Maxillary posterior intrusion mechanics with mini-implant anchorage evaluated with the finite element method

Muhsin Cifter and Muyesser Sarac
Istanbul, Turkey

Introduction: The goal of this study was to evaluate the effects of 3 maxillary posterior intrusion mechanics with mini-implant anchorage by using the finite element method. Methods: Finite element models were generated by assembling the images obtained by computed tomography and a laser surface scanner. For each posterior dental segment, a 300-g force was applied and distributed to the mini-implants in proportion to their calculated root surface areas. Results: The most balanced intrusion and the most uniform stress distribution were obtained by concurrent force applications from the vestibular and palatinal sides. In the models with transpalatal arches and buccal force application, vestibular tipping movement and overall stress values were prominent. In all models, increased stress values were identified at the apical region of the first premolar roots and at the apical region of the first molar mesial root. Conclusions: The results of this study suggest that the apical region of the first premolar roots and the apical region of the first molar mesial root should be considered to be prone to resorption during posterior intrusion treatment. Posterior intrusion systems with force application from counterbalancing sites lead to a more uniform stress distribution and balanced intrusion than the mechanics with a transpalatal arch. For a balanced intrusion, root surface areas should be considered when determining the appropriate forces. (Am J Orthod Dentofacial Orthop 2011;140:e233-e241)

Intrusion of the posterior teeth is regarded as a difficult orthodontic tooth movement. Several factors, such as magnitude and direction of the forces and orientation of the anchorage units, should be considered during posterior intrusion to prevent undesirable movement and root resorption. In most studies, it was reported that traditional posterior intrusion mechanics such as bite-blocks and fixed appliances with vertical elastics and multi-loop archwire therapy often have limited intrusion and side effects from insufficient anchorage. Once temporary anchorage devices in orthodontic treatments became established, the need for patient cooperation became obsolete, and side effects on the surrounding tissues were reduced significantly. The posterior intrusion methodology with temporary anchorage devices has also eliminated the need for orthognathic surgery for some borderline open-bite patients. Kuroda et al used titanium screw anchorage to treat open bites and stated that their results were similar to those obtained by 2-jaw orthognathic surgery. Sherwood et al aimed to treat anterior open bites by intruding the molars with titanium mini-plate anchorage; as a result, they accomplished true molar intrusion in adults. They stated that anterior open bites can be treated by posterior intrusion, resulting in reduced anterior vertical face height, a decreased mandibular plane angle, and counterclockwise rotation of the mandible.

Many studies have reported on the application and clinical efficiency of posterior intrusion mechanics; however, studies about biomechanical effects such as stress, strain, and displacements on the teeth and the surrounding tissues are limited. Since in-vivo studies are not quite sufficient in assessing biomechanical effects such as stress and strain, finite element analysis, a common method in engineering, became a valuable option for evaluation of biomechanical factors in orthodontics. Briefly, finite element analysis is a numeric method with which stress, strain, and deformation of structures...
with complex geometries can be studied in various loading and boundary conditions. Its philosophy is based on dividing complicated structures into manageable pieces, called elements, that can be easily defined with differential equations. These finite numbers of elements are then assembled to form an approximate mathematic model of the structure.

The main contributions of our study were to compare and evaluate the stress and displacement effects of 3 maxillary posterior intrusion mechanics with mini-implant anchorage by using the finite element method. The intrusive forces simulated in this study were assigned in proportion to the calculated root surface areas.

MATERIAL AND METHODS

A computer-aided design model of the maxillary bone was generated by using 3-dimensional Doctor software (version 4.0; Able Software, Lexington, Mass). For this process, computed tomography images taken from a maxillary bone at 0.625-mm intervals in the axial direction were assembled perpendicular to the occlusal plane. The premolars and molars were modeled manually as suggested by Wheeler11 with 3ds Max software (Auto-desk, San Rafael, Calif). Each tooth was located in the maxillary model according to the Roth12 prescription and aligned with reference to a maxillary medium Tru-Arch form (Ormco, Orange, Calif). Periodontal membrane, cortical bone, and alveolar bone layers were all defined by using the 3ds Max software. The periodontal membrane was assumed to have a thickness of 0.25 mm evenly. The thickness of the cortical bone was 2 mm at the palatal alveolar bone and decreased from 2 to 1 mm from the top of the alveolar bone to the nasal floor of the vestibular alveolar bone (Fig 1, C).13,14 The computer-aided design model of 0.018 × 0.025-in brackets, bands, and archwires with the same dimensions were generated with a laser surface scanner (NextEngine, Santa Monica, Calif) and 3ds Max software. Brackets were attached to the teeth so that the midpoint of the brackets overlapped the midpoint of the facial-palatal surface of the crowns. Mini-implants were modeled manually with the 3ds Max software.

In the first model, the posterior teeth were connected by full-dimension segmental archwires from the vestibular and palatal sides. Mini-implants were placed between the roots of the first and second premolars and the first and second molars from both vestibular and palatal sides (Fig 1).

In the second model, the posterior teeth were connected from the vestibular side, and the mini-implants were placed from the same side between the roots of the first and second premolars and the first and second molars. In this model, to balance the moments produced in the vestibular direction, transpalatal arches with a diameter of 1.4 mm were constructed connecting the first premolars and the first molars (Fig 2).

In the third model, as in the second model, the posterior teeth were connected from the vestibular side, and the mini-implants were placed from the same side between the roots of the first and second molars. In this model, only 1 transpalatal arch connecting the first molars was modeled (Fig 3).

In accordance with the clinical applications, transpalatal arches were adapted evenly 5 mm from the palatal bone to achieve clearance for the intrusion movement. In clinical applications, to minimize tipping movements, a rigid connection should be preferred at the transpalatal arch-band interface. Accordingly, to simulate a welded connection, the transpalatal arch-band interface was considered a fully bonded surface.

Finite element models were generated by importing solid models into ALGOR software (Autodesk). Three-dimensional discrete mesh generations of the solid models were realized by using hexahedral “brick” and tetrahedral elements. The total numbers of elements used in the finite element models were 94,630, 203,150, and 198,600 for the first, second, and third models, respectively. By this process, convergence analyses of finite element method models were completed. All nodes, except for the constrained ones, had 3 translational degrees of freedom.
(x, vestibulopalatinal; y, mesiodistal; z, vertical). Boundary conditions were assigned to the nodes on the floor of the nasal cavity as zero displacement in all directions. Mechanical properties of the materials in the models were assigned as shown in Table I. All materials in the finite element analysis were assumed to be homogeneous, isotropic, and linearly elastic. Bracket-tooth, bracket-archwire, and bone-implant interfaces were defined as fully bonded surfaces.

For all models, finite element analysis was realized by applying a total of 300 gf to each dental segment. Distributions of the forces were calculated in proportion to the root surface areas determined by the 3ds Max software (Figs 1-3). The root surface area ratio of the molars to the premolars was calculated as 1.936 and rounded to 2; the ratio of the vestibular roots along the segment to the palatinal roots was calculated as 1.36 and rounded to 1.5.

By using the finite element method, the initial vertical displacement of the posterior teeth and the Von Mises stress distribution along the root surface were evaluated. To determine tipping movements precisely, vertical displacements of the nodes, having the same coordinates in each model at the root apexes and the cusp tips, were assessed, and superimpositions were used.

**RESULTS**

Among all models, the lowest stress magnitudes were produced in the first model (Fig 4, A). The maximum Von Mises stress was 0.07855 N per square millimeter. The apical third of the first premolar roots and the same region of the first molar mesial root showed the highest stress magnitudes. Trifurcation areas of the first molar and regions adjacent to the force application sites also showed relatively high stress values.

In the second model, the maximum Von Mises stress was 0.49114 N per square millimeter (Fig 5, A). This value was about 6.3 times higher than the maximum stress value in the first model. Increased stress values were observed at the apical third of the first premolar roots. The palatal surfaces of the first molar and first

---

**Table I. Mechanical properties**

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Alveolar bone</td>
<td>1370</td>
<td>0.3</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>13700</td>
<td>0.26</td>
</tr>
<tr>
<td>Periodontal membrane</td>
<td>0.6668</td>
<td>0.49</td>
</tr>
<tr>
<td>Teeth</td>
<td>19613.3</td>
<td>0.15</td>
</tr>
<tr>
<td>Stainless steel</td>
<td>200000</td>
<td>0.3</td>
</tr>
</tbody>
</table>

---

**Fig 2.** Model 2: A, vestibular aspect with force distribution in proportion to the root surface area; B, diagonal aspect showing the double transpalatal arch with 1.4 mm diameter for intersegmental connection.

**Fig 3.** Model 3: A, vestibular aspect with force distribution in proportion to the root surface area; B, palatinal aspect showing 1 transpalatal arch with 1.4 mm diameter for intersegmental connection.
premolar roots and regions adjacent to the force application sites also showed increased stress magnitudes.

The third model had the highest maximum Von Mises stress value among all models, with 0.52708 N per square millimeter, which was approximately 6.7 times higher than the first model (Fig 6, A). Increased stress values were identified at the first molar roots, especially on the vestibular surfaces and at the apical third of the first premolar roots.

In the first model, the maximum intrusion values were identified at the second molar mesial root (Fig 4, B). Intrusion values calculated for the vestibular roots were
slightly higher than those of the palatal roots (Table II). This slight vestibular tipping of the teeth can also be realized from the slight variation of intrusion between the vestibular and palatal cusps (Fig 7, Table III). In the anteroposterior direction, there was no significant tipping or bowing of the dental segment (Fig 7).

In the second model, maximum intrusion movements occurred at the first and second molar vestibular roots (Fig 5, B; Table II). Intrusion of the vestibular roots was considerably higher than at the palatal roots (Fig 8). In contrast to the intrusion at the palatal roots, extrusion was evident at the palatal cusps, as the result of the prominent vestibular tipping movement of the dental segment (Table III).

In the third model, maximum intrusion was evident at the first molar vestibular roots, and intrusion values decreased progressively from the first molar to the anterior and posterior of the dental segment (Figs 6, B, and 9). All roots along the dental segment, other than the second molar palatal root, showed intrusion (Table II). As in the second model, also in this model, the vestibular roots showed considerably more intrusion than did the palatal roots. The third model experienced the greatest vestibular tipping movement among all models and, hence, the greatest extrusion at the palatal cusps (Figs 6, B, and 9) was observed in this model.

**DISCUSSION**

Intrusion of the posterior teeth has been a difficult issue in orthodontics because of the lack of anchorage. Temporary anchorage devices have allowed clinicians to gain anchorage from many different sites for balanced...
and at the apical region of the observed in the apical region of the and displacements were evaluated.

number of nodes in the critical regions where stress used elements as small as 1.1 mm to enhance the element analysis is the numbers of elements and nodes comprising the models. Therefore, in our study, we taken at 0.625-mm intervals. In addition, variable max-

Table III. Vertical displacement of the nodes at the cusp tips (mm)

<table>
<thead>
<tr>
<th></th>
<th>Model 1</th>
<th>Model 2</th>
<th>Model 3</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Premolar buccal</td>
<td>$14 \times 10^{-3}$</td>
<td>$5.8 \times 10^{-3}$</td>
<td>$4 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Premolar palatinal</td>
<td>$12.7 \times 10^{-3}$</td>
<td>$-6.7 \times 10^{-4}$</td>
<td>$-11.6 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Premolar buccal</td>
<td>$14.2 \times 10^{-2}$</td>
<td>$12.7 \times 10^{-3}$</td>
<td>$15.7 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Premolar palatinal</td>
<td>$13 \times 10^{-3}$</td>
<td>$-0.6 \times 10^{-3}$</td>
<td>$-6.7 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar mesiobuccal</td>
<td>$13.4 \times 10^{-3}$</td>
<td>$12.4 \times 10^{-3}$</td>
<td>$17.5 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar distobuccal</td>
<td>$14 \times 10^{-3}$</td>
<td>$13.7 \times 10^{-3}$</td>
<td>$22.6 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar mesiopalatinal</td>
<td>$12.6 \times 10^{-3}$</td>
<td>$-5.7 \times 10^{-4}$</td>
<td>$-10.2 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar distopalatinal</td>
<td>$13.1 \times 10^{-3}$</td>
<td>$-4.1 \times 10^{-3}$</td>
<td>$-9.2 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar mesiobuccal</td>
<td>$16 \times 10^{-3}$</td>
<td>$15.3 \times 10^{-3}$</td>
<td>$2.4 \times 10^{-3}$</td>
</tr>
<tr>
<td>1. Molar distobuccal</td>
<td>$13.2 \times 10^{-4}$</td>
<td>$10.3 \times 10^{-4}$</td>
<td>$1 \times 10^{-4}$</td>
</tr>
<tr>
<td>1. Molar mesiopalatinal</td>
<td>$14.4 \times 10^{-3}$</td>
<td>$-11.4 \times 10^{-4}$</td>
<td>$-10.3 \times 10^{-4}$</td>
</tr>
<tr>
<td>1. Molar distopalatinal</td>
<td>$11.2 \times 10^{-3}$</td>
<td>$-21.7 \times 10^{-4}$</td>
<td>$-13.6 \times 10^{-4}$</td>
</tr>
</tbody>
</table>

Positive values indicate intrusion; negative values indicate extrusion.

invasion with minimal side effects. But there are still unclear data concerning the biomechanical issues.

This finite element method study was carried out to evaluate the effects of various posterior intrusion mechanics with mini-implant anchorage. In finite element method studies, the reliability of the results depends on the accuracy of the models. In this study, to maximize the similarity of the models with the maxilla, models were generated from computed tomography images taken at 0.625-mm intervals. In addition, variable maxillary cortical bone thicknesses were generated manually; this is an important parameter in tooth movement. Another parameter affecting the precision of the finite element analysis is the numbers of elements and nodes comprising the models. Therefore, in our study, we used elements as small as 1.1 mm to enhance the number of nodes in the critical regions where stress and displacements were evaluated.

In all models, increased Von Mises stress values were observed in the apical region of the first premolar roots and at the apical region of the first molar mesial root. This observation can be attributed to the small surface area and geometric shape of these regions. Denoted sites should be considered highly prone to root resorption, according to several clinical studies.17-19 Even though total root surface areas and the applied forces were the same, the produced stresses will alter if the tooth geometries are different. Since stress is the cause of the biomechanical response rather than the applied force, root geometries should also be considered when determining orthodontic force magnitudes.15,20

Another common finding related to stress distribution was the increase in stress magnitudes adjacent to the force application sites. The smallest increase in stress was identified in the first model, and the greatest increase was observed in the third model. This finding for the first model was due to the simultaneous multiple force applications from different sites; this led to uniform stress distributions among all 3 models.

In the pilot study conducted with the first model, even intrusive forces (75 gf from each mini-implant) were applied to the anterior, posterior, vestibular, and palatalin.
sides of the posterior dental segment; as a result, severe anterior and palatinal tipping was observed. This behavior supports several clinical studies undertaken for this type of procedure.\textsuperscript{2,21} One explanation for this behavior was due to the variable distribution of root surface areas along the dental segment; this led to uneven stress distributions and various tipping movements. Therefore, in the final study, we have calculated the intrusive forces according to the root surface area ratios. As a result, the applied forces according to the selected root surface area ratios gave a virtually uniform intrusion movement, which was particularly evident in the first model. Nonetheless, it should be considered that the distribution of root surface areas is not the only variable for a balanced intrusion. Variations in tooth morphologies and root angles, inclination differences of the vestibular and palatinal slopes of the alveolar bone, and anisotropic and nonlinear properties of the tissues also can have significant effects on the stress distribution and the path of the intrusion movement.

The most balanced intrusion was identified in the first model, with only a slight vestibular tipping movement observed. One reason for this slight vestibular tipping might be the assumption made for the root surface area ratio of the vestibular roots to the palatinal roots, calculated as 1.36 and rounded to 1.5. However, as mentioned previously, several factors other than root surface area ratios can influence the intrusion movement. At this stage, it could be considered that the most salutary feature of the mechanics used in the first model is the ability to clinically change the properties of the forces applied through 4 different aspects. This leads to virtually full control over the movement of the dental segment during treatment. However, particularly if intrusion of both right and left segments is considered, 4 mini-implants for each segment will be too much for a patient. Thus, although the mechanics used in the first model are biomechanically ideal, clinical application can be difficult.

In the second and the third models, a transpalatal arch was used to balance the produced moments and
inhibit vestibular tipping movements. Compared with the first model, the vestibular tipping movements in the second and third models were more prominent; these caused extrusion of the palatal cusps. In clinical situations, extrusion of the palatal cusps can create interferences between the antagonist teeth and lead to decreased overbite. However, the static finite element analysis used in this study only simulated the initial tooth movement in the periodontal membrane because of the extremely large difference between Young’s modulus of the periodontal membrane and the bone layers. In clinical situations, if a transpalatal arch with sufficient resistance is used, it will exhibit its uprighting effect through a long-term process of bone remodeling, and most of the initial interferences will disappear with time by intrusion of the palatinal cusps. The other side effect of the vestibular tipping movement identified in the second and third models was the increase in overall stress magnitudes, which clinically increase the probability of root resorption. Thus, in most open-bite patients, it is crucial to prevent vestibular tipping during posterior intrusion. With simultaneous force application from the vestibular and palatinal sides, this can easily be controlled. However, through mechanics with vestibular force application and a transpalatal arch, the horizontal component of the forces at each segment should be intersegmentally balanced. For this process to have a sufficient force transition between the segments, the resistance of the transpalatal arch should be adequate, and the connection between the transpalatal arch and the teeth should be rigid. In the second and third models, prominent vestibular tipping was due to insufficient resistance of the transpalatal arch. Clinically, with similar force levels, a thicker transpalatal arch would lead to better stress distribution and better vestibular tipping control. In this study, the connection between the teeth and the transpalatal arch was considered to be fully bonded. In clinical applications, a welded or soldered connection would be appropriate to prevent any rotational movement at this junction.

The highest stress magnitudes and the most severe vestibular tipping movements were observed in the third model, having 1 transpalatal arch. Also for this model, application of the total intrusive force from 1 point caused a bowing effect in the vertical direction. The first and second molars tipped distally and mesially, respectively, from the bowing effect. The reason for this initial displacement in the periodontal membrane was the insufficient resistance of the vestibular arch. But, as previously described, the movement of the teeth differs with bone remodeling by time. In clinical situations, sufficient time is needed to allow the full-dimension archwires to compensate for the initial tipping of the molars. Nonetheless, second-order antitip bends can also be used to prevent tipping movements at the first and second molars. In this study, an 0.018-in system was simulated with full-dimension archwires. If a 0.022-in system or a stiff cap splint were used, the initial intrusion and the stress distribution along the segments could be more homogeneous; thus, the initial bowing of the segment would be minimized.

In most open-bite patients, to obtain an ideal occlusion, intrusion of the posterior segment is needed more at the molars than at the premolars because of the hinge movement of the mandible and the excessive eruption of the molars. With separate force applications from premolar and molar sites, such as those in the first and second models, it is possible to prescribe intrusion amounts, thereby adjusting the mesiodistal cant of the segments.

Because of individual variations, it is essential to use unique mechanics and force systems for each patient. Even with perfect mechanics and exact force systems, after the initial tooth movement, the biomechanical effect of the force system changes, and modifications are required during treatment. In this study, static finite element analysis only simulated the initial tooth movement in the periodontal membrane and the initial stress distribution along the root surfaces. During the treatment cycle, ongoing movements and stresses can differ because of the changes in force systems and biologic responses. Other limitations of this study were the constant values used for the physical properties of the tissues, which would normally alter clinically through the histologic process, and the assumption that the periodontal membrane was homogeneous, isotropic, and uniform in thickness. These limitations can cause differences between clinical applications and simulation studies. Also, because of individual variations, it is impossible to simulate an exact mathematical model to validate each case. However, similarities between the results of this study and clinical studies with parallel mechanics show that the finite element models generated were accurate enough to simulate clinical conditions.

**CONCLUSIONS**

1. The apical region of the first premolar roots and the apical region of the first molar mesial root experienced increased stress levels; thus, these sites should be considered to be prone to resorption.
2. To obtain a balanced intrusion, root surface area should be considered when determining the appropriate forces.
3. Segmental posterior intrusion mechanics with force applications from counterbalancing sides (anterior-posterior, vestibular-palatine) lead to a more...
uniform stress distribution and balanced intrusion than the mechanics with a transpalatal arch.

Ideally, the results gained through finite element analysis should be integrated with clinical experiences to maximize accuracy.

We thank Ata Muğan, Istanbul Technical University, Faculty of Mechanical Engineering, for his guiding comments.

REFERENCES