ORIGINAL ARTICLE AJO-DO

Effects of monocortical and bicortical mini-implant anchorage on bone-borne palatal expansion using finite element analysis

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Introduction: Bone-borne palatal expansion relies on mini-implant stability for successful orthopedic expansion. The large magnitude of applied force experienced by mini-implants during bone-borne expansion may lead to high failure rates. Use of bicortical mini-implant anchorage rather than monocortical anchorage may improve mini-implant stability. The aims of this study were to analyze and compare the effects of bicortical and monocortical anchorages on stress distribution and displacement during bone-borne palatal expansion using finite element analysis. **Methods:** Two skull models were constructed to represent expansion before and after midpalatal suture opening. Three clinical situations with varying mini-implant insertion depths were studied in each skull model: monocortical, 1-mm bicortical, and 2.5-mm bicortical. Finite element analysis simulations were performed for each clinical situation in both skull models. Von Mises stress distribution and transverse displacement were evaluated for all models. **Results:** Peri-implant stress was greater in the monocortical anchorage model compared with both bicortical anchorage models. In addition, transverse displacement was greater and more parallel in the coronal plane for both bicortical models compared with the monocortical model. Minimal differences were observed between the 1-mm and the 2.5-mm bicortical models for both peri-implant stress and transverse displacement. **Conclusions:** Bicortical mini-implant anchorage results in improved mini-implant stability, decreased mini-implant deformation and fracture, more parallel expansion in the coronal plane, and increased expansion during bone-borne palatal expansion. However, the depth of bicortical mini-implant anchorage was not significant. (Am J Orthod Dentofacial Orthop 2017;151:887-97)

Transverse maxillary deficiency has been reported to affect 8% to 23% of adolescent patients and fewer than 10% of adult patients.[1-5] Rapid palatal expansion (RPE), which typically uses a tooth-borne appliance with a center jackscrew, is a well-established and reliable technique to correct this problem for adolescent patients.[6-8] For adults, however, nonsurgical RPE with a tooth-borne appliance can result in dentoalveolar tipping that may cause unfavorable periodontal effects because of the interdigitated midpalatal suture and the decreased elasticity of bone in adults.[9-11] Therefore, in adults, skeletal orthopedic expansion is necessary to prevent these issues and to correct transverse maxillary deficiency.[12-14]

Surgically assisted RPE is the conventional treatment of choice to correct transverse maxillary deficiency in adults.[9-11,15] However, surgically assisted RPE is an invasive process that can result in lateral rotation of the 2 maxillary halves with minimal horizontal translation.[9-11] In addition, surgically assisted RPE may be detrimental to the periodontium and has been shown to result in a large amount of relapse during the postretention period.[16,17]

Recently, bone-borne palatal expanders have been reported in several case presentations to have the capability to correct transverse maxillary deficiency in adults, making it a potential alternative to surgically assisted RPE.[18-21] Bone-borne expanders have also been shown to prevent the dentoalveolar tipping seen in adults when attempting to use traditional tooth-borne RPE

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All authors have completed and submitted the ICMJE Form for Disclosure of Potential Conflicts of Interest, and none were reported.

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appliances.[20,22,23] For adolescent patients, bone-borne expansion has been shown to produce greater transverse skeletal expansion while minimizing dental side effects such as dental tipping, alveolar bending, and vertical alveolar bone loss compared with tooth-borne RPE appliances.[24] Bone-borne expansion also has been combined with a facemask for maxillary protraction, which has been shown to reduce adverse effects such as mesialization of anterior teeth.[25]

Bone-borne palatal expansion relies on skeletal anchorage obtained through mini-implants to directly apply force to the basal bone. Thus, mini-implant stability is essential for successful skeletal orthopedic expansion. Mini-implant loss and loosening rates for orthodontic tooth movement range from 6.9% to 28.0%, and their success depends on several factors including the magnitude and direction of the applied force; operator experience; insertion site; quality of cortical bone; surface contact area in cortical bone; length, depth, diameter, thread configuration, and shape of the mini-implant; and patient's age.[26-35] Although no specific reports have analyzed mini-implant failure rates during bone-borne expansion in mature patients, such failure rates are likely to be higher than in orthodontic tooth movement because of the increased magnitude of the applied force necessary to split the interlocking suture. Therefore, new approaches to improve mini-implant stability during bone-borne expansion are needed.

Bicortical mini-implant anchorage has been demonstrated in orthodontic tooth movement applications to be biomechanically more favorable than monocortical anchorage. As such, bicortical anchorage should be considered for clinical situations requiring heavy anchorage.[32,36] Bone-borne expanders, which require heavy anchorage, represent a good clinical situation for bicortical anchorage that has not yet been explored in the literature. In this study, we sought to determine the differences between bicortical and monocortical mini-implant anchorage on skeletal orthopedic expansion.

Finite element analysis (FEA) is a numeric approximation technique that is widely used to assess biomechanical problems. FEA has been applied to study different aspects of bone-borne expanders, primarily focusing on stress distribution and displacement of different expander designs as well as its biomechanical effects on craniofacial sutures.[37-41] However, no study has compared bicortical and monocortical anchorage for bone-borne expanders using FEA. Thus, the aim of this study was to analyze and compare the effects of bicortical and monocortical anchorage on stress distribution and displacement during bone-borne palatal expansion using FEA.

MATERIAL AND METHODS

A finite element model was generated using volumetric data from a cone-beam computed tomography scan (slice thickness, 0.30 mm) of a dry adult skull using Mimics software (version 15.0; Materialise, Leuven, Belgium). Threshold segmentation was performed generating a 3-dimensional (3D) virtual surface model of the dry skull. Individual masks of sutures 1.5 to 2 mm wide were manually generated for the midpalatal, median nasal, lateral nasal, pterygomaxillary, zygomaticotemporal, and zygomaticomaxillary sutures.[40-42] The thicknesses of the cortical bone and the masticatory mucosa in the hard palate were determined using the studies by Farnsworth et al[43] and Studer et al,[44] respectively. Two 3D surface models of the dry skull were generated. The first model contained the interlocking midpalatal suture and represented the skull before midpalatal suture opening (Fig 1, *A*). The second model did not contain the interlocking midpalatal suture and represented the skull after midpalatal suture opening without sutural resistance against expansion force (Fig 1, *B*). Bicortical and monocortical anchorages were compared in both models using measurements at 3 points (Fig 1, *B*). These 3D skull surface models were imported into 3-matic software (version 7.0; Materialise) to generate a finite element volumetric mesh.

The mini-implant (diameter, 1.5 mm; length, 11.0 mm) (ACR Series; BioMaterials Korea, Seoul, Korea) and a specific design of bone-borne palatal expander, the maxillary skeletal expander (MSE; BioMaterials Korea) used in this study were constructed with computer-aided design software (SolidWorks, version 2011; Dassault Systemes, Velizy, France) with the design specifications provided by the manufacturer. These models were exported from SolidWorks as 3D surface stereolithography files. The stereolithography files of the mini-implant and the maxillary skeletal expander were then also imported into the 3-matic software for finite element volumetric mesh generation.

In the 3-matic software, the expander was positioned similar to a patient using clinical photos and cone-beam computed tomography scans as positioning aids (Fig 1). The mini-implants were positioned, using posteroanterior cephalograms as a positioning aid, to have varying insertion depths representing 3 clinical situations: monocortical, 1-mm bicortical, and 2.5-mm bicortical (Fig 2). The expander was in the same position for all 3 clinical situations with only the vertical position of the mini-implants varying between each clinical situation. All 3 clinical situations were analyzed in both skull models.

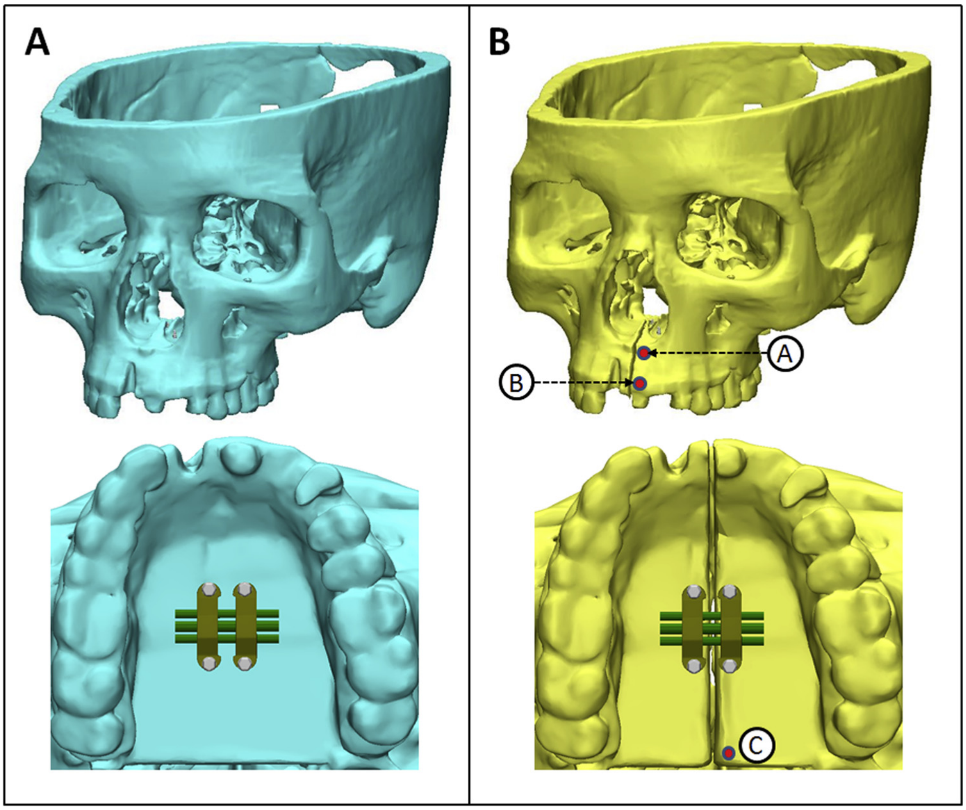

Fig 1. Three-dimensional virtual models of a dry skull with bone-borne expander: **A,** model to be used for FEA simulation of expansion before midpalatal suture opening; **B,** model to be used for FEA simulation of expansion after midpalatal suture opening. Transverse displacement will be measured at points *A*, *B*, and *C*.

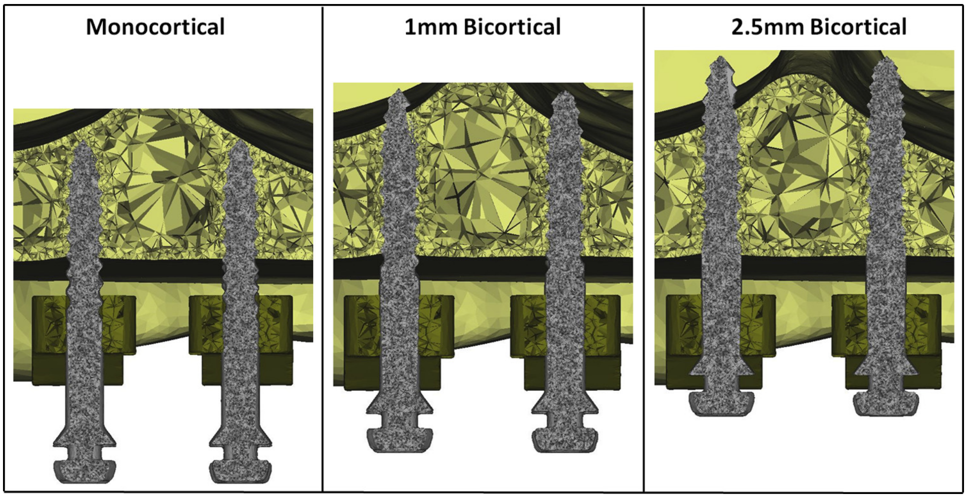

Fig 2. Coronal plane cut view of mini-implant positions in 3 clinical situations: monocortical, 1-mm bicortical, and 2.5-mm bicortical. The expander is in the same position for all 3 clinical situations with only the vertical position of the mini-implants varying between each situation.

Table I. Material properties

	Young's modulus (MPa)	*Poisson's ratio*
Cortical bone	13,700	0.30
Cancellous bone	1,370	0.30
Suture	10	0.49
Masticatory mucosa	25	0.30
Titanium	113,000	0.33
Stainless steel	210,000	0.30

Tetrahedral elements were used for volumetric mesh generation. Each skull was composed of about 4,500,000 elements and 1,200,000 nodes. For the skull generation, the maxilla and the sutures were locally remeshed to contain more fine elements than elsewhere on the skull. Each mini-implant was composed of about 85,000 elements and 16,000 nodes. The expander was composed of approximately 30,000 elements and 9,000 nodes.

The finite element models of the skull, mini-implants, and expander were imported into Abaqus FEA software (version 6.13; Dassault Systemes) to perform FEA simulations. The material properties used are shown in Table I.[37,45,46] Each material was considered to be homogeneous and isotropic. The boundary conditions applied were setting the nodes of the foramen magnum to be completely fixed in all degrees of freedom.[47]

In the model simulating bone-borne expansion before midpalatal suture opening, the expander was activated transversely by 0.5 mm in the transverse plane and was unfixed in the sagittal and coronal planes to prevent interference with the resultant movement.[37,38] In the model simulating bone-borne expansion after midpalatal suture opening, the expander was activated transversely by 0.25 mm for 20 steps resulting in a total of 5 mm of expansion. Similar to the first model, the expansion was also activated in the transverse plane and was unfixed in the sagittal and coronal planes to prevent interference with the resultant movement. In both models, Von Mises stress distribution and transverse displacement were evaluated.

RESULTS

Von Mises stress at the peri-implant site was measured for the skull model containing the interlocking midpalatal suture and was found to be clearly higher in the monocortical anchorage model compared with both bicortical anchorage models (Fig 3). In all models, the Von Mises stress was localized around the initial cortical bone layer. Minimal difference was observed between the 1-mm and the 2.5-mm bicortical models. The total Von Mises stress at the bone-implant interface was calculated for each model: 476,000 MPa for the monocortical model, 234,000 MPa for the 1-mm bicortical model, and 227,000 MPa for the 2.5-mm bicortical model. The difference between the monocortical model and the 1-mm bicortical model was 68.17%, whereas the difference between the monocortical model and the 2.5-mm bicortical model was 70.84%, and that between the 1-mm and 2.5-mm bicortical models was 3.04%.

Von Mises stress of the mini-implants was also measured in the skull model containing the interlocking midpalatal suture and was found to be significantly higher in the monocortical model compared with both bicortical anchorage models (Fig 4). In all models, the Von Mises stress on the implant was localized at the bone-implant interface around the initial cortical bone layer. Total Von Mises stress values were measured at the bone-implant interface and were determined to be 5,831,000 MPa for the monocortical model, 3,576,000 MPa for the 1-mm bicortical model, and 3,845,000 MPa for the 2.5-mm bicortical model. The difference between the monocortical model and the 1-mm bicortical model was 47.94%, the difference between the monocortical model and the 2.5-mm bicortical model was 41.05%, and that between the 1-mm and the 2.5-mm bicortical models was 7.25%. For the monocortical model, the maximum principal stress at the bone-implant interface was 664.49 MPa, and the minimum principal stress was 229.94 MPa. For the 1-mm bicortical model, the maximum principal stress at the bone-implant interface was 270.246 MPa, and the minimum principal stress was 53.95 MPa. For the 2.5-mm bicortical model, the maximum principal stress at the bone-implant interface was 289.87 MPa, and the minimum principal stress was 75.94 MPa. Bending of the mini-implants was clearly evident. Bending in all 3 mini-implants was measured for all 3 models, and the mean amounts of bending were calculated to be 4.55° for the monocortical model, 1.94° for the 1-mm bicortical model, and 1.71° for the 2.5-mm bicortical model.

Transverse displacement was measured on the left side of the skull model not containing an interlocking midpalatal suture and was determined for each step, 20 steps in total (Fig 5). These 20 steps were equivalent to 20 turns of 0.25 mm each, for a total of 5 mm of expansion (2.5 mm on each side). Left-side transverse displacement was measured at points A, B, and C (Fig 1, *B*) and plotted in Figure 6. The total and mean transverse displacements are recorded in Table II. At point A, the total transverse displacements were 1.608 mm for the monocortical model, 1.988 mm for the 1-mm bicortical model, and 2.067 mm for the 2.5-mm bicortical model. The difference at point A for total transverse displacement between the monocortical model and the

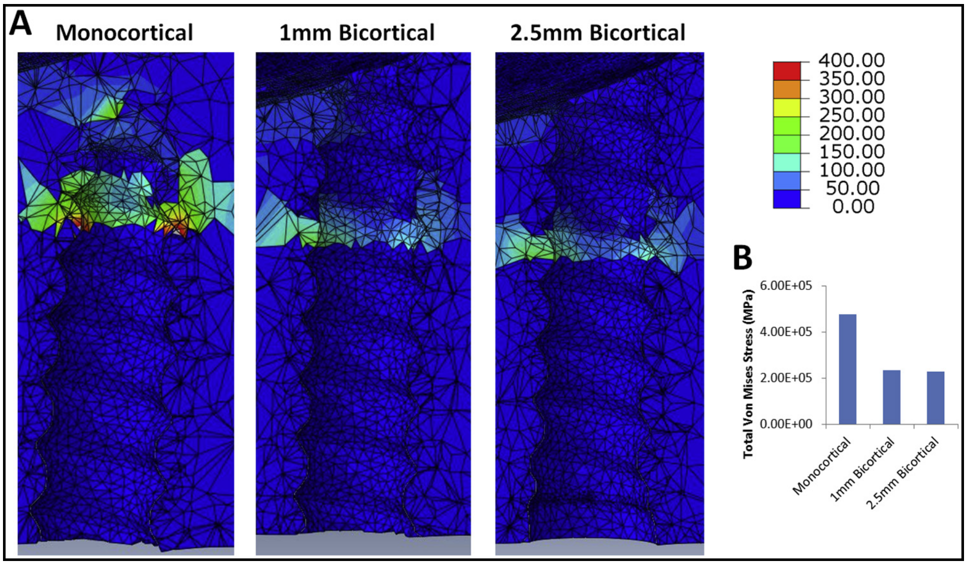

Fig 3. A, Von Mises stress of the peri-implant site for the skull model with midpalatal suture for the monocortical, 1-mm bicortical, and 2.5-mm bicortical models; **B,** bar graph showing total Von Mises stress in megapascals for all 3 anchorage models.

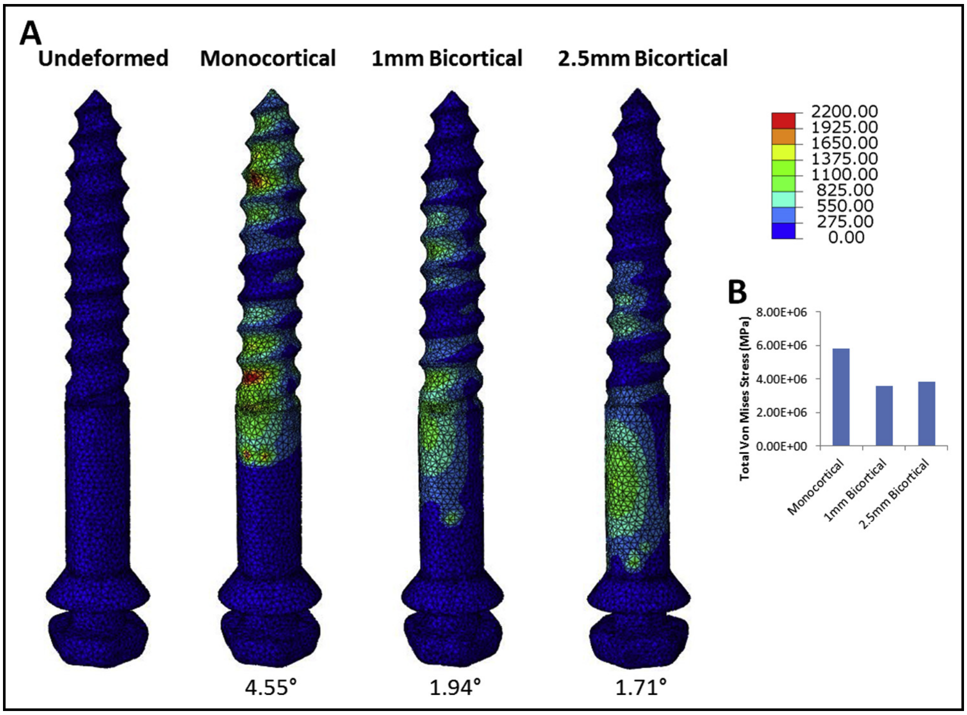

Fig 4. A, Von Mises stress of the mini-implant for the skull model with midpalatal suture for the monocortical, 1-mm bicortical, and 2.5-mm bicortical models, with the degree of bending of the mini-implants reported; **B,** bar graph showing total mini-implant Von Mises stress in megapascals for all 3 anchorage models.

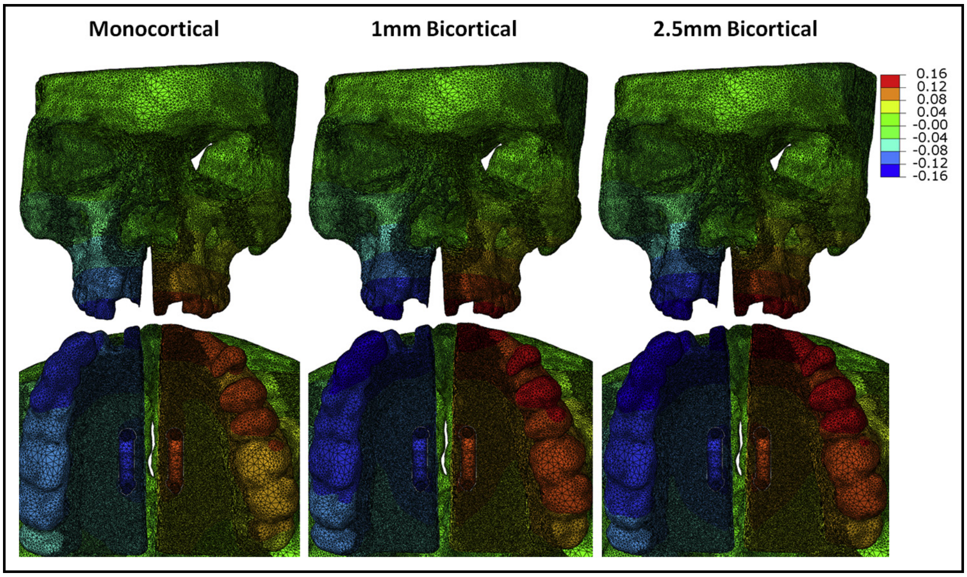

Fig 5. Frontal and occlusal views of step 20 (5 mm of expansion) of the skull model simulation after midpalatal suture opening with a contour map showing transverse displacement.

1-mm bicortical model was 21.13%, the difference between the monocortical model and the 2.5-mm bicortical model was 24.98%, and that between the 1-mm and the 2.5-mm bicortical models was 3.90%. At point B, the total transverse displacements were 2.215 mm for the monocortical model, 2.744 mm for the 1-mm bicortical model, and 2.848 mm for the 2.5-mm bicortical model. The difference at point B for total transverse displacement between the monocortical model and the 1-mm bicortical model was 21.33%, the difference between the monocortical model and the 2.5-mm bicortical model was 25.00%, and that between the 1-mm and the 2.5-mm bicortical models was 3.72%. At point C, the total transverse displacements were 1.141 mm for the monocortical model, 1.444 mm for the 1-mm bicortical model, and 1.442 mm for the 2.5-mm bicortical model. The difference at point A for total transverse displacement between the monocortical model and the 1-mm bicortical model was 23.44%, the difference between the monocortical model and the 2.5-mm bicortical model was 23.31%, and that between the 1-mm and the 2.5-mm bicortical models was 0.14%.

The total transverse displacement at step 20 was measured at levels D and E, located at the coronal midplane of the bone-borne palatal expander (Fig 7). The ratio between D and E was calculated to compare the amount of displacement measured at levels D and E. The closer the ratio was to 1.000, the more parallel the expansion. The ratios were 0.634 for the monocortical model, 0.692 for the 1-mm bicortical model, and 0.701 for the 2.5-mm bicortical model. The difference between the monocortical model and 1-mm bicortical model was 8.72%, the difference between the monocortical model and the 2.5-mm bicortical model was 10.06%, and that between the 1-mm and the 2.5-mm bicortical models was 1.34%.

DISCUSSION

Bone-borne palatal expanders have been shown to be a viable treatment option to correct a transverse maxillary deficiency in adults in several reports showing evidence of clinical success.[18-21,48-50] Since bone-borne expanders rely on skeletal anchorage obtained by mini-implants applying force directly to the basal bone, mini-implant stability is integral to successful skeletal orthopedic expansion. Bicortical mini-implant anchorage has been demonstrated to be superior compared with monocortical mini-implant anchorage for orthodontic tooth movement but has not been explored for bone-borne palatal expansion.[32,36] Therefore, this study was designed to evaluate whether bicortical anchorage likewise increased stability and improved skeletal orthopedic expansion compared with monocortical anchorage.

We used 2 skull models to study the effects of bicortical and monocortical anchorage before and after midpalatal suture opening. The midpalatal suture was removed in the model that represented postmidpalatal suture opening to allow for expansion in the FEA simulation. Three clinical situations of varying mini-implant

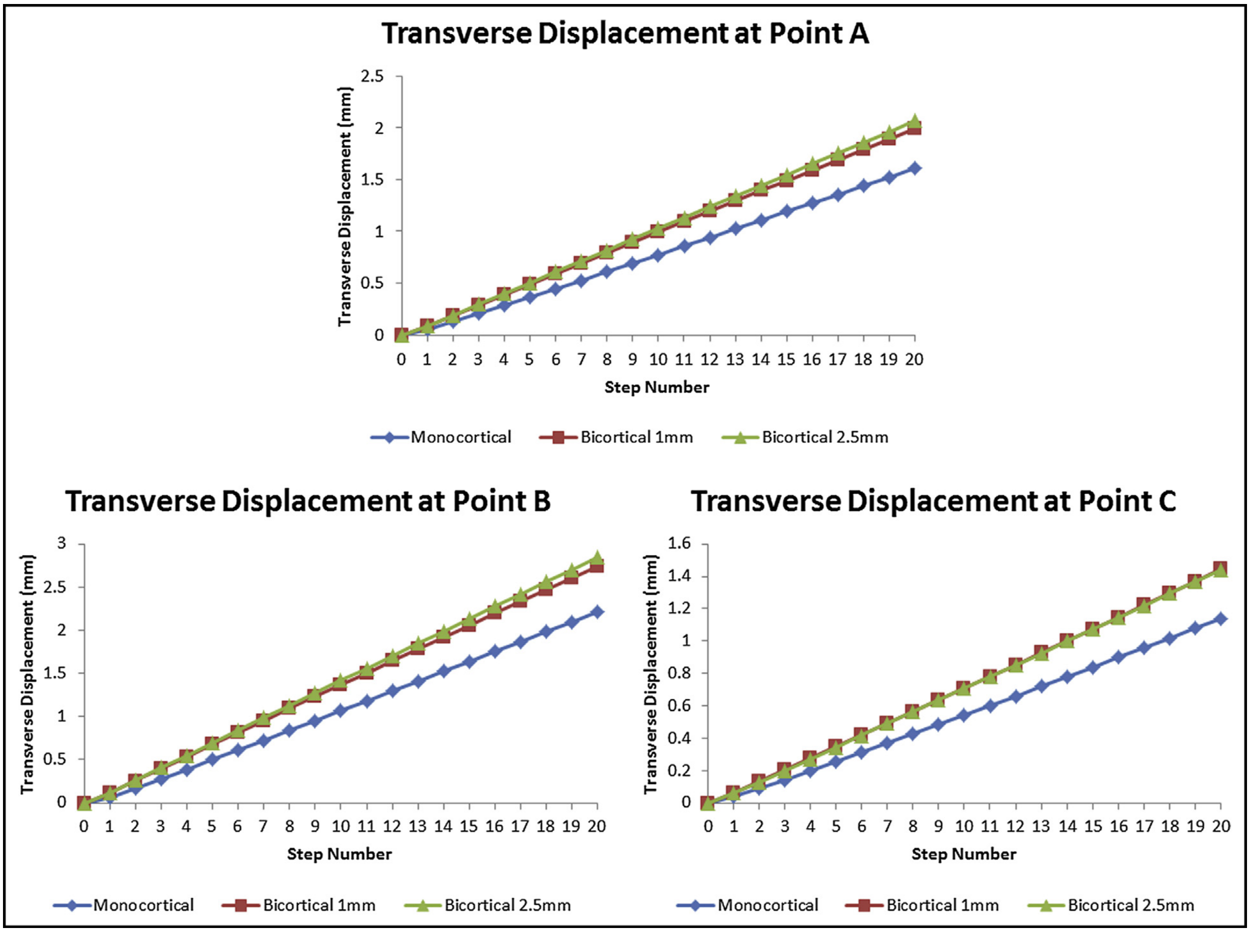

Fig 6. Line graphs showing transverse displacement at each step during expansion.

Table II. Left-side transverse displacements (mm) after midpalatal suture opening

	Monocortical A	*Monocortical B*	*Monocortical C*	*Bicortical 1-mm A*	*Bicortical 1-mm B*	*Bicortical 1-mm C*	*Bicortical 2.5-mm A*	*Bicortical 2.5-mm B*	*Bicortical 2.5-mm C*
Total	1.608	2.215	1.141	1.988	2.744	1.444	2.067	2.848	1.442
Mean	0.080	0.111	0.057	0.099	0.137	0.072	0.103	0.142	0.072

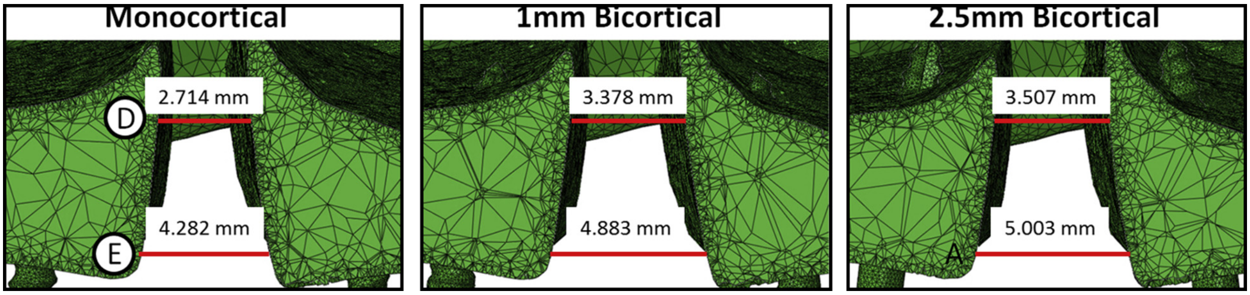

Fig 7. Cut view at the coronal midplane of the bone-borne palatal expander. Total displacements at levels D and E were measured for each model.

insertion depth were used for both skull models: a monocortical model, a 1-mm bicortical model, and a 2.5-mm bicortical model. In all 3 clinical situations, the expander was in the same position, and only the mini-implants varied in vertical position. All 3 of these clinical situations have been observed in patients treated at the University of California at Los Angeles School of Dentistry and were chosen to explore the differences between monocortical and bicortical anchorage as well as to determine whether the depth of bicortical anchorage is significant. Operator experience may also play a role in the varying depths of implantation seen clinically and has been reported to be a factor in mini-implant stability.[27,29]

Overloading of the peri-implant bone can lead to loss of primary stability of orthodontic mini-implants.[51] In addition, there is a decreased risk of mini-implant loosening if the stress in the cervical region of the peri-implant bone region is low.[52] In the skull model containing the midpalatal suture, this study demonstrated that there is significantly lower stress at the peri-implant site in the bicortical models compared with the monocortical model, suggesting that mini-implants placed bicortically decrease the risk of mini-implant loosening. Minimal differences were observed between the 2 bicortical models. These findings are consistent with previous studies showing that in bone-borne expansion, bicortical anchorage is more favorable than monocortical anchorage and that the depth of bicortical anchorage has a minimal impact on stability.[32,36] In addition, this finding is also supported through Wolff's law and the maximum principal stress values reported in this study.[53] The monocortical model had an increased maximum principal stress value compared with the bicortical models. A high principal stress value, as in the monocortical model, may place the bone remodeling in the "pathologic overload window" in which stress fractures and bone resorption, not coupled to formation, occur, leading to overloaded implants and implant loosening.

A greater magnitude of force experienced by mini-implants increases the likelihood of deformation and mini-implant fracture.[54] The authors of this study found that monocortical mini-implants experienced significantly greater stress at the bone-implant interface, specifically around the initial cortical bone layer, compared with bicortical mini-implants. There were minimal differences between the mini-implant stress levels of the 2 bicortical models. In addition, the monocortical mini-implants were found to have more than double the bending compared with the 2 bicortical models. Again, there was a minimal difference between the bending in the 2 bicortical models. These findings suggest that mini-implant fracture is most likely to occur at the initial cortical bone layer and demonstrate that mini-implant deformation and fracture in bone-borne expansion are more likely to occur with monocortical anchorage rather than bicortical anchorage and that the depth of bicortical anchorage has little impact on mini-implant deformation and fracture.

Transverse displacement was measured in the skull model that did not contain the interlocking midpalatal suture for 20 steps. Each step was equivalent to a 0.25-mm turn of the palatal expander for a total of 5 mm of simulated expansion. Analyzing the bone-borne expansion for multiple turns of the expander allowed for more in-depth analysis than previous FEA studies of expansion using only 1 static step. Furthermore, this stepwise model was more representative of a clinical situation.

Transverse displacement was found to be significantly lower in the monocortical model at all 3 points of measurement and after every turn compared with both bicortical models. Minimal differences in transverse displacements were observed between the 2 bicortical models. The difference in transverse displacement between the monocortical and bicortical models may be due to the greater surface contact area in cortical bone of the bicortical models; this allowed for more uniform force transfer. Mini-implant contact surface area in cortical bone has been shown to be a more significant contributor to mini-implant stability than cancellous bone.[33,55] In addition, the monocortical model may have experienced less transverse displacement because of its increased bending. This increased amount of bending created a greater discrepancy between the mini-implant orientation and the line of applied force. Any discrepancy between mini-implant orientation and line of applied force has been shown to decrease load distribution uniformity leading to disproportionate load distribution at the bone-implant interface that would most likely decrease transverse displacement.[56] These findings therefore demonstrate that bicortical anchorage leads to increased expansion compared with monocortical anchorage and that the depth of bicortical anchorage has minimal impact on the amount of expansion.

The ratios between levels D and E were significantly greater for both bicortical models compared with the monocortical model. There was a minimal difference between the ratios of the 2 bicortical models. A larger ratio between levels D and E indicated more parallel expansion in the coronal plane. These results demonstrate that bicortical engagement produces more parallel expansion of the maxillary complex in the coronal plane compared with monocortical engagement.

The V-shaped expansion in the coronal and occlusal planes with traditional tooth-borne expanders makes it difficult to attain precise width coordination between the maxillary and mandibular basal bones without causing excessive dentoalveolar expansion.[54,55] The monocortical model, similar to previous bone-borne expansion models, produced significantly more parallel expansion than traditional tooth-borne expanders.[11,45,57,58] More parallel expansion is favorable for patients because it improves stability and increases the amount of expansion in the posterior region of the maxilla where expansion is often necessary and difficult to achieve.[11,56,59] However, whereas the monocortical model and previous bone-borne expansion models were better than tooth-borne expanders, they still produced a partial V-shaped expansion indicating that further parallel expansion was needed. The bicortical models meet this need by producing even greater parallel expansion. Even distribution of force on both layers of the cortical bones and less bending of the mini-implants may have played significant roles in producing bodily expansion of the 2 halves of the maxilla. On the other hand, transverse displacement discrepancies between points A and C were not significant, indicating that bone-borne palatal expansion produced relatively parallel expansion in the occlusal plane for all 3 models and suggesting that bicortical engagement plays a more significant role in producing parallel expansion in the coronal vs the occlusal planes.

In this study, we applied FEA, a computational numeric approximation technique, to a dry skull model. Our results and numeric findings may differ from actual clinical results because clinical situations vary in many factors such as maturity of the suture, density of the bones, biologic considerations, and shape of the palate and other anatomic structures, which all affect biomechanical systems of maxillary expansion. Therefore, a single finite element model will not represent every clinical situation. In addition, finite element modeling always includes numerous simplifications and assumptions, which decrease the accuracy of the model. In this model, simplifications we applied that decreased the accuracy of the model included modeling the sutures, material properties, and boundary conditions. Because of the inherent limitations of FEA and the assumptions made in this study that decreased the accuracy of the model, future studies using mechanical tests and conventional clinical model analysis are necessary to confirm our results. Constantly improving software and modeling techniques may allow for future studies to decrease the amount of necessary assumptions, leading to more accurate FEA simulations.

CONCLUSIONS

Within the limitations of this study, the following conclusions were drawn.

1. Bicortical mini-implant anchorage results in improved mini-implant stability, decreased mini-implant deformation and fracture, more parallel expansion in the coronal plane, and increased expansion in bone-borne palatal expansion.
2. The depth of bicortical mini-implant anchorage has little impact on mini-implant stability, deformation, and transverse displacement in bone-borne palatal expansion.

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