortical deepening model (34,987.75 MPa) and the monocortical model (36,768.88 MPa) was 4.96%, and that between the bicortical deepening model (20,774.25 MPa) and the bicortical model (23,506.00 MPa) was 12.34%.

Table 3. Mean von Mises stress (MPa) of the miniscrew implants for the skull. model with midpalatal suture for all 4 anchorage models.

Miniscrew Implant	Monocortical	Monocortical Deeping	Bicortical	Bicortical Deeping
Stainless steel	36,768.88	34,987.75	23,506.00	20,774.25
Titanium alloy	30,898.75	19,196.25	15,361.25	13,165.75

The mean von Mises stress in titanium alloy miniscrews was also calculated for each anchorage model. The difference between the bicortical model (15,361.25 MPa) and the monocortical model (30,898.75 MPa) was 67.17%, the difference between the bicortical deepening model (13,165.75 MPa) and the monocortical model (30,898.75 MPa) was 80.49%, the difference between the bicortical model (15,361.25 MPa) and the monocortical deepening model (19,196.25 MPa) was 22.19%, and that between the bicortical deepening model (13,165.75 MPa) and the monocortical deepening model (19,196.25 MPa) was 37.27%. The difference between the monocortical deepening model (19,196.25 MPa) and the monocortical model (30,898.75 MPa) was 6.72%, and that between the bicortical deepening model (13,165.75 MPa) and the bicortical model (15,361.25 MPa) was 15.01%.

The von Mises stress was obviously lower in the two bicortical anchorage models compared to the two monocortical anchorage models. The titanium alloy mini-implants had lesser stress compared to the stainless steel mini-implants. In all four models, the von Mises stress in the implant was localized at the neck around the initial cortical bone layers (Figure 6a–h).

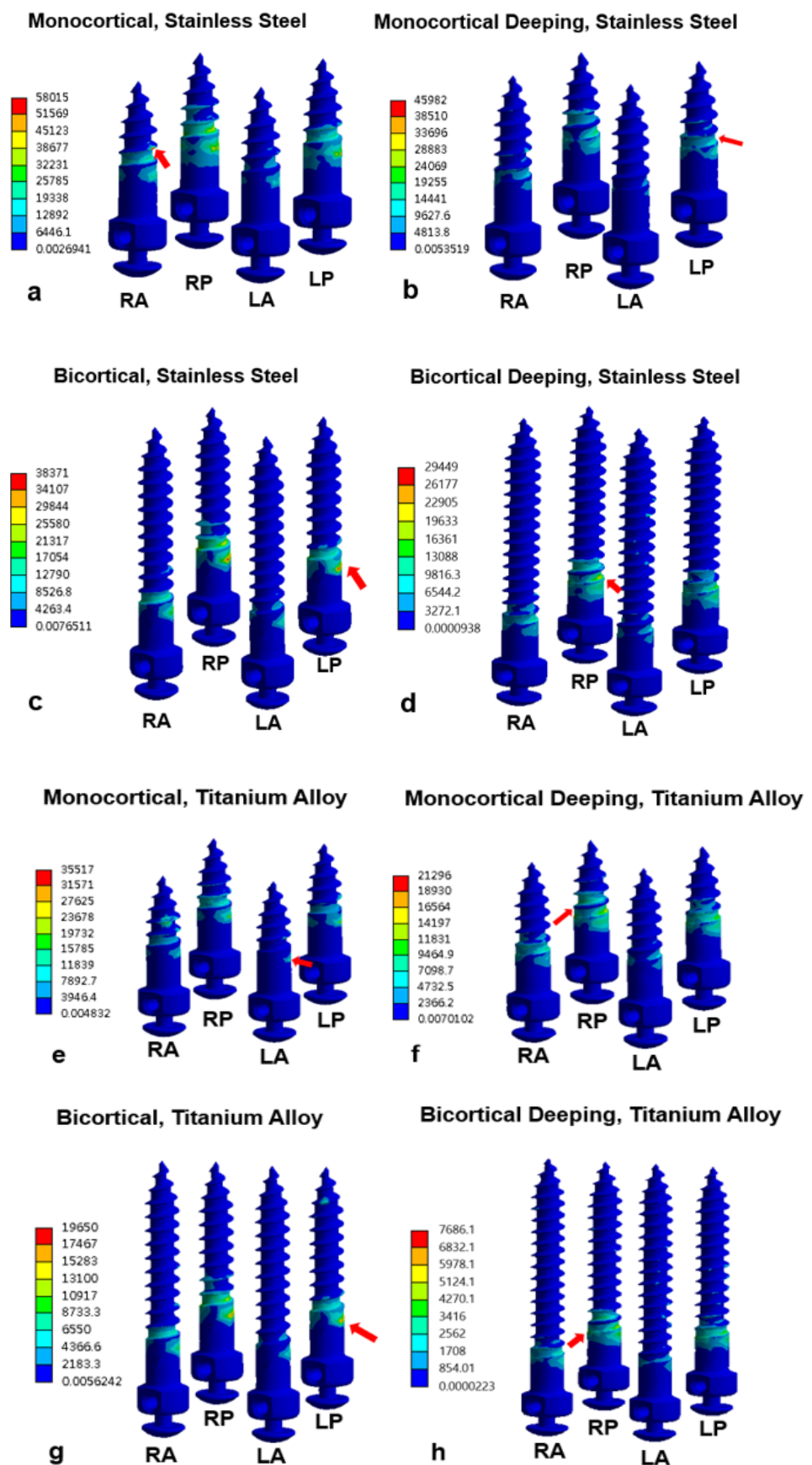


Figure 6. (a–d). Mean von Mises stress (MPa) of the miniscrew implants of stainless steel for the skull model with midpalatal suture for all 4 anchorage models. RA, right anterior; RP, right posterior; LA, left anterior; LP, left posterior. Red arrow represents the location of the peak stress in the

FE model. (e–h) Mean von Mises stress (MPa) of the miniscrew implants of titanium alloy for the skull model with midpalatal suture for all 4 anchorage models. RA, right anterior; RP, right posterior; LA, left anterior; LP, left posterior. Red arrow represents the location of the peak stress in the FE model.

3.3. Bending in Miniscrews

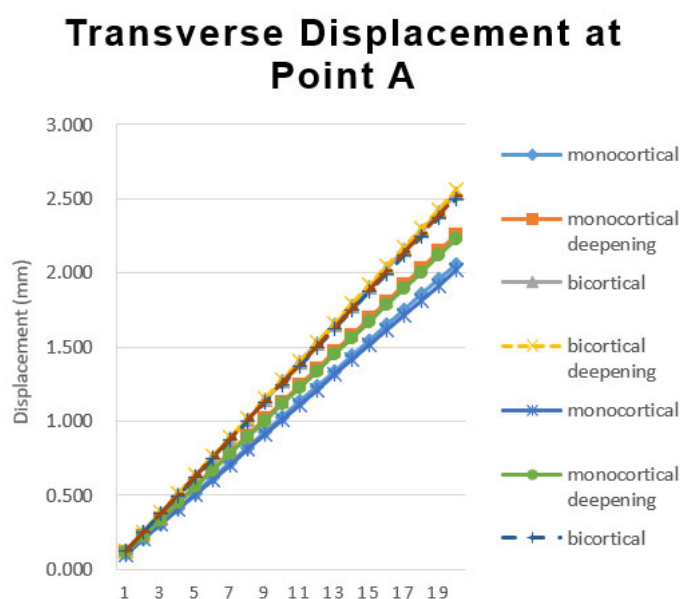
Bending was clearly evident in the miniscrews. Table 4 shows the bending measured in the four miniscrews composed of stainless steel and titanium alloy. The mean degree of bending in stainless steel miniscrews was 3.89° for the monocortical model, 3.19° for the monocortical deepening model, 2.41° for the bicortical model, and 2.33° for the bicortical deepening model. The mean degree of bending in titanium alloy miniscrews was 4.15° for the monocortical model, 3.64° for the monocortical deepening model, 3.23° for the bicortical model, and 3.01° for the bicortical deepening model. The titanium alloy miniscrew implants had a greater mean degree of bending compared to the stainless steel miniscrew implants for the same model.

Table 4. Mean degree of bending of the miniscrew implants for the skull. Model with midpalatal suture for all 4 anchorage models.

	Monocortical	Monocortical Deeping	Bicortical	Bicortical Deeping
Stainless steel	3.89°	3.19°	2.41°	2.33°
Titanium alloy	4.15°	3.64°	3.23°	3.01°

3.4. Midpalatal Suture Opening

Transverse displacement was measured on the right side of the skull model with the midpalatal suture and was determined for each step, which was 20 total steps. These 20 steps were equivalent to 20 turns of 0.25 mm each, for a total expansion of 5 mm (2.5 mm on each side). Figure 1 lists the right-side transverse displacement measurements at points A, B, and C; Figure 7 plots the displacement.



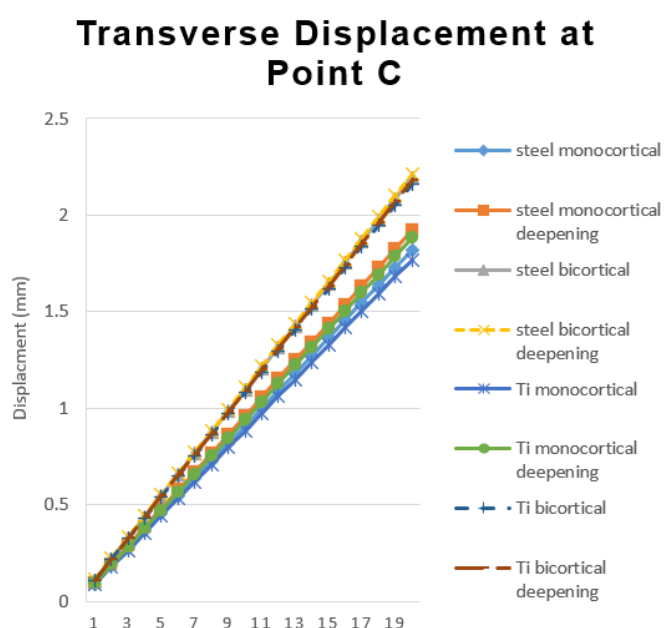
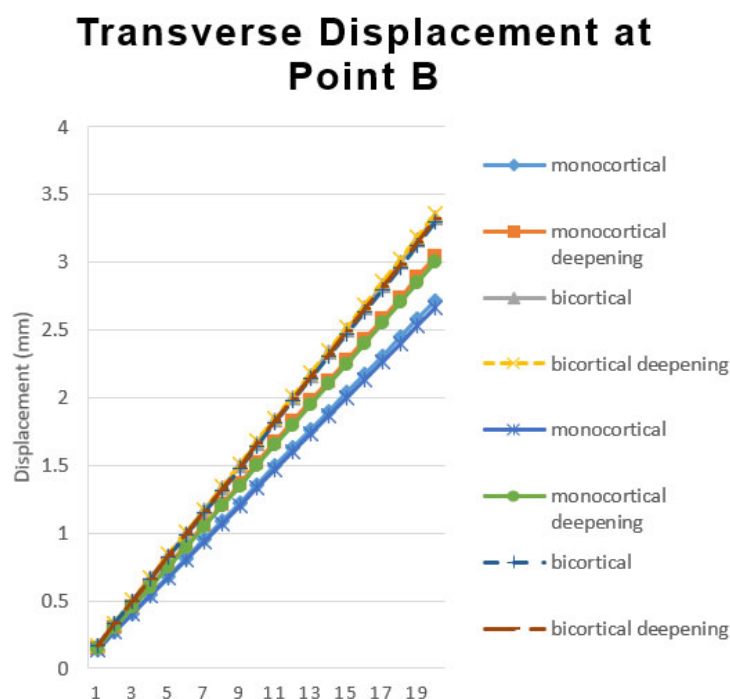


Figure 7. (a–c) Transverse displacement was recorded and evaluated in the three landmarks, points A, B, and C. The total transverse displacement at step 20 was measured at levels A and B, located at the coronal plane, and at levels B and C, located at the axial plane.

The difference at point A for the total transverse displacement caused by expansion screws with stainless steel mini-implants between the bicortical model (2.534 mm) and the monocortical model (2.056 mm) was 20.83%, the difference between the bicortical deepening model (2.556 mm) and the monocortical model (2.056 mm) was 21.68%, the difference between the bicortical model (2.534 mm) and the monocortical deepening model (2.257 mm) was 11.56%, and that between the bicortical deepening model (2.556 mm) and the monocortical deepening model (2.257 mm) was 12.42%. At point A, the difference in

the total transverse displacement between the monocortical deepening model (2.257 mm) and the monocortical model (2.056 mm) was 9.32%, and that between the bicortical deepening model (2.556 mm) and the bicortical model (2.534 mm) was 0.86%.

The difference at point B for the total transverse displacement caused by expansion screws with stainless steel mini-implants between the bicortical model (3.332 mm) and the monocortical model (2.713 mm) was 20.48%, the difference between the bicortical deepening model (3.357 mm) and the monocortical model (2.713 mm) was 21.22%, the difference between the bicortical model (3.332 mm) and the monocortical deepening model (3.037 mm) was 9.26%, and that between the bicortical deepening model (3.357 mm) and the monocortical deepening model (3.037 mm) was 10.01%. At point B, the difference in the total transverse displacement between the monocortical deepening model (3.037 mm) and the monocortical model (2.713 mm) was 11.27%, and that between the bicortical deepening model (3.357 mm) and the bicortical model (3.332 mm) was 0.75%.

The difference at point C for the total transverse displacement caused by expansion screws with stainless steel mini-implants between the bicortical model (2.191 mm) and the monocortical model (1.818 mm) was 18.61%, the difference between the bicortical deepening model (2.210 mm) and the monocortical model (1.818 mm) was 19.46%, the difference between the bicortical model (2.191 mm) and the monocortical deepening model (1.919 mm) was 13.24%, and that between the bicortical deepening model (2.210 mm) and the monocortical deepening model (1.919 mm) was 14.10%. At point C, the difference in the total transverse displacement between the monocortical deepening model (1.919 mm) and the monocortical model (1.818 mm) was 5.41%, and that between the bicortical deepening model (2.210 mm) and the bicortical model (2.191 mm) was 0.86%.

The difference at point A for the total transverse displacement caused by expansion screws with titanium alloy mini-implants between the bicortical model (2.496 mm) and the monocortical model (2.015 mm) was 21.33%, the difference between the bicortical deepening model (2.515 mm) and the monocortical model (2.015 mm) was 22.07%, the difference between the bicortical model (2.496 mm) and the monocortical deepening model (2.228 mm) was 11.34%, and that between the bicortical deepening model (2.515 mm) and the monocortical deepening model (2.228 mm) was 12.10%. The difference at point A for the total transverse displacement between the monocortical deepening model (2.228 mm) and the monocortical model (2.015 mm) was 10.04%, and that between the bicortical deepening model (2.515 mm) and the bicortical model (2.496 mm) was 0.76%.

The difference at point B for the total transverse displacement caused by expansion screws with titanium alloy mini-implants between the bicortical model (3.293 mm) and the monocortical model (2.662 mm) was 9.87%, the difference between the bicortical deepening model (3.315 mm) and the monocortical model (2.662 mm) was 10.01%, the difference between the bicortical model (3.293 mm) and the monocortical deepening model was 9.35%, and that between the bicortical deepening model (3.315 mm) and the monocortical deepening model was 10.01%. The difference at point B for the total transverse displacement between the monocortical deepening model and the monocortical model (2.662 mm) was 11.91%, and that between the bicortical deepening model (3.315 mm) and the bicortical model (3.293 mm) was 0.67%.

The difference at point C for the total transverse displacement caused by expansion screws with titanium alloy mini-implants between the bicortical model (2.160 mm) and the monocortical model (1.768 mm) was 19.96%, the difference between the bicortical deepening model (2.177 mm) and the monocortical model (1.768 mm) was 20.74%, the difference between the bicortical model (2.160 mm) and the monocortical deepening model (1.879 mm) was 13.91%, and that between the bicortical deepening model (2.177 mm) and the monocortical deepening model (1.879 mm) was 14.69%. The difference at point C for the total transverse displacement between the monocortical deepening model (1.879 mm) and the monocortical model (1.768 mm) was 6.09%, and that between the bicortical deepening model (2.177 mm) and the bicortical model (2.160 mm) was 0.78%.

3.5. Transverse Displacement of Midpalatal Suture in Coronal and Axial Planes

The total transverse displacement at step 20 was measured at levels A and B in the coronal plane and at levels B and C in the axial plane. The ratio of displacement at level A to displacement at level B was calculated to compare the amount of displacement measured at levels A and B. The closer the ratio was to 1.000, the more parallel the expansion was.

The difference between the bicortical model and the monocortical model (0.743) was 2.00%, the difference between the bicortical deepening model (0.761) and the monocortical model was 2.39%, the difference between the bicortical model (0.758) and the monocortical deepening model (0.754) was 0.53%, and that between the bicortical deepening model (0.761) and the monocortical deepening model (0.754) was 0.92%. The difference between the monocortical deepening model (0.754) and the monocortical model (0.743) was 1.47%, and that between the bicortical deepening model (0.761) and the bicortical model (0.758) was 0.39% in stainless steel miniscrew implants.

The difference between the bicortical model (0.756) and the monocortical model (0.728) was 3.77%, the difference between the bicortical deepening model (0.759) and the monocortical model (0.728) was 4.22%, the difference between the bicortical model (0.756) and the monocortical deepening model (0.743) was 1.74%, and that between the bicortical deepening model (0.759) and the monocortical deepening model (0.743) was 2.13%. The difference between the monocortical deepening model and the monocortical model (0.728) was 2.04%, and that between the bicortical deepening model (0.759) and the bicortical model (0.756) was 0.40% in titanium alloy miniscrew implants.

The ratio of displacement at level B to displacement at level C was also calculated for the axial plane. The difference between the bicortical model (0.658) and the monocortical model (0.632) was 4.03%, the difference between the bicortical deepening model (0.670) and the monocortical model (0.632) was 5.84%, the difference between the bicortical model (0.658) and the monocortical deepening model (0.648) was 1.53%, and that between the monocortical deepening model (0.648) and the bicortical deepening model (0.670) was 3.34%. The difference between the monocortical deepening model (0.648) and the monocortical model (0.632) was 1.81%, and that between the bicortical deepening model (0.670) and the bicortical model (0.658) was 0.03% in stainless steel miniscrew implants.

The difference between the bicortical model (0.658) and the monocortical model (0.627) was 4.82%, the difference between the bicortical deepening model (0.664) and the monocortical model was 5.73%, the difference between the bicortical model (0.658) and the monocortical deepening model (0.656) was 0.30%, and that between the bicortical deepening model (0.664) and the monocortical deepening model (0.656) was 1.21%. The difference between the monocortical deepening model (0.656) and the monocortical model was 0.91%, and that between the bicortical deepening model (0.664) and the bicortical model (0.658) was 0.52% in titanium alloy miniscrew implants.

4. Discussion

Effective anchorage management is essential for good orthodontic outcomes. Various authoritative works agree that bicortical fixation improves miniscrew implant stability [25]. Bicortically anchored miniscrews are often used in challenging clinical situations requiring strong anchorage [26]. Since detailed biomechanical comparisons between bicortical and monocortical anchorage systems are rarely available in the literature, the objective of this study was to address this gap in the literature.

Three-dimensional models of a hard palate with bicortical and monocortical anchorage were constructed by FEA. To avoid systematic error, conclusions were not based on absolute values. Only the differences in simulation results were considered for comparison purposes.

4.1. Monocortical Anchorage vs. Bicortical Anchorage

This FEA study indicated that cortical bone stress was higher in monocortical miniscrew placements compared to bicortical miniscrew placements. Compared with bicortical miniscrews, monocortical miniscrews provided inferior anchorage resistance and greater cortical bone stress.

The stress differences between the monocortical variants were clearly significant, but the extent of the differences was much lower than between bicortical and monocortical anchorage. With monocortical anchorage relatively high stresses were induced in the region of the cervical peri-implant bone. These were somewhat higher than in monocortical deepening anchorage, but significantly higher than in both bicortical anchorage placements. Cervical bone stress was clearly lower in the two bicortical anchorage placements compared to the two monocortical anchorage placements.

Since miniscrews are not osseointegrated, their anchorage potential usually depends on the quantity of bone into which they are placed [27]. Clinically relevant dimensions of bone available for palatal miniscrew implant anchorage include both the outer cortical layers of the oral hard palate and the nasal floor. Bicortical anchorage (oral and nasal cortical layers) is a major determinant of the success of a miniscrew implant, and if the expander is too distant from the palatal mucosa, miniscrews may not reach the nasal cortical bone. Moreover, the application of force at too far a distance from the implant–bone interface increases the potential for miniscrew deformation and fracture.

4.2. Transverse Displacement of Midpalatal Suture

In MARME, miniscrew implants apply direct force to the maxillary center of resistance, which practically eliminates inclination forces of the posterior teeth and promotes a more parallel opening of the midpalatal suture in coronal and axial views [28].

When an expander is placed at a more posterior position, the concentration of force is near the pterygoid plates, which are highly resistant to palatal expansion [29]. Therefore, a parallel opening of the midpalatal suture differs from that induced by a conventional rapid palatal expansion, in which the opening tends to have a triangular shape, whereas the expansion tends to have a “V” shape (i.e., the greatest width is in the anterior and inferior regions).

4.3. Stainless Steel vs. Titanium Alloy Miniscrews

Primary stability is an important factor in miniscrew implant survival. Despite the many differences between stainless steel and titanium alloy, they had a similar rate of success in meeting the most important mechanical requirements for the good stability of the miniscrew implants. The primary stability of a miniscrew implant, which is a critical factor in the success of a miniscrew treatment, depends on the insertion depth rather than on the implant material [30].

Instead of titanium alloy, stainless steel was the selected miniscrew composition because it had superior toughness (resistance to fracture) when placed in relatively dense cortical bone. Stainless steel also has a long history as the preferred material for orthopedic applications requiring sharp self-drilling screws with a high toughness (resistance to fracture) [31]. In thin cortical bone, however, titanium alloy miniscrews have a lower failure rate compared to stainless steel miniscrews. The likely explanation for the lower failure rate of titanium is its slight biocompatibility advantage in resisting bone resorption at the miniscrew interface in thin cortical bone [32].

In a recent clinical study [31], both stainless steel infrazygomatic miniscrews and titanium alloy infrazygomatic miniscrews had overall success rates of 93.7%; therefore, either stainless steel or titanium alloy are suitable for most clinical applications. The FEA in this study revealed greater mean bending in titanium alloy miniscrew implants compared to stainless steel miniscrew implants used in the same hard palate model. Further clinical

studies are needed to compare the stability between miniscrew implants composed of stainless steel and those composed of titanium alloy in MARME.

5. Conclusions

The finite element analysis could efficiently investigate the biomechanical effects of the suture displacement in the rapid maxillary expansion treatment for comparing different parameters of screw material and screw insertion depth. The conclusion was that the stress magnitude of the miniscrews was in inverse proportion to the inserted depth of the miniscrews; moreover, the material of the miniscrews in the titanium alloy was less than that in the stainless steel. The same tendency was still detected in the stress distribution of the peri-implant bones. Furthermore, the effects of the miniscrews bending during expansion stage showed that the deflection degrees of the miniscrews were in inverse proportion to the inserted depth of the miniscrews. The miniscrews in the titanium alloy demonstrated the deflection effects easier than that in the stainless steel. For comparing the suture displacement of points A, B, and C, the tendency of the expansion displacement in the stainless steel was larger than the titanium alloy. Additionally, the suture displacement of the rapid maxillary expansion was in proportion to the inserted depth of the miniscrews. The risk of miniscrew penetration into the nasal cavity should be avoided when rapid maxillary expansion is performed. Therefore, for obtaining better treatment of the rapid maxillary expansion in patients with a narrow upper arch, both bicortical anchorage and stainless steel miniscrews should be considered the preferred anchorage in challenging clinical situations requiring heavy anchorage, such as maxillary skeletal expansion with MARME.

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