

Article

Biomechanical Evaluation of the Second Molar Uprighting with Retromolar Mini-Implants in the Presence and Absence of the Third Molar

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Highlights

What are the main findings?

- Displacements along the X, Y, and Z axes, as well as regions of peak stress, were determined for two three-dimensional (3D) models of the left mandibular segment, with a retromolar mini-implant (MI) placed in both models.
- Finite element analyses revealed that the presence of the third molar significantly modulates the biomechanics of the second molar uprighing via retromolar MIs.

What are the implications of the main findings?

- Within the limitations of the simulated scenarios, the biomechanical response of the permanent second molar to orthodontic displacement is strongly influenced by the presence or absence of the third molar within the dental arch, exhibiting a reduced tendency for deformation during distalization and a slower displacement when the third molar is present.
- Based on a hypothesis derived from a computational model, the study suggests that the third molar should be retained when feasible, with benefits such as reduced stress on the second molar and the potential for both molars to be aligned; however, this conclusion is limited to simulated conditions and does not directly translate into clinical decision-making.



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Abstract

Background/Objectives: The uprighting of mesially tipped mandibular second molars following first molar loss is a complex surgical and orthodontic challenge. Conventional methods often result in reciprocal anchorage loss. Mini-implants (MIs) have emerged as essential temporary anchorage devices (TADs) that provide absolute anchorage and enable more predictable tooth movements. **Methods:** Numerical simulations were performed to evaluate the forces required for mandibular second molar uprighting under two conditions: first, only with the second molar present, and second, with both the second and the third molars present. Although the periodontal ligament exhibits nonlinear and viscoelastic behavior in vivo, a linear elastic approximation was adopted to allow for a reliable evaluation of comparative stress distribution and initial displacement patterns within the scope of this exploratory biomechanical study. Stress distribution in the roots, periodontal ligament, and alveolar bone was assessed for each scenario. Two three-dimensional (3D) models of the left mandibular segment were created from scans of a human mandible and its teeth. The first model included the canine, the first and second premolars, and the second molar. A second model additionally incorporated the third molar. A retromolar MI was placed in both models. Molar uprighting was simulated using a spring connecting the implant to a button bonded on the mesial surface of the second molar. A force of 200 g was applied because in clinical orthodontic practice, forces that exceed approximately 2 N may cause pain or undesirable tooth mobility. Displacements along the X, Y, and Z axes, as well as regions of peak stress, were analyzed. **Results:** Model 1 showed maximum displacements at the furcation/mid-root, distal root apex, and distal crown, with von Mises stresses of 0.470 to 0.371 MPa. In Model 2, peak displacements occurred at the mesial root and crown, with stresses of 0.185 and 0.149 MPa, respectively. The magnitude of displacements was in the order of 10^{-5} mm. Such values represent initial mechanical responses rather than clinically observable tooth movements. However, the differences between models (e.g., the stress reduction) are expected to be clinically meaningful. **Conclusions:** Since clinical measurements regarding the stress distribution on teeth and surrounding tissues during orthodontic molar uprighting movements are impossible to perform, the finite element method (FEM) can offer insight into these aspects. The presence of the third molar significantly modulates the biomechanics of second molar uprighting via retromolar MIs. When the third molar is present, the second molar exhibits a reduced tendency for deformation during distalization, although this leads to a slower displacement. This FEM provides biomechanical insights but does not support direct clinical decision-making. The present findings should be viewed as theoretical biomechanical tendencies that require confirmation through clinical, experimental, and longitudinal studies before translation into clinical practice.

Keywords: mini-implant; molar uprighting; human mandible; 3D models; Finite element analysis; von Mises stress; orthodontics; tooth movements; numerical simulations

1. Introduction

The first permanent molars are the earliest teeth to erupt in the permanent dentition. They play a crucial role in establishing sagittal and transverse occlusal relationships [1]. Owing to their early eruption and prolonged exposure to the oral environment, these teeth are particularly susceptible to carious lesions. Consequently, the premature loss of first permanent molars, often associated with inadequate oral hygiene, can have a significant negative impact on occlusal stability and craniofacial development [2]. In orthodontic

practice, such a loss frequently leads to mesial tipping and rotation of the adjacent second permanent molars and, in some cases, even the third molars [3].

These positional changes are commonly associated with periodontal complications, distal drifting of canines and premolars, and extrusion of the opposing first permanent molar [4]. Furthermore, premature contacts and occlusal interferences in centric relation and during mandibular excursions may result in traumatic occlusion [5]. Therefore, the oral rehabilitation of these patients becomes more complex.

Prosthetic preparation of a tipped molar often requires excessive reduction of the mesial surface. This increases the risk of pulp chamber exposure, the need for endodontic treatment for prosthetic purposes, and an abnormal distribution of occlusal forces [6,7]. Tilted mandibular molars represent a frequent clinical challenge, particularly for patients who require implant-supported prosthetic rehabilitation. Most commonly, this medical condition develops following premature tooth loss. It produces a mesial or distal tilt of adjacent teeth, which compromises the establishment of an optimal 3D prosthetic treatment plan [8]. Periodontal deterioration may occur, characterized by inflammation, localized bone loss, and periodontal pocket formation, especially at the mesial root of the tipped molar. In severe cases, occlusal interferences, premature contacts, and supraeruption of the opposing molar may further complicate or even preclude prosthetic restoration of the edentulous space [9].

Therefore, timely management of first permanent molar loss through orthodontic space closure or through prosthetic replacement is essential in order to prevent functional and anatomical disturbances [10]. In this respect, molar uprighting constitutes an effective therapeutic approach, as it promotes improved periodontal health, restores proper axial loading of occlusal forces, and recreates sufficient space for implant-supported restorations [11]. Conventional orthodontic uprighting techniques that are currently employed in clinical practice include cantilever springs, prefabricated Sander springs, uprighting loop springs, NiTi coil springs, T-loop archwires, and distal jet appliances, as well as other auxiliary systems [12]. Molar uprighting requires good anchorage control. Afterward, the placement of a fixed orthodontic appliance on the arch is usually necessary. Among other methods, the Uprighter Jet developed by Carano offers full control of molar uprighting, minimizes extrusion, and does not require brackets or special patient cooperation [13]. Another uprighting alternative involves combining it with rapid maxillary expansion (RME). In this case, vertical elastic forces are directed from a hook on the RME device to an orthodontic attachment bonded to the tooth, thus bringing it into an upright position. This method saves time, does not require additional anchorage preparation, and is more physiological. According to Pogrel, the surgical uprighting of lower second molars is a quick procedure with minimal risk and a good long-term prognosis. Most uprighted teeth remained stable, with excellent bone formation and good periodontal health after 18 months of follow-up [14].

However, all conventional orthodontic approaches used to achieve molar uprighting have been associated with issues such as molar extrusion, bulky appliances, unwanted movements of anchorage units, the frequent need for additional anchorage reinforcement accessories, and extended treatment duration [15]. Therefore, modern methods employ temporary anchorage devices (TADs) such as palatal implants, mini-implants (MIs), and miniplates. These devices expand the limits of tooth movement, especially for uncooperative patients or for those with multiple missing teeth or complex malocclusions. TADs allow forces to be applied exclusively to the targeted tooth and in the desired direction. Such techniques do not need to involve the other teeth in the oral cavity [16].

Recent investigations continue to refine our understanding of stress distribution around MIs and the surrounding bone under orthodontic loading conditions. In particular,

the stress distribution that results from MIs has been extensively studied in the literature, including in recent studies. Existing biomechanical research primarily focuses on anchorage design, force systems, or implant stability. In contrast, the influence of anatomical variations, particularly the presence or absence of the mandibular third molar during second molar uprighting, has not been sufficiently investigated. Consequently, the extent to which the third molar modifies stress distribution within the periodontal ligament, roots, and alveolar bone during uprighting mechanics remains unclear.

The rationale for investigating the presence versus the absence of the third molar is based on both clinical relevance and biomechanical considerations. The decision to extract or retain the third molar during second molar uprighting remains controversial, as its presence may alter distal resistance, anchorage conditions, and stress distribution within the posterior mandibular segment. While the third molar could increase mechanical resistance and restrict distal movement, its absence may facilitate displacement of the second molar. Despite these plausible effects, quantitative biomechanical evidence remains limited. Finite element analysis (FEA) allows for controlled standardization of all variables except the presence of the third molar. This enables the isolation of its specific mechanical contribution. Hence, the inclusion of both configurations was designed to address a clinically significant yet insufficiently quantified question in orthodontic biomechanics. Today, FEA has become a powerful and reliable computational tool in orthodontic biomechanics, as it enables a detailed simulation of the stress–strain behavior that occurs between teeth and their supporting periodontal structures under controlled force systems. By integrating anatomically accurate 3D models with experimentally derived material properties, FEA allows for a precise evaluation of tooth displacement patterns, load transfer mechanisms, and stress distribution within the periodontal ligament and alveolar bone. Such aspects cannot be directly measured *in vivo*. In the context of molar uprighting following the premature loss of the first permanent molar, such biomechanical insight is particularly critical, as inappropriate force magnitude or anchorage design may compromise periodontal health and treatment predictability [17]. However, previous FEA studies have not addressed a direct comparison between the presence and absence of the third molar under identical loading conditions.

Considering all the above, the present study aims to evaluate, through 3D FEA, the biomechanical behavior of mandibular second molar uprighting using a retromolar MI in models with and without the third molar present. The method includes quantifying tooth displacement, force vectors, and stress distribution within the periodontal ligament and the surrounding alveolar bone. The study further seeks to identify biomechanically favorable force application strategies that optimize anchorage efficiency while minimizing potentially harmful stress concentrations, thereby providing evidence-based guidance for clinical molar uprighting protocols.

The study hypothesis is that the presence of the third molar significantly modulates the biomechanics of second molar uprighting via retromolar MIs.

2. Materials and Methods

In the present study, 3D FEA-based numerical simulations were undertaken to rigorously quantify and compare the biomechanical force systems associated with the orthodontic uprighting of the second permanent molar under two clinically relevant scenarios: (1) in the absence of the third molar and (2) in the presence of an adjacent third molar within the dental arch. This comparative analysis was designed to elucidate the influence of the third molar presence on force magnitude, displacement behavior, and stress distribution within the tooth–periodontal ligament–alveolar bone complex. Additionally, the study aims to

assess the stress distribution within the roots, periodontal ligament and alveolar bone in each scenario.

Accordingly, the finite element method (FEM) was utilized as a computational framework to evaluate the biomechanical effects of orthodontic uprighting of the mandibular molar. Two 3D left hemimandibular models were selected, considering that molar uprighting is a unilateral and localized procedure, with minimal influence from the contralateral side. This approach has reduced the computational demand while preserving anatomical fidelity and allowing for enhanced mesh refinement in the region of interest. The anatomical dataset was anonymized and utilized exclusively for computational simulation. As no patient interaction or identifiable data were involved, and as demographic variables were not relevant to the biomechanical objectives, no additional ethical concerns were associated with the model selection. Both models included the canine, the first premolar, the second premolar and the second molar. In the second model, the third molar was additionally incorporated to allow for a comparative analysis. The added third molar was modeled in the eruption phase, positioned under the occlusal plane. This anatomical configuration was intentionally preserved in order to accurately reproduce the clinical scenario of molar uprighting during an incomplete eruption.

The geometries of the mandible and dental units were derived from 3D scans of an actual human mandible and teeth. The scanning was performed using a Roland PICZA 3D laser scanner, model LPX-1200 (Scantech Co., Ltd., Hangzhou, China) (Figure 1a), which ensures an accurate replication of anatomical structures for subsequent FEA.

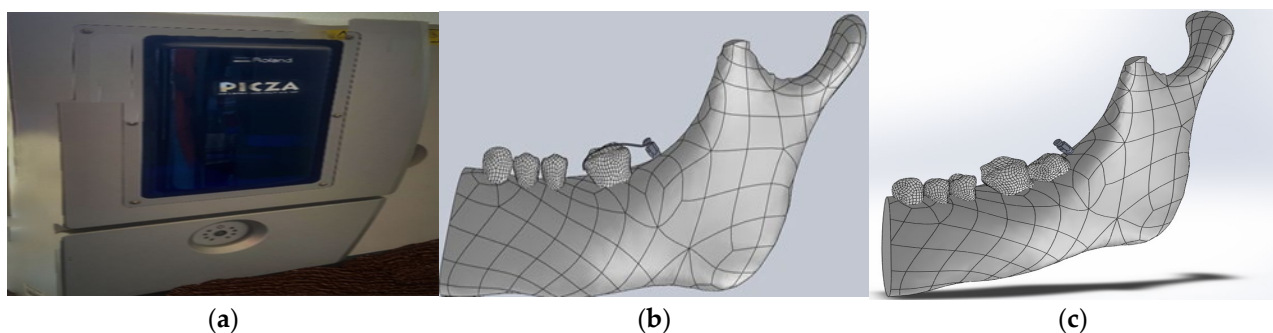


Figure 1. (a) Roland PICZA 3D laser scanner model LPX-1200, utilized for obtaining the models; (b) Model 1 and (c) Model 2 of the considered mandible.

The 3D scanned image was processed using reverse engineering techniques in order to obtain a solid geometric model compatible with the numerical analysis software.

The numerical analysis of the clinical situations described above was carried out on two cases, namely on two three-dimensional models, as follows:

Model no. 1 presents the mandibular segment, including the following teeth: the canine, the first premolar, the second premolar, and the permanent second molar, which is inclined at an angle of 30° (Figure 1b).

Model no. 2 includes all the elements of Model no. 1, and in addition, the wisdom tooth is present, being in the process of eruption (Figure 1c).

In both 3D models, a retromolar MI was positioned under the assumption that sufficient arch space was available to permit this approach. Molar uprighting mechanics in both models were simulated using a spring system, with one end attached to the retromolar MI and the other end connected to a button bonded to the mesial surface of the second permanent molar. This configuration allowed for a controlled application of forces in order to achieve tooth uprighting while minimizing undesirable effects on adjacent teeth and surrounding structures.

Subsequently, the models were transferred into the ANSYS 2019 R2 Academic software (Simtec, Thessaloniki, Greece) for performing numerical simulations. The numerical analysis of the cases was carried out by considering the linear elastic behavior of the tissues, described by their elastic properties, Young's modulus, and Poisson's ratio, as presented in Table 1. The content and consistency of the periodontal ligament vary significantly depending on the portion of the root, with values that range between 0.39 and 5.08 MPa [18]. Therefore, a higher fiber content corresponds to a greater resistance to orthodontic movements. In our study, the maximum value was applied, based on previously reported peak PDL values in the middle region of the root and during transverse movements.

Table 1. Young's modulus and Poisson's ratio for the different materials involved in FEA.

Material	Young's Modulus (MPa)	Poisson's Ratio
Cortical bone	1.37×10^4	0.30
Cancellous bone	7.90×10^3	0.30
Tooth	2.07×10^4	0.30
TMA	8.00×10^4	0.30
Miniscrew	1.05×10^5	0.33
PDL	50	0.49

Although the periodontal ligament exhibits nonlinear and viscoelastic behavior in vivo, the linear elastic approximation was adopted to allow for a reliable evaluation of comparative stress distribution and initial displacement patterns within the scope of this exploratory biomechanical study. Identical material properties were applied to both models, ensuring that the comparative outcomes reflect the investigated variable rather than differences in material modeling assumptions.

Since identical material properties were assigned to both models, the comparative analysis remains valid despite the simplified constitutive assumptions. Accordingly, a linear elastic and isotropic approximation was deemed appropriate for the objectives of this study.

For the simulations, teeth were isolated from the mandibular models: Model 1 utilized the permanent second molar, and Model 2 included both the second and third molars (Figure 2a).

FEM was implemented by discretizing the 3D geometric model into a finite number of predefined elements, each governed by specific interpolation (shape) functions. The mesh consisted of 3D, primarily tetrahedral or hexahedral in geometry, with nodes located at the element vertices and, when applicable, along their edges. Based on the principles of continuum mechanics, nodal displacements were computed for each element. The corresponding interpolation functions were subsequently utilized to estimate the displacement field at any point within the element. Thus, the geometric models corresponding to the two analysis cases were discretized into tetrahedral finite elements, with a total of 20,204 elements with 34,905 nodes for a molar (Figure 2b). An analysis of mesh convergence was performed during the development of the preliminary model to ensure numerical stability and reliable stress distribution. Refinements were concentrated in regions of high interest, including the second molar roots and periodontal ligament. Based on these preliminary checks, further mesh refinement did not result in significant changes in stress values or displacement patterns.

Boundary conditions were applied to the condylar region and to the inferior border of the hemimandible to prevent rigid-body motion. The same constraints were used in both models to ensure result comparability.

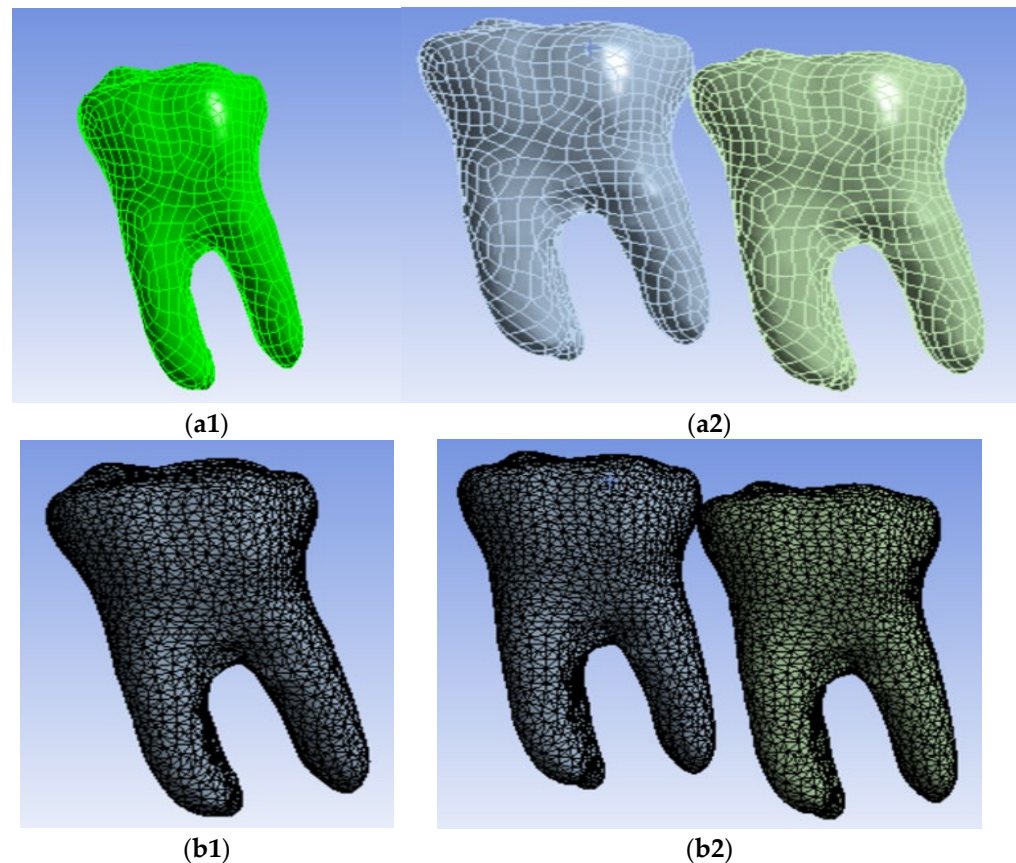


Figure 2. (a) Model and (b) discretization of the (1) one and (2) two molars considered models.

In this study, since the second and third molars (when present) were modeled with small separations, no frictional or sliding contact was applied. The stress was transmitted primarily through the PDL and the bone. The contact conditions in FEM were defined to reflect the physiological behavior of the mandibular system during a second molar uprighting. The bonded contact was assumed at the tooth–PDL and PDL–bone interfaces. This ensures continuity and prevents relative motion, allowing the PDL to act as the primary load-transferring and deformable structure, which is consistent with its biological role. This assumption leads to a realistic stress distribution, with forces transmitted through the PDL rather than across artificial interface gaps. This enables an accurate representation of stress concentrations in clinically relevant regions such as the cervical and apical areas of the root. The interaction between the orthodontic appliance and the tooth was simplified as bonded (or alternatively frictional), thus providing a stable and consistent force application. Although simplified, this does not significantly alter the overall biomechanical response.

Tooth displacements were obtained further on, as presented in the following. In both simulations, the force was applied to the permanent second molar in order to achieve an uprighting movement using the spring method, anchored on one side to a button cemented on the mesial surface of the second molar and on the other side to an MI placed in a vertical retromolar position. A force of 200 g (i.e., 2 N) was introduced, originating from the cemented button and directed toward the MI position, through which displacements along the three axes (X, Y, and Z) were observed, as well as the areas that are most prone to deformation and stress.

3. Results

The X-axis displacements are in the mesio-distal direction. Their positive values indicate the displacement of the roots toward the mesial, as suggested by the direction of the X-axis

arrow in Figure 3. These displacements occur as a reaction to the uprighting movement that was applied to the permanent second molar.

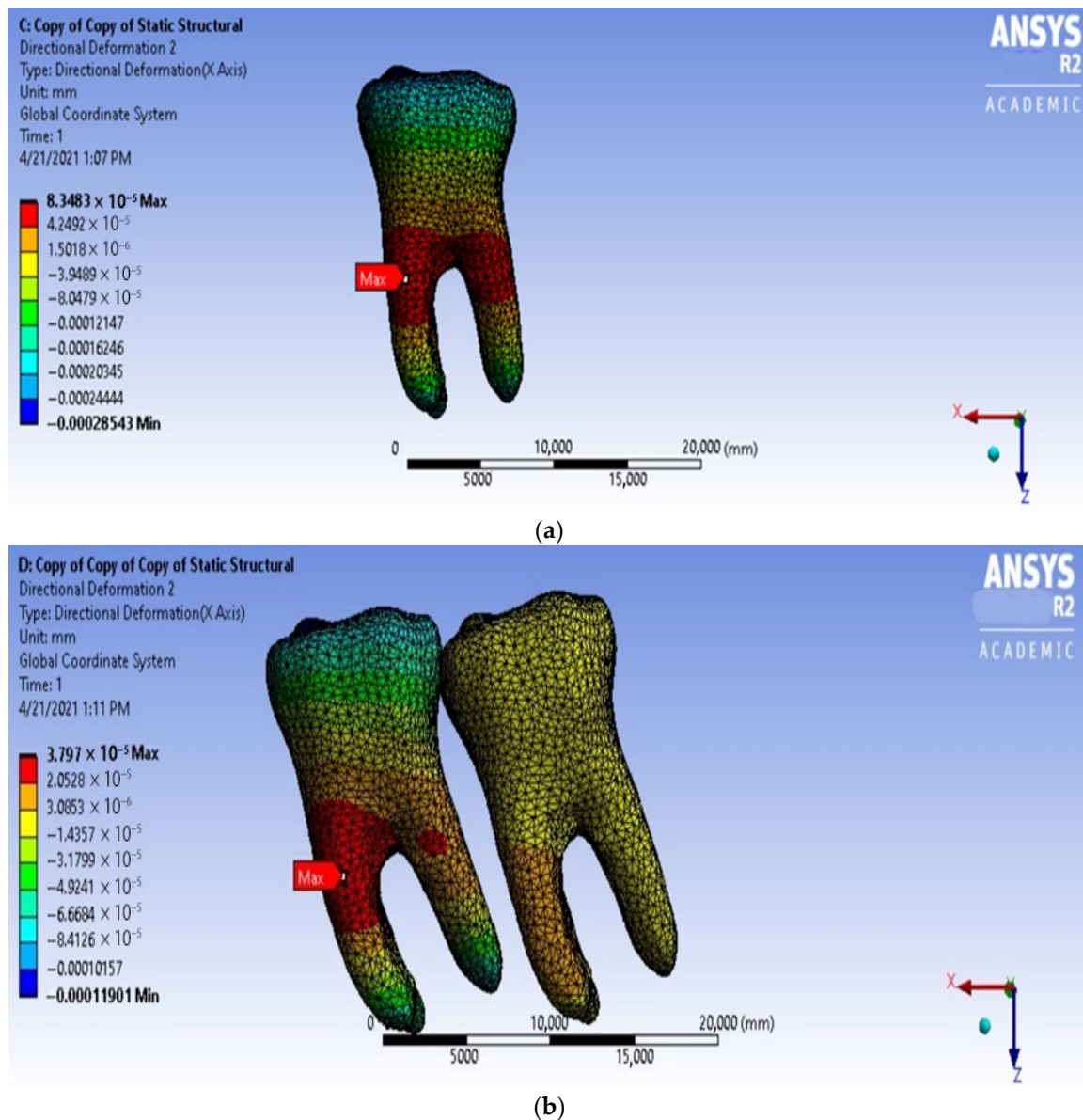


Figure 3. The X-axis displacement for (a) one molar (Model 1) and for (b) two molars (Model 2).

Regarding these X-axis displacements, when only the permanent second molar is present on the arch, deformation occurs at the level of the furcation and the middle third of both roots (Figure 3a). At this level, the highest stress is observed on the roots and alveolar bone. Also, the processes of bone apposition and resorption that are necessary for achieving dental movements take place.

In contrast, when the third molar is also present in the arch, the deformation occurs mainly at the level of the mesial root. The maximum stress was observed in the middle and coronal thirds of the mesial root (Figure 3b).

Therefore, when only the permanent second molar is present in the arch, the displacement stresses at the root level are distributed across both roots. However, when the third molar is present as well, the displacement stress is observed only at the mesial root.

The Y-axis displacements are in the bucco-lingual direction. When only the permanent second molar is present in the arch, these displacements (and consequently the processes

of bone apposition and resorption) occur at the apex of the distal root, where the largest displacement deformation can be observed (Figure 4a).

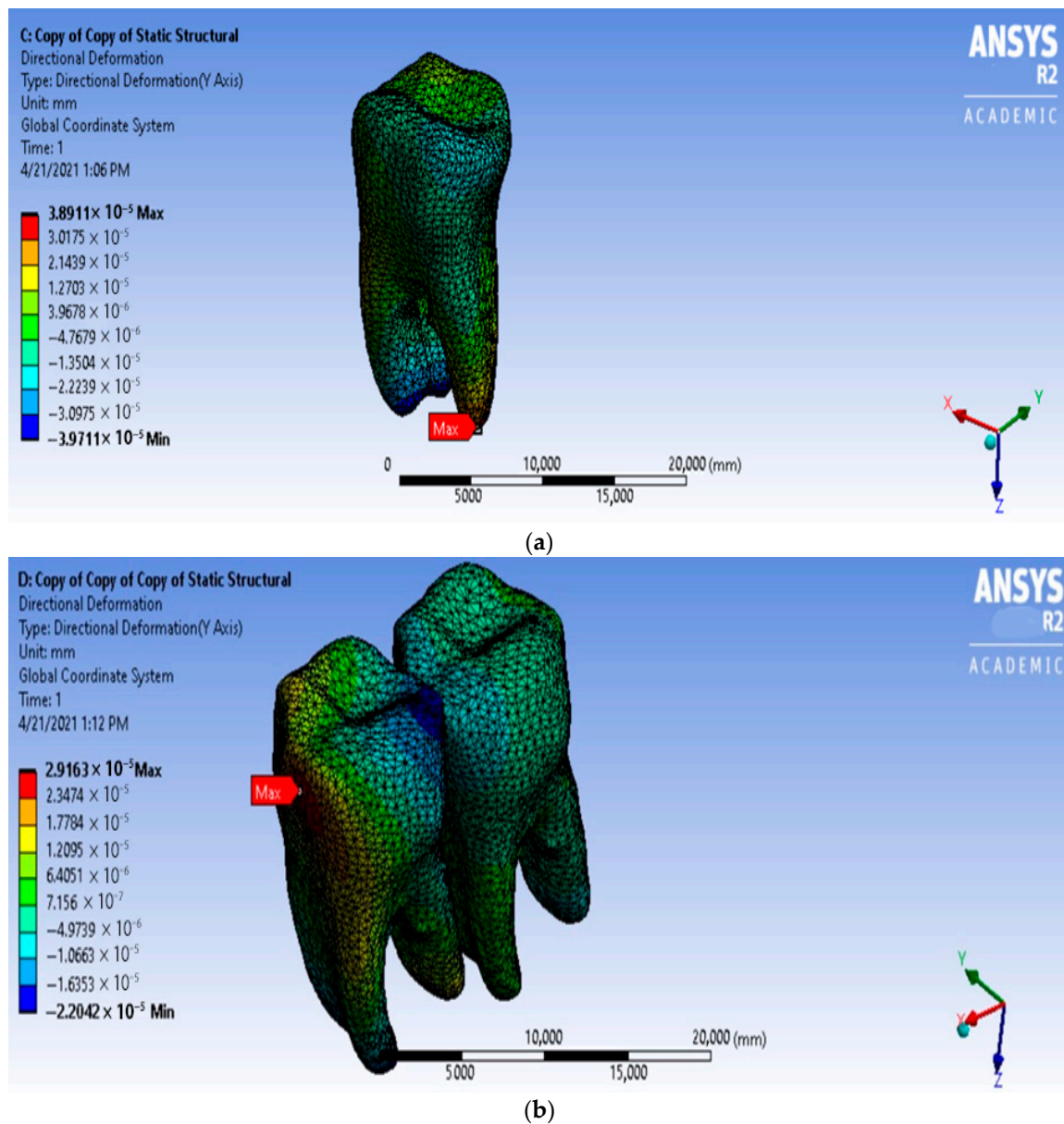


Figure 4. The Y-axis displacement for (a) one molar (Model 1) and for (b) two molars (Model 2).

In contrast, when the third molar is also present in the arch, the displacement occurs predominantly at the mesial surface of the crown of the second molar. The maximum stress is observed at the anchorage point, which is the button cemented on the mesial surface of the second molar (Figure 4b).

When only the permanent second molar is present, root-level stresses are concentrated primarily at the mesial root apex. In contrast, the presence of a third molar shifts the maximum stresses from the root level to the anchorage unit. Therefore, keeping the third molar in the arch increases the stress at the anchorage unit but decreases the stress on the second molar root, potentially lowering the risk of root resorption.

The Z-axis displacements are in the superior–inferior direction. Thus, the intrusion occurs at the level of the second molar on which the uprighting forces are applied. One can observe that in both simulations—Model 1 and Model 2—the intrusion occurs at the distal surface and at the distal root of the second molar (Figure 5). This is precisely the

goal of molar uprighting treatment: not only to correct the tooth's axial inclination but also to intrude the distal cusps, thereby eliminating potential premature contacts and occlusion interferences.

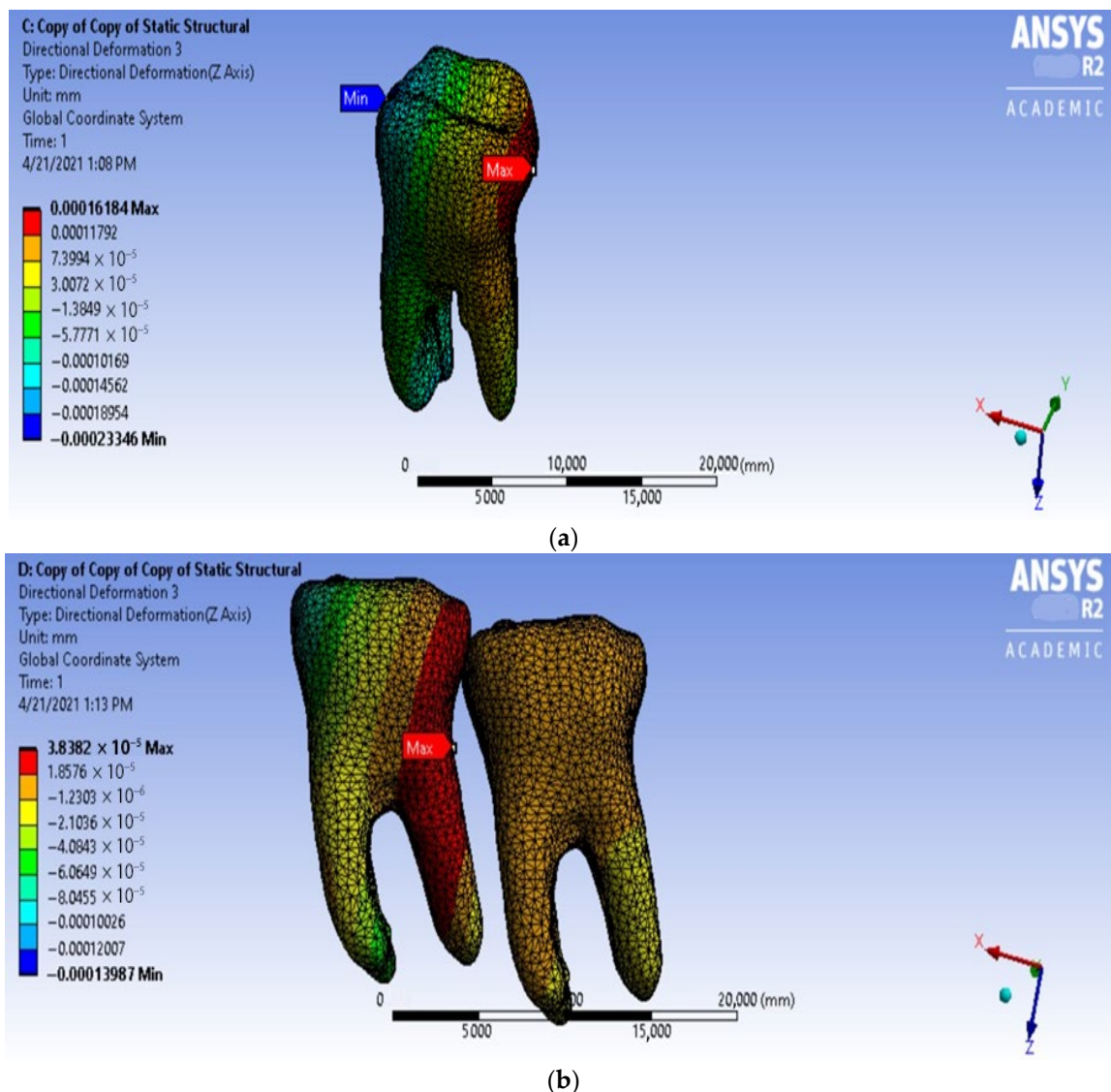


Figure 5. The Z-axis displacement for (a) one molar (Model 1) and for (b) two molars (Model 2).

However, when the third molar is present in the arch, the displacement value is significantly lower than when only the second molar is present. This suggests that the third molar acts as an obstacle to the intrusion of the second molar.

The *von Mises stress* obtained for the two considered models is presented in Figure 6. The results obtained from the simulations of the uprighting movement performed on Model 1 are as follows:

- The maximum displacement (i.e., the deformation during the movement) along the X-axis (in the mesio-distal direction) is observed at the level of the furcation and of the middle third of both roots, reaching a value of 8.348×10^{-5} mm.
- The maximum displacement along the Y-axis (i.e., in the bucco-lingual direction) is observed at the apex of the distal root of the second molar, with a value of 3.891×10^{-5} mm.

- The maximum displacement along the Z-axis (i.e., in the superior–inferior direction) is observed at the distal surface of the crown of the second molar, with a value of 1.618×10^{-4} mm.
- The von Mises stress values at the distal surfaces of the roots of the permanent second molar are 0.46955 MPa for the mesial root and 0.37051 MPa for the distal root.

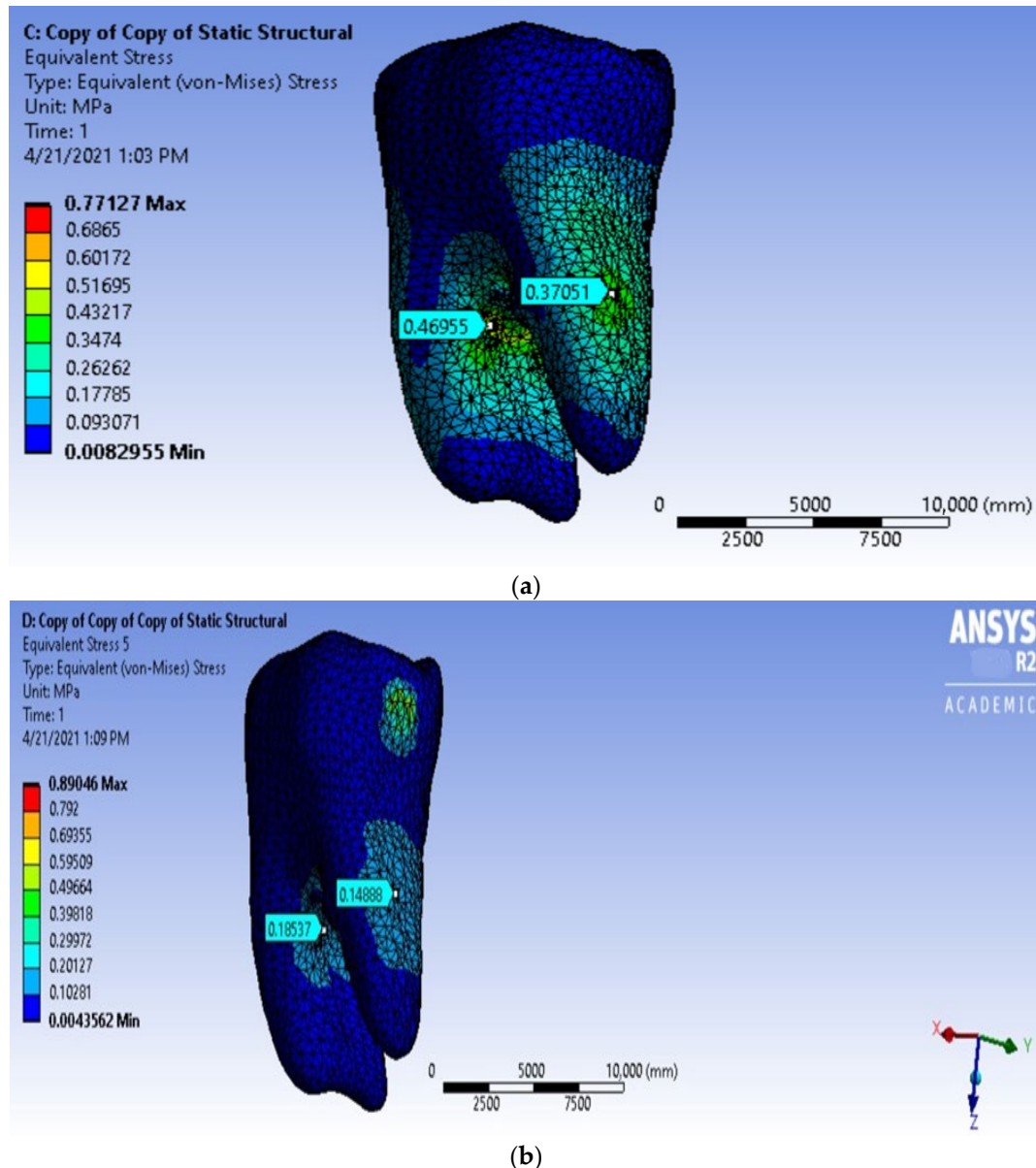


Figure 6. The von Mises stress obtained for (a) Model 1 and (b) Model 2.

The results obtained from the uprighting movement simulations performed on Model 2 are as follows (Figure 7):

- The maximum displacement along the X-axis is observed at the roots of the second molar, at the mesial root, reaching a value of 3.797×10^{-5} mm.
- The maximum displacement along the Y-axis is observed at the mesial surface of the crown of the second molar, where the anchorage button is cemented, and it has a value of 2.916×10^{-5} mm.
- The maximum displacement along the Z-axis is observed at the distal surface of the crown of the second molar, and it has a value of 3.8382×10^{-5} mm.

- The maximum von Mises stress values at the distal surfaces of the roots of the permanent second molar are 0.18537 MPa for the mesial root and 0.14888 MPa for the distal root.



Figure 7. Graphical representations of (a) the displacement values (where blue represents the X-axis, orange the Y-axis, and green represents the Z-axis displacements) and (b) the von Mises stress for the mesial root (in blue) and for the distal root (in orange).

It should be highlighted that all reported displacements represent instantaneous mechanical responses obtained under controlled loading conditions. Such displacements are typically very small in finite element simulations of orthodontic force application. Therefore, even displacement values in the order of 10^{-5} mm can be considered biologically relevant in the context of orthodontic tooth movement. Such magnitudes are more indicative of the initial elastic response of the tooth–periodontal ligament system immediately after force application rather than true biological tooth movement, which involves periodontal ligament remodeling and occurs over a longer time scale [19].

4. Discussion

The various uprighting techniques utilized in numerous studies can affect the targeted teeth in different ways, with variations that occur in stress and force distribution. Available therapeutic approaches encompass surgical uprighting and tooth transplantation, as well as a range of orthodontic mechanics that include tip-back cantilevers, spring-based systems, sectional archwires, and skeletal anchorage using miniplates or MIs, as well as the use of a simple tube [20].

In conventional uprighting methods, multiple teeth are usually involved for anchorage purposes, or even the entire dental arch in cases where the patient already wears a fixed orthodontic appliance [21]. Unintended movement of the anchorage teeth (including tipping, rotation, intrusion, or extrusion of adjacent teeth) can occur during orthodontic mechanics. To mitigate these side effects, segmented approaches such as T-loop springs have been recommended, while specialized uprighting devices like the Sander spring have been developed for this purpose. However, the use of these auxiliaries can be technically demanding for both the clinician and the patient. Also, their application may cause trauma to the buccal mucosa, particularly in cases with shallow vestibular depth [22].

The introduction of TADs has marked significant progress in orthodontic biomechanics. They allow for precise force application to designated dental units while minimizing undesired movement of neighboring teeth [23]. From this category, MIs are widely utilized. Made from pure Ti or Ti alloy (1 to 2 mm in diameter and 8 to 20 mm in length), these devices depend on mechanical retention and osseointegration to provide the necessary stability for effective orthodontic tooth movement.

Determining the site where an MI must be inserted for orthodontic treatment is an important step in the treatment plan. The insertion site must meet certain criteria to ensure primary stability: it should have adequate thickness, proportional to the length of the MI, as well as a width of at least 2.5 mm. Also, according to Holm et al., it is recommended to place the MIs in different regions depending on the patient's age. Possible mandibular sites include the symphysis or parasymphysis, the alveolar process (i.e., between tooth roots), and the retromolar area [24]. A 3D-designed and fabricated surgical guide integrating CBCT data with intraoral scans provides superior accuracy for MI placement [25]. The thickness of the soft tissues influences the failure rate as well, since a mucosa thicker than 3 mm displaces the point of force application too far from the MI's center of resistance. This generates a higher moment that negatively affects the stability of the MI [26].

The aim of this study was to evaluate, by using 3D FEA, the biomechanical response of the permanent second molar to MI-assisted uprighting in two contrasting scenarios characterized by the absence or presence of the third molar within the dental arch. The results revealed marked variation.

A distalizing force of 2 N was selected because it is often utilized in clinical settings for various orthodontic applications, including canine retraction, where forces of 1.5 N to 2.5 N are commonly recommended [27]. In clinical orthodontic practice, forces that exceed approximately 2 N may cause pain or undesirable tooth mobility. This makes this value a reasonable upper physiological limit for controlled tooth movement. Moreover, although uprighting mechanics commonly employ forces around 1.5 N, the use of 2 N in FEA enhances the identification of the center of rotation and improves the distinction between tension and compression areas within the PDL. This is particularly valid when considering the larger size and root surface of molars. Also, previous FEA studies have shown that a 2 N distalizing force combined with appropriate counter-tipping and counter-rotation torques produces a more favorable and evenly distributed minimum principal stress in the PDL. This supports the biomechanical validity of this magnitude in the displacement behavior, the stress localization, and the anchorage efficiency [28].

Thus, for Model 1, the maximum displacement (i.e., the deformation during movement) was recorded along the X-axis (in the mesio-distal direction) at the level of the furcation and the middle third of both roots. It reached 8.348×10^{-5} mm. Along the Y-axis (in the bucco-lingual direction), the highest displacement occurred at the apex of the distal root of the second molar, with a value of 3.891×10^{-5} mm. In contrast, the largest displacement along the Z-axis (in the superior-inferior direction) was observed at the distal surface of the crown of the second molar, with a value of 16.18×10^{-5} mm. Additionally, the von Mises stress values at the distal surfaces of the roots of the permanent second molar were 0.46955 MPa for the mesial root and 0.37051 MPa for the distal root.

The results obtained from the uprighting movement simulations performed on Model 2 indicated that the maximum displacement along the X-axis occurs at the mesial root of the second molar, reaching 3.797×10^{-5} mm. Along the Y-axis, the highest displacement is observed at the mesial surface of the crown, where the anchorage button is cemented, with a value of 2.916×10^{-5} mm. The maximum displacement along the Z-axis appears at the distal surface of the crown, reaching 3.838×10^{-5} mm. Furthermore, the maximum von Mises stress values at the distal surfaces of the roots of the permanent second molar are 0.18537 MPa for the mesial root and 0.14888 MPa for the distal root.

The present findings demonstrate that the biomechanical response of the permanent second molar to an orthodontic displacement along the X-axis is strongly influenced by the presence or absence of the third molar within the dental arch. Distinct patterns of deformation and stress distribution were observed, highlighting the role of adjacent teeth in modifying the force transmission and the periodontal load distribution during an orthodontic tooth movement. Therefore, the study hypothesis has been confirmed.

When the permanent second molar is the sole posterior tooth present, deformation is distributed across the furcation region and the middle thirds of both roots. These areas coincide with zones where elevated stress values are observed not only within the dental roots but also in the surrounding alveolar bone. Such stress localization is consistent with classical biomechanical principles of multi-rooted teeth, in which forces applied along the horizontal axis are dissipated through both roots and concentrated at the furcation area. This distribution promotes simultaneous bone apposition and resorption in multiple regions of the periodontal ligament and alveolar bone. This facilitates bodily tooth movement while maintaining relative mechanical balance between the roots.

In contrast, the presence of the third molar significantly alters this stress distribution pattern. Under the X-axis displacement, the deformation becomes predominantly localized to the mesial root of the second molar, with the maximum stress concentrated in its middle and coronal thirds. This shift suggests that the third molar acts as a posterior constraint: it limits the distal displacement of the second molar and redirects applied forces anteriorly. As a result, the mesial root becomes the primary structure responsible for absorbing and transmitting orthodontic loads, while the distal root experiences a comparatively reduced stress.

These findings are particularly relevant from a biological perspective, as regions of elevated stress correspond to areas of active bone remodeling. In the absence of the third molar, remodeling processes are distributed across a broader periodontal surface. This potentially allows for more uniform tooth movement and reduces the risk of localized overload. Conversely, stress concentration on the mesial root in the presence of the third molar may intensify the remodeling activity in a more confined region. This could influence the rate of tooth movement and increase susceptibility to adverse effects such as root resorption or localized periodontal disease.

The observed differences in stress localization have important implications for orthodontic treatment planning. The presence of a third molar reduces the efficiency of force

distribution during second molar movement. This requires adjustments in force magnitude, direction, or anchorage strategy. These results support the rationale for careful evaluation of the third molar status prior to orthodontic mechanics that involve posterior tooth movement. This is particularly important when a bodily translation of the second molar is desired. In some cases, third molar extraction may contribute to a more favorable biomechanical environment by restoring a more symmetrical stress distribution across both roots of the second molar. Overall, the variation in deformation and stress patterns observed in this study underscores the complex interplay between dental anatomy and orthodontic force systems. Thus, the third molar significantly modifies the mechanical behavior of the second molar under an X-axis displacement. Therefore, it influences both the distribution of stresses and the biological processes that govern the movement of the tooth. These findings contribute to a more nuanced understanding of posterior orthodontic biomechanics, which may inform more predictable and biologically sound treatment approaches.

The bucco-lingual displacement of the permanent second molar along the Y-axis is markedly influenced by the presence of the third molar. The result is fundamentally different biomechanical behavior within the tooth–periodontal complex. These findings underscore the critical role of posterior anatomical constraints in modulating force transmission, displacement patterns, and stress localization during orthodontic tooth movement.

When only the permanent second molar is present in the dental arch, the displacement is predominantly concentrated at the apex of the distal root. This pattern is indicative of a tipping-type movement, in which the applied bucco-lingual forces are transmitted along the root structure toward the apical region. The concentration of the deformation at the distal root apex corresponds to localized regions of bone apposition and resorption. It reflects an apically driven biological response. Such behavior is consistent with established biomechanical models, in which an asymmetric force application to a multi-rooted tooth results in rotational moments and apical stress accumulation, particularly in the absence of posterior support. This apical concentration of stress may facilitate tooth movement but also raises concerns regarding increased susceptibility to root resorption in areas subjected to prolonged or excessive loading. In contrast, the presence of the third molar produces a pronounced shift in the displacement pattern, with deformation predominantly localized at the mesial surface of the second molar crown. Under these conditions, the maximum stress is concentrated at the anchorage site, specifically at the button cemented on the mesial crown surface. This finding suggests a substantial alteration in the force transmission pathway. Thus, the applied forces are largely absorbed at the site of application rather than being propagated along the root system. The third molar appears to act as a stabilizing posterior boundary, restricting root displacement and limiting apical stress concentration. Consequently, deformation becomes more crown-dominated, reflecting a more constrained biomechanical environment.

From a biological standpoint, this altered stress distribution implies a reduction in apical bone remodeling activity when the third molar is present. This potentially lowers the risk of apical root resorption. However, the increased stress concentration at the anchorage site may introduce new challenges, including heightened stress at the enamel–adhesive interface and on the surrounding periodontal tissues. This localized loading may compromise anchorage stability, increase the likelihood of attachment failure, or promote unwanted crown tipping if the magnitude and direction of the force are not carefully controlled.

Clinically, these findings have important implications for an orthodontic treatment plan that involves the bucco-lingual movement of the second molar. In cases where the third molar is absent, clinicians should anticipate a greater apical involvement and tipping tendencies. This requires a careful force calibration to minimize adverse biological effects.

Conversely, when the third molar is present, the restricted root movement and the crown-focused deformation may require modifications in appliance design, anchorage strategy, or force application in order to achieve the desired tooth movement efficiently and safely. Therefore, the third molar status may be essential in optimizing biomechanical outcomes and in minimizing treatment-related risks.

Overall, the observed differences in Y-axis displacement patterns highlight the significant influence of the third molar presence on the biomechanical response of the permanent second molar. By altering both stress localization and displacement behavior, the third molar plays a decisive role in shaping the efficiency, predictability, and biological impact of bucco-lingual orthodontic forces.

In both simulations (i.e., Model 1 and 2), regarding the displacement along the Z-axis, intrusion is primarily observed at the distal surface and along the distal root of the second molar. This displacement pattern aligns with the fundamental objectives of molar uprighting therapy, which aims not only to correct axial inclination but also to achieve intrusion of the distal cusps. Therefore, it reduces the risk of premature contacts and occlusal interferences. Notably, when the third molar is present in the arch (i.e., in Model 2), the magnitude of the intrusion is markedly reduced compared to Model 1, which includes only the second molar. This clearly indicates that the third molar may mechanically limit or impede the intrusive movement of the second molar. Thus, the analysis of Z-axis displacement provides important insight into the vertical biomechanical effects associated with molar uprighting mechanics. In both models, intrusion is consistently observed at the distal surface and distal root of the second molar. This aligns with the primary objectives of molar uprighting treatment.

Despite this shared displacement pattern, the magnitude of intrusion differs markedly between the two models. In the absence of the third molar scenario, an increased vertical displacement of the second molar is observed. This shows a more effective transmission of intrusive forces to the distal region of the tooth. This finding supports the hypothesis that the third molar acts as a physical and biomechanical obstacle: it absorbs or redistributes part of the applied forces. Therefore, it diminishes the intrusive effect on the second molar.

These results underscore the influence of third molars on the effectiveness of molar uprighting mechanics. Reduced intrusion in the presence of a third molar may compromise the desired correction of occlusal discrepancies. This potentially prolongs treatment or necessitates adjustments in force magnitude or mechanics. Therefore, a careful evaluation of the third molar presence and position is essential during treatment planning, as its removal or management may be required for optimizing vertical tooth movement and for achieving predictable uprighting outcomes.

Although previous clinical studies have reported no significant effect of third molar removal on second molar uprighting [22,23], our FEA indicates that the presence of the third molar does, in fact, influence displacement patterns and stress distribution. Specifically, the removal of the third molar allows for a more uniform stress distribution across the roots. In contrast, its presence concentrates the stress on the mesial root and on the crown region. This discrepancy suggests that subtle biomechanical differences, which may not be detectable in clinical outcomes, could nonetheless affect anchorage planning and the efficiency of uprighting procedures.

Caminiti reported third molar removal in 86.9% of 255 second molar uprighting cases, primarily to avoid interference with the desired result or based on orthodontic referral preferences. Although contemporary guidelines support individualized third molar assessment, the author suggested that retention may restrict access and limit the controlled tooth movement. Direct comparisons between uprighting performed with and without the third molar preservation remain limited, largely due to the routine extraction

of third molars during orthodontic treatment [29]. The present study addresses this gap by providing a controlled FEA of second molar uprighting with and without the third molar's retention. The results demonstrate distinct differences in displacement and stress distribution and suggest a stabilizing biomechanical effect of the presence of the third molar during uprighting.

The discrepancy between clinical observations and the present FEA findings may be explained by fundamental methodological differences. Clinical studies have evaluated final treatment outcomes influenced by multiple biological, mechanical, and patient-specific factors, whereas FEA isolates the mechanical behavior under controlled conditions. Therefore, the biomechanical differences observed in the present study may represent initial force transmission patterns rather than clinically observable treatment endpoints. The apparent divergence between clinical reports and the results of the computational simulation underscores the complementary rather than the contradictory roles of clinical research and biomechanical modeling.

These findings offer objective evidence to support more selective, evidence-based third molar management. Also, they highlight the necessity for further clinical and biomechanical investigations to refine extraction protocols. One must point out that the observations made in this study are derived from static simulations without biological remodeling. Therefore, clinical advantages of third molar retention are clearly distinguished from biomechanical findings (e.g., stress redistribution).

The biomechanical accuracy of forces applied to upright a tilted inferior second molar is not solely dependent on the magnitude of the applied orthodontic force but is markedly influenced by the properties and placement of the mini-implant, as well as by the presence or absence of the third molar [30]. FEA demonstrated that varying MI placement created distinct force vectors and moment-to-force ratios at the second molar. This impacts the predictability and efficiency of uprighting mechanics. The presence of the third molar altered the stress distribution and displacement patterns in the second molar's periodontal ligament and alveolar bone. This suggests increased resistance and modified force transmission compared to models without a third molar.

The present study confirms that FEA is a vital computational tool in biomechanics, as it models complex structures and allows for an analysis of their mechanical properties. This allows for the visualization of stress distribution and displacement patterns, as well as for the quantification of force location, intensity, and direction within anatomical structures. This provides detailed insights into how biological systems and medical implants react to various physical conditions [31]. Furthermore, as a non-destructive numerical method, FEA allows for repeatable simulations without altering material properties. The process involves discretizing a geometry into interconnected finite elements via nodes, where the element density and configuration determine the accuracy of the results. By defining specific material properties, loading parameters, and boundary conditions within the engineering software, researchers can precisely simulate and quantify structural stresses and deformations [32].

In the last decade, FEA has been utilized in numerous orthodontic studies, leading to conclusions that would have been difficult or even impossible to obtain in vitro. This has opened new horizons in this branch of dental medicine.

Barone et al. developed a digital workflow integrating imaging, scanning, and modeling to optimize orthodontic aligner design using FEA. This methodology simulates biomechanical interactions between patient-specific anatomical models and aligner geometries in order to evaluate tooth movement. Their study demonstrated that specific auxiliary element configurations significantly influence incisor tipping efficiency [33].

Saga A.Y. used a CBCT-reconstructed FEA model to evaluate periodontal ligament stress during simulated maxillary incisor intrusion with varied force application points. Their study concluded that, regardless of the force's application point, stress consistently concentrates in the apical region. Thus, clinical application of FEA supports tailored treatment planning, lessens iatrogenic complications, and fosters an evidence-based, collaborative approach. However, FEA serves as a supplementary tool and must not supersede comprehensive clinical assessment and professional judgment [34].

Benaissa et al. determined that MI stress peaks at the neck–cortical bone interface and diminishes toward the apex. In the alveolar bone, the maximum stress concentrates near the cervical region. This is aligned with patterns of clinical bone loss. That study further emphasized that periodontal ligaments serve as essential mediators in force transmission [35].

The issue of retaining the third molar in the arch is a topic that has not been sufficiently discussed, and there is limited data in the literature, especially referring to studies that utilize FEM. Hence, the present study may support the community in adopting the most appropriate clinical strategy.

Within the limitations of the present model, the study has demonstrated that the presence of the third molar influences the stress distribution and displacement patterns during a simulated second molar uprighting with a retromolar MI. Specifically, the presence of the third molar is associated not only with a lower overall deformation of the second molar but also with a reduced displacement magnitude. These findings suggest that adjacent posterior teeth may modify the biomechanical environment during orthodontic molar uprighting. However, it must be highlighted that these results represent comparative biomechanical tendencies rather than precise physiological predictions. Thus, despite the biomechanical insights provided by this FEA, several limitations should be considered. Specifically, the materials included in the model were assumed to be linearly elastic, homogeneous, and isotropic, whereas biological tissues, particularly the periodontal ligament, exhibit nonlinear and viscoelastic behavior that may influence the stress distribution during orthodontic loading. In addition, the simulations were performed under static loading conditions. This represents only the initial mechanical response to force application. Orthodontic tooth movement is a dynamic process that involves time-dependent force decay and tissue adaptation. Furthermore, the model does not incorporate biological remodeling of the periodontal ligament and alveolar bone, which plays a fundamental role in clinical tooth movement. Therefore, the obtained results should be interpreted as theoretical biomechanical tendencies rather than direct predictors of clinical outcomes. As clinical decisions regarding third molar management should not be based solely on computational simulations, further clinical and experimental studies are required to clarify the clinical significance of these biomechanical observations.

5. Conclusions

Clinical measurements regarding the stress distribution on teeth and surrounding tissues during orthodontic molar uprighting movements are impossible to perform. In contrast, FEA is a method that allows for simulating reality, although it does not represent the exact real situation. Therefore, some biomechanical factors in the present study may differ from those in *in vivo* studies. Additionally, the individual characteristics and biological responses of anatomical elements, teeth, and surrounding tissues to different forces must be considered.

This work demonstrates that the presence of the third molar significantly modulates the biomechanics of second molar uprighting via retromolar MIs. Also, it is associated with altered force transmission and a reduced magnitude of tooth displacement in the considered computational models. When the third molar is present, the second molar

exhibits a reduced tendency for deformation during distalization. However, this also leads to a slower displacement compared to the situation in which the third molar is no longer present.

It must be highlighted that the recommendation of third molar retention is a potential biomechanical hypothesis that must undergo clinical validation. Thus, the broader clinical implications of the present theoretical simulation should be interpreted with caution. This FEA provides biomechanical insights into stress distribution and displacement patterns during simulated second molar uprighting but does not support direct clinical decision-making. An orthodontic treatment plan depends on multiple patient-specific biological and anatomical factors that are beyond the scope of this model. Therefore, the present findings should be viewed as theoretical biomechanical tendencies that require confirmation through clinical, experimental, and longitudinal studies before translation into clinical practice.

While empirical simulations have limitations due to infinite clinical variations, the primary recommendation that can be extracted from the study is to retain the third molar when feasible. This approach offers benefits such as reduced stress on the second molar and the potential for both molars to be aligned. This can close the space and possibly eliminate the need for prosthetic restorations, as natural teeth are superior to any artificial replacement.

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Abbreviations

The following abbreviations are used in this manuscript:

3D	Three-dimensional
FEM	Finite element analysis
CT	Computed tomography
TAD	Temporary anchorage device
RME	Rapid maxillary expansion
PDL	Periodontal ligament

FEM	Finite element method
MI	Mini-implant
OTM	Orthodontic tooth movement
MRI	Magnetic resonance imaging

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