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Finite Element Analysis of Two Panthera MADs: Implications for Biomechanical Stress on Teeth and Periodontal Ligaments



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ABSTRACT

Aim: This study compares the biomechanical effects of 2 mandibular advancement devices (MADs), Panthera D-SAD Classic (ID1) and Panthera D-SAD X3 (ID2), on dental and periodontal structures using finite element analysis (FEA). The objective is to assess differences in stress distribution, displacement patterns, and mechanical interactions with teeth and periodontal ligaments (PDLs). This study provides insights to improve MAD design and clinical outcomes in the treatment of obstructive sleep apnea syndrome (OSAS).

Methods: 3D anatomical models, including teeth and PDLs, were reconstructed from cone beam computed tomography (CBCT) scans of an OSA patient. Two Panthera MADs were digitally modeled, processed in CAD software, and integrated with the anatomical models. Finite element simulations were performed in ANSYS, applying uniform boundary conditions, material properties (including a hyperelastic Ogden model for PDLs), and simulated mandibular advancement forces. Stress and displacement distributions in dental structures and MADs were evaluated and compared between the 2 devices.

Results: The Panthera D-SAD X3 (ID2) showed a more uniform distribution of stresses across dental and periodontal structures, attributable to a 23% increase in device volume, despite potential patient discomfort. In contrast, the Panthera D-SAD Classic (ID1) generated higher stress peaks, particularly in the mandibular molars and PDLs (eg, up to 135.85 kPa on tooth 37 and 1.44 kPa on the corresponding PDL). ID2 exhibited lower peak stresses but a more pronounced load on premolars and canines. Overall, the X3 device led to a more balanced mechanical response across the dentition.

Conclusion: The MAD design influences biomechanical interactions with dental and periodontal tissues. The Panthera D-SAD X3 demonstrated superior stress uniformity and mechanical efficiency performance, suggesting its potential to reduce localized overloading. Future research should focus on simulating the effects across a larger number of patients by considering multiple anatomical geometries. Moreover, long-term effects must be assessed using an appropriate FEM model.

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Introduction

Obstructive sleep apnea (OSA) is a sleep disorder characterized by repetitive episodes of partial or complete upper airway collapse during sleep, leading to airflow cessation

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(apnea) or reduction (hypopnea).¹ These episodes result in intermittent hypoxia and sleep fragmentation, which contribute to excessive daytime sleepiness, impaired cognitive function, and an increased risk of cardiovascular diseases, stroke, and hypertension.² 425 million adults worldwide suffer from moderate-to-severe OSA, while 936 million individuals aged 30–69 have mild to severe OSA.³ The estimated prevalence of OSA varies across populations. Still, it has been reported that a male-to-female ratio of 2:1 to 3:1 exists in the general population.⁴

The standard treatment for OSA is continuous positive airway pressure (CPAP), which maintains airway patency by providing a constant pneumatic splint. However, CPAP therapy suffers from low patient compliance due to discomfort, upper airway dryness, and difficulties with long-term use.⁵ As an alternative, mandibular advancement devices (MADs) have emerged as an effective, non-invasive treatment option for mild to moderate OSA and for patients with severe OSA who are intolerant to CPAP.⁶ These oral appliances advance and stabilize the mandible forward, enlarging the pharyngeal airway space and reducing tissue collapse.

Despite their efficacy in reducing respiratory events and improving symptoms, MADs are associated with potential side effects, including temporomandibular joint (TMJ) discomfort, muscular tenderness, dental soreness, occlusal changes, and long-term modifications in overjet and overbite.⁷ Additionally, their biomechanical effects on teeth and periodontal ligaments (PDLs) remain an area of ongoing investigation. Understanding these effects is crucial to optimizing MAD designs and improving their long-term tolerability and effectiveness.

Finite Element Analysis (FEA) has proven to be a powerful tool in investigating biomechanical behavior in dentistry and orthodontics, providing insights into stress distribution, force application, and structural integrity of dental devices. Previous studies have successfully applied FEA to analyse the impact of oral appliances on TMJ mechanics and to evaluate the stress distribution in the teeth and PDLs during orthodontic treatment.⁸ Beyond biomechanical modeling, other quantitative computational approaches, such as fractal analysis of radiographic images, have been investigated to characterize trabecular bone architecture and explore its potential association with clinical outcomes, including early dental implant loss.⁹

The effects of MADs on dental structures, particularly stress distribution within the periodontal ligament, require further investigation. FEA enables simulation of forces exerted by MADs, providing a non-invasive means of understanding their biomechanical impact on teeth and supporting tissues. Recent studies have highlighted the importance of considering different MAD designs and their mechanical behavior to enhance treatment efficacy and minimize side effects. Additionally, variations in material properties, adjustment mechanisms, and force transmission across the dental arches influence treatment outcomes and long-term patient comfort. Predicting these biomechanical interactions through numerical simulations can help refine MADs, thereby improving therapeutic strategies for OSA patients. However, research on the biomechanical effects of different MAD designs on dental structures has highlighted distinct stress

and strain distributions within the PDL.^{10–12} This research offers valuable insights into the structural behavior of MADs under physiological loads using advanced numerical simulations, enabling clinicians to make informed decisions for patient-specific treatment planning. Furthermore, understanding the mechanical response of these devices may help develop next-generation MADs with improved comfort, durability, and therapeutic effectiveness.

This study aims to evaluate the biomechanical effects of 2 Panthera MADs, the Panthera D-SAD Classic and the Panthera X3, using FEA. The Panthera D-SAD Classic is a patient-specific, digitally printed nylon device that allows titration in 0.5 mm increments using interchangeable rods, with an adjustment range of 16–34 mm. Once titration is complete, no further modifications are required. The Panthera X3, in contrast, is a dorsal fin appliance known for its strength, lightweight properties, and adaptability. Both devices are 3D-printed (via additive manufacturing) and made from the same PA-2200 (polyamide 12). Given the importance of optimizing MAD designs to enhance treatment efficacy while minimizing adverse effects, this study will compare the stress distributions, force application, and biomechanical impacts of these 2 devices on teeth and PDLs.

Materials and methods

The methodology is based on the study presented in a prior publication.¹³ The first step is to acquire physical devices and anatomical parts used by OSA patients. Then it proceeds to a geometric reconstruction of the model using CAD software. Subsequently, the 3D model is implemented in Ansys, a structural analysis software, where the materials and boundary conditions are defined, and the mesh is created. Once the finite element model has been simulated, the biomechanical effects of MADs on the PDLs and teeth can be evaluated. The current work presents a simulation to compare the devices in [Table 1](#).

Model acquisition and reconstruction

The 3D reconstruction of all the anatomic structures has been achieved starting from the Cone Beam Computed Tomography (CBCT) images of a female patient affected by OSA syndrome through a 3D reconstruction software already performed in a recent work (Voxel size 75 micron, FOV 11 × 13 cm. NewTom Giano, Cefla, Italy).¹³ The reconstructed anatomy consists of a complete set of teeth (except the unerupted wisdom teeth). Since the periodontal ligament (PDL) is a soft tissue with a thickness comparable to or smaller than the CT slice resolution, it cannot be reliably identified in CT images. Although the PDL exhibits a non-uniform, hourglass-shaped morphology (about 0.15–0.38 mm), it was reconstructed in a 3D NURBS (Non-Uniform Rational B-Spline) surface modeling environment by applying a uniform offset to the tooth-root surface. A homogeneous thickness of 0.3 mm was assumed, as it represents a widely reported average value and a common simplification in finite element modeling to reduce geometric complexity and ensure numerical stability, consistent with previous studies.^{14–16}

Table 1 – Features of simulated devices.

ID	MAD	Upper Plateau	Lower Plateau	Upper Bands	Lower Bands
1	Panthera D-SAD Classic				
		Mesial of the canine to the distal of the 2nd premolar.	Distal 1st molar to distal 1st molar.	The lingual band has an incisor overlap, with no contact.	The lingual band has an incisor overlap, with no contact.
2	Panthera D-SAD X3				
					

Reconstruction of the Panthera's MADs

In this work, 2 different MADs were evaluated (Panthera Dental, Quebec, CN). Firstly, the STL model was imported into the 3D modeling software Rhinoceros (v.7.0, McNeel & Associates, Seattle, WA) to separate the upper and lower splint meshes for each MAD, thereby enabling reconstruction of one splint at a time. Then, each splint was imported into Geomagic Design X, 3D scanning and data processing software. Several steps were performed with this software. Initially, the Automatic Correction command was used to address various mesh defects. The Automatic Surface command automatically creates networks of curves on the mesh. It adapts surface patches to the network to maintain the accuracy of the underlying mesh. In this step, the surface was reconstructed with maximum precision and a tolerance of 0.05 mm. A closed geometry of the splint is obtained at the end of these passages. The anatomical model is also imported into Geomagic Design X to perform a Boolean subtraction between the MAD and the teeth, removing any interpenetration. At this point, the reconstructed geometry of the MAD can be exported in STP format to the FE software Ansys v. 2023 R1 (Ansys, Inc., Canonsburg, PA), along with the anatomical model comprising teeth and PDLs.

Finite element model processing

Finite element model processing involves preparing and analysing a numerical representation of a physical structure to

predict its mechanical behaviour under different conditions. The model setup includes defining the material properties used in constructing the components, specifying the contact interactions between bodies, and applying external loads and boundary conditions to reproduce a realistic operating environment. Once these parameters are introduced, the model is discretized into finite elements and solved computationally to evaluate characteristics such as stress distribution, deformation, and overall structural performance.

Properties of materials

Each material was assigned as described in the prior research,¹³ in which the same assumptions were made. The material properties of the components representing MAD and teeth were assumed to be homogeneous, isotropic, and linearly elastic, as listed in Table 2. The PA2200 material was used for the MADs: a biocompatible polyamide ($E = 1700$ MPa, $\nu = 0.4$).⁷⁻¹⁸ For PDLs, a hyperelastic model based on a strain-dependent energy function was used, as expressed in.¹⁸

$$W = \sum_{i=1}^N \frac{2\mu_i}{\alpha_i^2} (\bar{\lambda}_1^{\alpha_i} + \bar{\lambda}_2^{\alpha_i} + \bar{\lambda}_3^{\alpha_i} - 3) + \sum_{i=1}^N \frac{1}{D_i} (J - 1)^{2i}$$

W is the strain energy function, $\lambda_1, \lambda_2, \lambda_3$ are the principal stretches of the left Cauchy-Green tensor, and J is the determinant of the elastic deformation gradient. The parameter N is the order of the polynomial. At the same time, the constant μ_i is related to the initial (around the undeformed configuration) shear modulus of the material. The coefficients D_i

Table 2 – Mechanical properties of the model's components.

Material	Model	Young's modulus [MPa]	Poisson's ratio	Coefficient values
Tooth	Linear elastic	18300 ^{20,21}	0.31	
MAD (PA2200)	Linear elastic	1700	0.4 ¹⁷	
PDL	Hyperelastic Ogden 2nd order	-	-	$\alpha_1 = -3.4761$ $\alpha_2 = 18.679$ $\mu_1 = -0.034004$ MPa $\mu_2 = 0.00088691$ MPa $d_1 = 0$ $d_2 = 0$

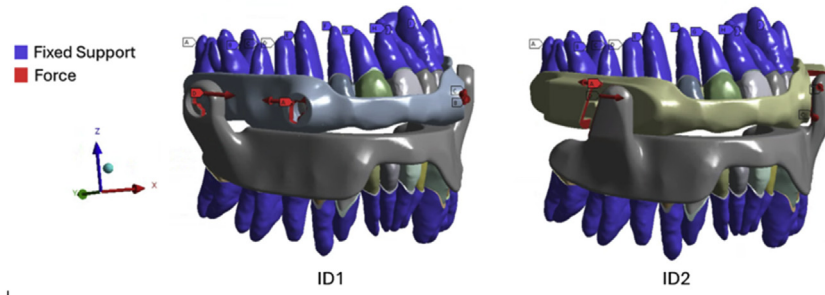


Fig. 1 – Boundary conditions representation across the 2 devices. In blue are the regions scoped to fixed support, while in red are the load vectors.

correspond to the material's initial bulk modulus. Since the PDLs exhibit almost incompressible behavior, they were set to 0. Lastly, the parameters α_i govern the nonlinearity of the stress-strain relationship and enable the model to fit experimental stress-strain curves with strong nonlinearity.

For the best fit to uniaxial test data, a 2nd-order Ogden model was chosen to represent PDLs' behaviour.^{13,19}

The simulation model evaluates the interactions between Panthera's MADs and anatomical structures. To this end, defining the stress state and deformation during OSA treatment that cause mandibular protrusion for millimetric adjustment of the MAD (measured on the occlusal plane) is crucial.

In this work, all boundary conditions are held constant across simulations, since results can only be compared in this way. Thus, in all simulations, the constraint applied between teeth and PDLs was a shared topology to achieve a conformal mesh at the interface between bodies.

Bonded contact was used to connect the MAD to the teeth; no sliding or separation between the faces or edges is allowed. This condition was applied to all teeth, except the upper and lower incisors, which were excluded because there is a 0.2 mm gap between them and the lingual bands of the devices. A frictionless contact was used between them and the MAD's splint to handle the incisors' gap, permitting separation and free sliding with a zero friction coefficient. By computing the cloud-to-mesh distance between each MAD splint and the dental arches, it was possible to define the radius of a pinball region useful for contact detection. It was set to 0.15 mm to prevent actual contact pressure, but the elements are close enough that contact may occur if the loads or displacements change slightly (near-open status). A pinball radius of 0.02 mm was chosen for the bonded contact between the 2 splints and teeth to ensure retention in the proper region.

A no-separation contact was established between the 2 splints, allowing relative sliding without separation.

A fixed support constraint was applied to the outer surface of the maxillary and mandibular PDL to prevent rigid-body motion of the model. Although this simplification may influence stress values in the immediate vicinity of the boundary, according to the Saint-Venant Principle, its effects are expected to be negligible within the regions of interest.^{22,23} Consistently, a previous study¹⁵ reported minimal differences in PDL von Mises stress between models with and without explicit bone representation. However, a dedicated

comparison performed at the dental level (Figure S1) showed that, while average stress values were only modestly affected (6% mean variation), maximum von Mises stress on the dental elements showed a more pronounced variation (27% on average) between models with and without bone, indicating that bone structures contribute more substantially to the localization and magnitude of peak stress than to average stress distribution. This simplification was therefore applied consistently across both MAD configurations under comparison, so that, although the absolute magnitude of peak stress values should be interpreted with caution, the relative comparison between the 2 devices, the primary aim of the present study, is not expected to be substantially affected.

The components of the MADs related to the connection mechanism have been neglected during the simulation. This assumption aligns with previous studies,^{23,24} thus permitting a comparison of results. The coupling mechanism was replaced by 2 equal and opposite forces acting along the contact element between the splints. This schema allowed the study of the mutual exchange of forces, representing the TMJ's restorative forces. According to the article,²⁵ a total load of 11.18 N was applied to each splint. Half of the total load was applied to each side, with an amplitude of 5.59 N, to best describe the MAD's operating principle. Figure 1 reports the representation of the boundary conditions application.

Mesh setting

The discretization phase follows the CAD reconstruction and represents the mechanical continuum through discrete elements, interconnected at a discrete number of nodal points, locally equivalent, from a static point of view, to the corresponding portions of the continuous medium. Thus, this process permits passing from a physical to a numerical model.

The models were discretized using 3-dimensional, linear, 4-node tetrahedral elements suitable for modeling complex shapes, such as the irregular surfaces of teeth and PDLs. A general mesh size of 20 mm was applied, with the capture curvature option enabled to refine the mesh in regions with high curvature. For specific areas of interest, such as the teeth and PDL, local mesh sizing was employed to improve accuracy. Conversely, a coarser mesh was applied to the MAD's external surface to reduce the number of nodes and elements, thereby decreasing the model's computational complexity (Figure 2).

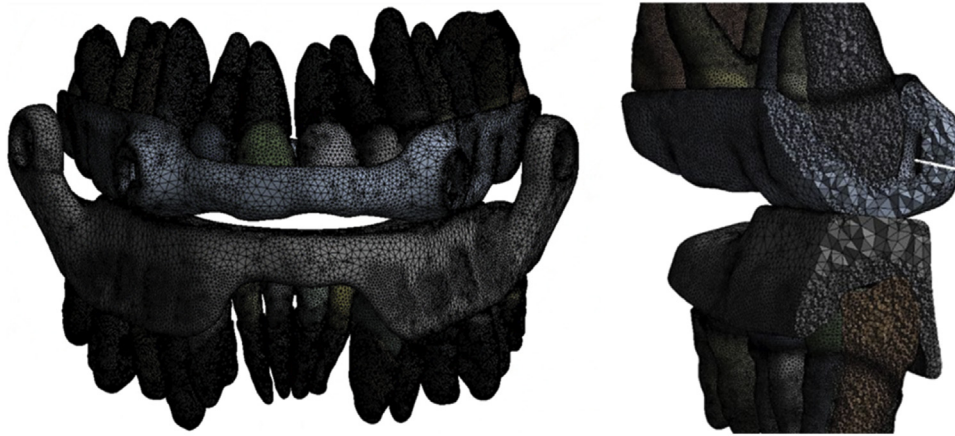


Fig. 2 – Frontal view (left) and cross-sectional view (right) of the mesh for model ID1.

The choice of the element size is strictly related to the body thickness; the thinner the body, the finer the element size. This concept is essential for understanding how the mesh sizing values were determined for each model component. In all simulations, the PDLs (0.3 mm thick) were meshed with an element size of 0.2 mm and a minimum size of 0.1 mm to capture curvature and improve accuracy, given their thinness. This meshing strategy is based on a previously published study by the authors,²⁶ which identified this configuration as an appropriate compromise between mesh quality and computational cost. Moreover, a mesh sensitivity analysis was performed considering a simplified model (half of the arches) to reduce the discretization error.²² The PDL element size was progressively reduced by a factor of approximately 1.3, from 0.26 mm to 0.115 mm, over 4 refinement steps, and von Mises stress was checked for convergence at 5 specific points on the PDL, 4 points of the tooth, and 5 points on the corresponding region of the MAD, according to the procedure described in Dhammayannarangi et al.,²⁷ as illustrated in Figure S2.

The results (Figure S3) show that the stress values at most monitored locations approach a stable solution when the element count reaches about 6 million elements (that is, between a PDL element size of 0.20 and 0.15 mm), except for points 1 and 5 of the PDL, which are close to the cervical margin of the tooth and point T1 of the tooth (buccal cusp). However, the additional analysis indicates that the solution becomes progressively less sensitive to mesh refinement as the PDL element size is decreased to 0.115 mm, while the computational cost increases disproportionately beyond this range (11052103 elements and a 52 GB results file). It is worth noting that the mesh quality, as judged by the Skewness criterion, remained stable across all iterations, with an average

value of 0.238 ± 0.130 , indicating excellent quality. Thus, a PDL element size of 0.20 mm was selected as a practical engineering compromise between numerical accuracy and computational feasibility for the full model.

A local minimum size of 0.2 mm was applied to teeth, and a contact sizing in the regions of contact between MAD and teeth of 0.4 mm was used. Figure 2 (right) shows the inner parts of the meshed ID1 model to make the thickness of each component easier to understand. It is also possible to adjust the local mesh settings to improve accuracy. The FEA was performed using the Ansys Workbench 2023 R1 finite element software.

Results

This section treats the results obtained from the simulations of Panthera's models ID1 and ID2, in terms of displacement of teeth and von Mises stress on teeth and PDLs. Von Mises stress was used to evaluate and compare the mechanical response of the analysed structures under different appliance configurations. Although originally developed for ductile materials, it is widely used in finite element analyses of dental and orthodontic systems because it condenses complex multiaxial stresses into a single comparable parameter, facilitating the identification of stress concentration areas and differences between experimental conditions.^{22,28-30}

They present a dissimilar position of connection elements, localized in correspondence with mandibular 2nd premolars and 1st molars, and likewise for the maxilla in ID2. In a geometric comparison, ID2 shows a 23% increase in volume (Figure 3) due to differences in device shape.

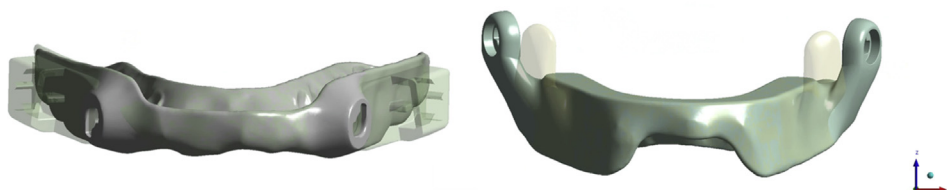


Fig. 3 – Geometric comparison between the upper (on the left) and lower (on the right) splints of ID1 and ID2 (in transparency).

Starting with tooth displacement, Figure 4 presents histograms of the maximum total displacement for each tooth in the upper and lower arches. In contrast, Figure 5 presents a colorimetric representation of the results in different views (frontal and occlusal) using a vector plot with arrows indicating the direction and magnitude of displacement. Concerning the mandible, it is possible to note a different trend, especially in the maxillary canines and mandibular molars. The incisors are not solicited since there is no contact between them and the devices. ID1 shows higher solicitation in the mandibular molars and 2nd premolar regions due to the proximity of the connection elements compared to ID2. Tooth 37 shows the highest value ($1.32 \mu\text{m}$), an increase of 44% compared to ID1 ($83.1 \mu\text{m}$). Conversely, ID2 shows slight increases in deformation in the 1st premolar and canine regions (20% in tooth 33, with a maximum of $69.5 \mu\text{m}$).

The opposite behavior is observed in maxillary teeth (Figure 4). ID1 shows higher solicitations in the 1st premolars/canines (tooth 13 shows an increment of 40%), whilst ID2 shows higher solicitations in the molars and 2nd premolars (maximum increment of 57% recorded in tooth 27). This is because of the different types of MAD, characterized by different positions of the connection elements. The same trend is confirmed by the results on the displacement of the PDLs' inner faces (Figure 6). In Figure 7, the comparative maximum total displacement between the 2 MADs is evident:

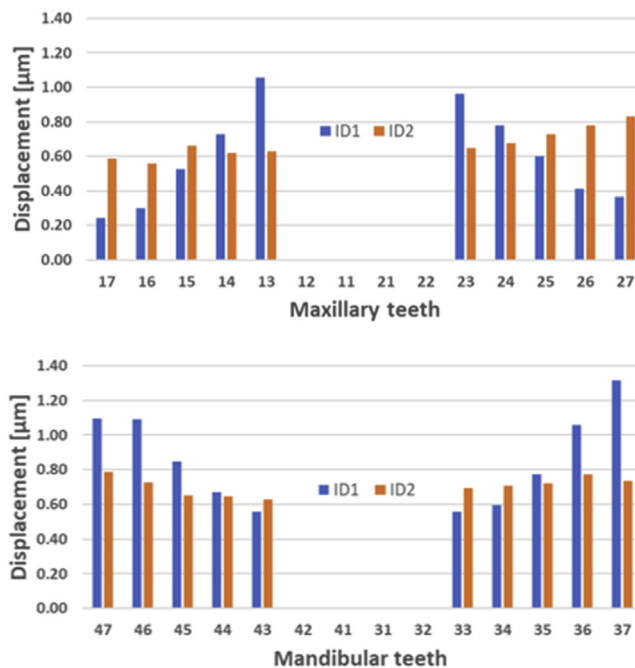


Fig. 4 – Comparison of maximum total displacement (μm) over maxillary (top) and mandibular (bottom) teeth of ID1 (blue) and ID2 (orange) models.

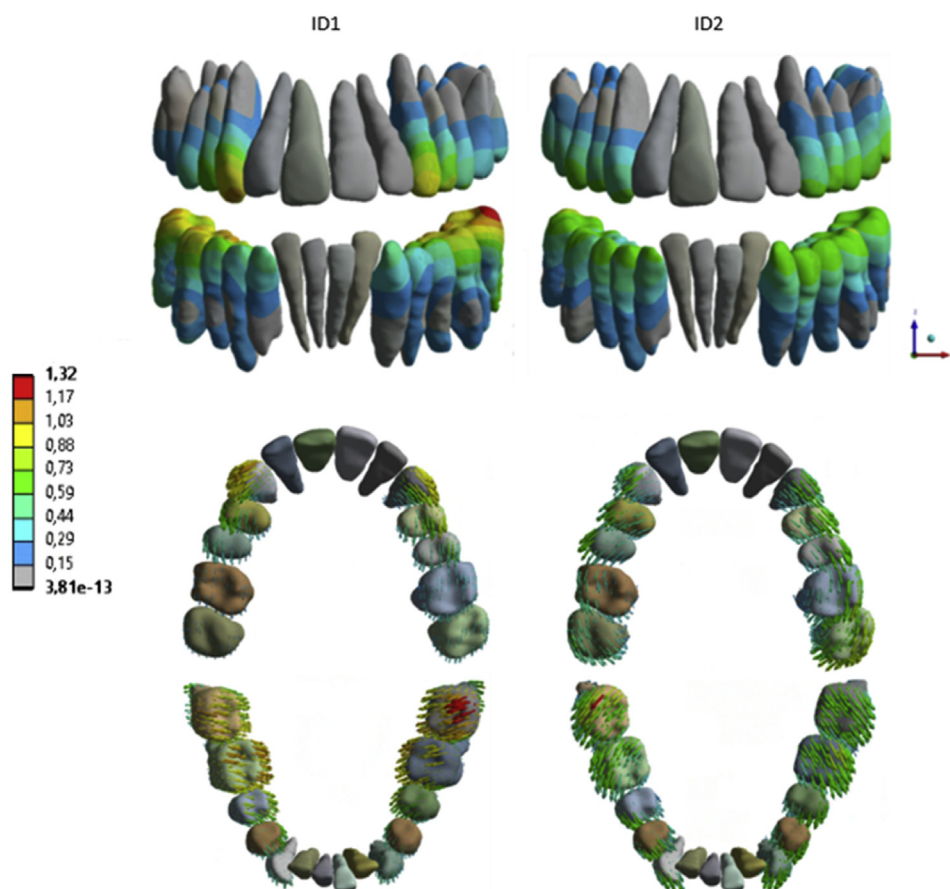


Fig. 5 – Comparative total displacement (μm) over teeth in ID1 (on the left) and ID2 (on the right), in front view (top row) and occlusal view (bottom row).

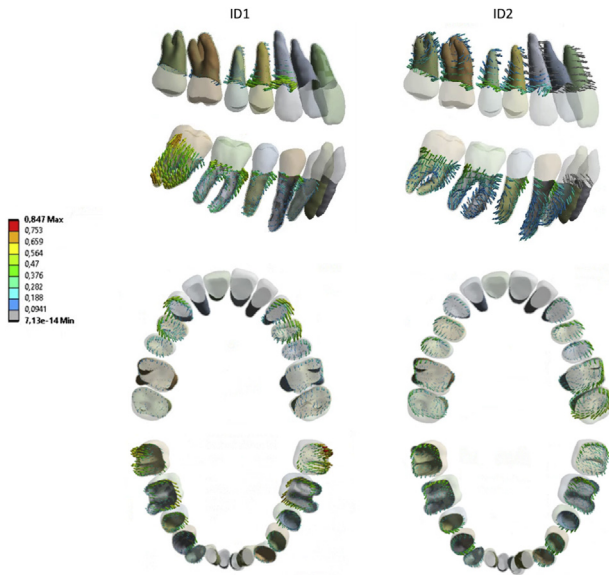


Fig. 6 – Comparative total displacement (μm) over PDL inner faces in ID1 (on the left) and ID2 (on the right), in lateral view (top row) and occlusal view (bottom row).

comparable values are observed in the upper splint ($3.7 \mu\text{m}$ vs $4.2 \mu\text{m}$ in ID1 and ID2, respectively), consistent with the lower displacement increment on the teeth observed in the maxilla compared to the mandible. Indeed, lower displacement peaks are observed in the maxilla, mainly in the canine region of ID1 ($1.06 \mu\text{m}$ vs $1.32 \mu\text{m}$ in mandibular molar 37). Instead, in the lower splint, an evident increase in ID1 is observed ($25 \mu\text{m}$ vs $19 \mu\text{m}$), as the connection pins are most solicited in ID1, leading to greater displacement in the molar region.

A different trend between the simulations can be noted concerning the average stress on teeth (Figure 8).

Mandibular teeth report the highest stress values on the 2nd molars in ID1 (an increment of 73% is evaluated on tooth 37 concerning ID2) since they are subjected to MAD's high solicitations. Premolars and canines show higher values in ID2, with lower but valuable increments (40% in tooth 45, 91.84 kPa). An opposite trend characterizes maxillary teeth. 1st premolars and canines show higher stress in ID1 (tooth 13 reports a stress increment of 31% compared to ID2), whilst

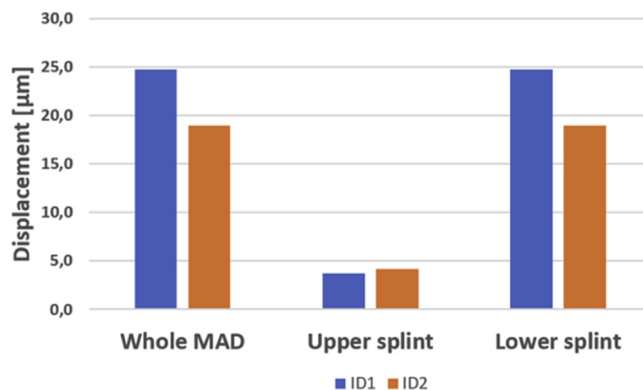


Fig. 7 – Total maximum displacement (μm) on MADs of ID1 (blue) and ID2 (orange) models.

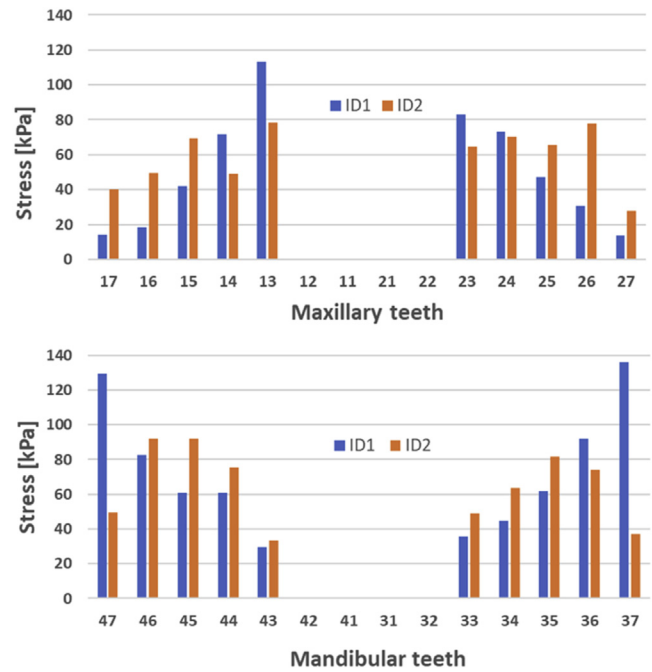


Fig. 8 – Comparison of the average stress over maxillary (top) and mandibular (bottom) teeth of ID1 (blue) and ID2 (orange) models.

molars and 2nd premolars are the most solicited areas in ID2. To better appreciate the results, Figure 9 presents a contour plot of the maximum stress developed in the teeth from frontal and occlusal views.

This behavior is explained by the trend in average stress on MADs, as shown in Figure 10 Figure : the upper splint shows almost identical values (65.5 kPa vs 57.5 kPa), consistent with the lower increment in stress on teeth observed in the maxilla compared to the mandible. Instead, in the lower splint, an evident increase in ID1 is observed (121 kPa vs 74.6 kPa), as the connection pins are most solicited in ID1, leading

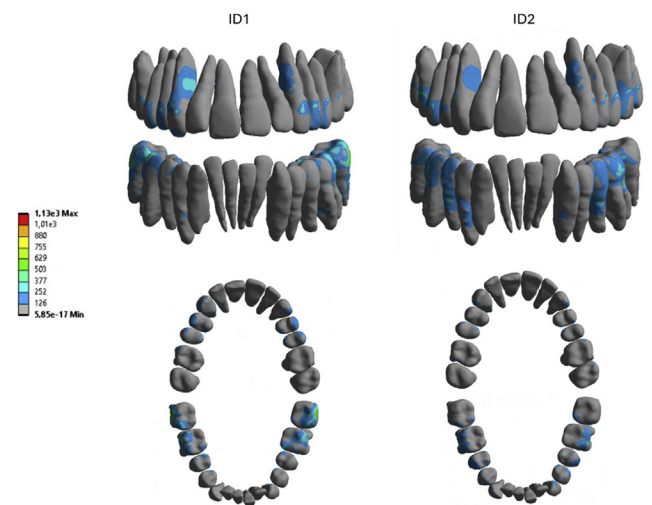


Fig. 9 – Comparison of the maximum stress (kPa) over teeth of ID1 (left) and ID2 (right) models, according to a frontal view (top) and occlusal view (bottom).

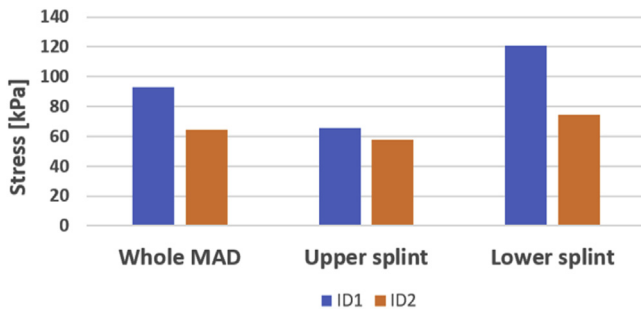


Fig. 10 – Average stress on MADs of ID1 (blue) and ID2 (orange) models.

to higher stresses in the molar region. A color-coded representation of these findings is reported in Figure 11.

Ultimately, average stress on PDLs can also be evaluated (Figure 12), reproducing the trend already described. Mandibular PDLs report the highest stress on molars in ID1 (53% increment of stress in PDL 37, compared to ID2). The PDLs of premolars and canines are more solicited in ID2, with a 42% increase in PDL 34 (0.84 kPa). Maxillary PDLs exhibit an opposite trend and lower solicitations than mandibular ones. Indeed, the highest values are reached on PDL 26 of ID2 (0.81 kPa) and on PDL 13 of ID1 (0.76 kPa). In Figure 13, the contour plot of the maximum stress over PDLs is shown.

Discussion

The FEA results for the Panthera D-SAD Classic (ID1) and Panthera D-SAD X3 (ID2) devices provide valuable insights into the stresses and deformations that occur during MAD application with devices of different structural characteristics. The study highlights how the various connection elements between ID1 and ID2 splints affect stress and deformation distributions, particularly in the molar and premolar regions and in the PDL and teeth.

Firstly, it was found that the highest stress concentrations occur near the connection elements, and that the stress distribution in ID1 is less uniform than in ID2. This achievement is evident in the greater deformations and stresses recorded in the mandibular molars with ID1. In contrast, in ID2, the stresses are more evenly distributed across the entire device surface, especially in the maxilla. This behavior is consistent with previous studies, such as,¹⁰⁻¹² which emphasize that devices with lateral connection elements, such as ID2, exhibit a more uniform stress distribution, similar to other devices

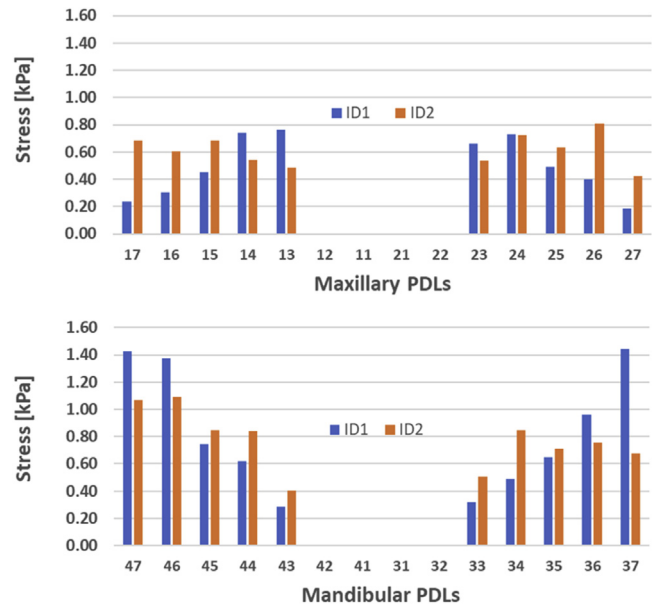


Fig. 12 – Comparison of the average stress over maxillary (top) and mandibular (bottom) PDLs of ID1 (blue) and ID2 (orange) models.

with lateral fins, than those with connection points closer to the posterior teeth.

The uniformity of stress distribution in ID2, despite the 23% increase in volume, represents an advantage. This result suggests a potential reduction in the risk of excessive wear or damage to the teeth and PDL from concentrated stresses, which are more prevalent in ID1. The higher stress peaks recorded in the mandibular molars in ID1 (135.85 kPa compared to 92.17 kPa in ID2) confirm that the position of the connection elements plays a key role in the device's biomechanical performance. Furthermore, the PDLs also experience higher stress in ID1 compared to ID2, which could lead to an increased risk of tissue fatigue. Another interesting finding is the behavior of stresses in premolars and canines, which are more stressed in ID2 than in ID1. This conclusion suggests that, despite the more uniform stress distribution in ID2, certain areas of the device remain subject to increases in stress, particularly near the connection elements. However, this more evenly distributed stress may reduce the risk of overloading individual teeth, a common issue observed in ID1. These results raise essential clinical considerations regarding MADs with different designs, particularly their mechanical impact on dental structures. A critical aspect to consider is the long-term effect of stress distribution on both

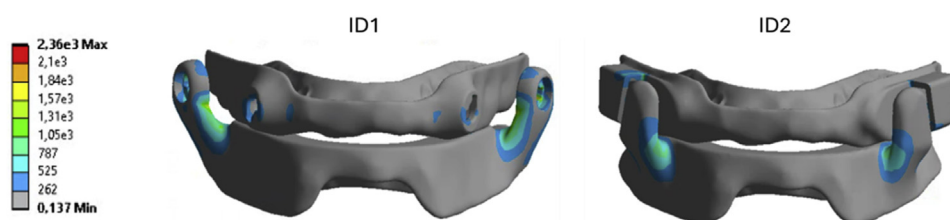


Fig. 11 – Maximum stress (kPa) on MADs of ID1 (left) and ID2 (right) models.

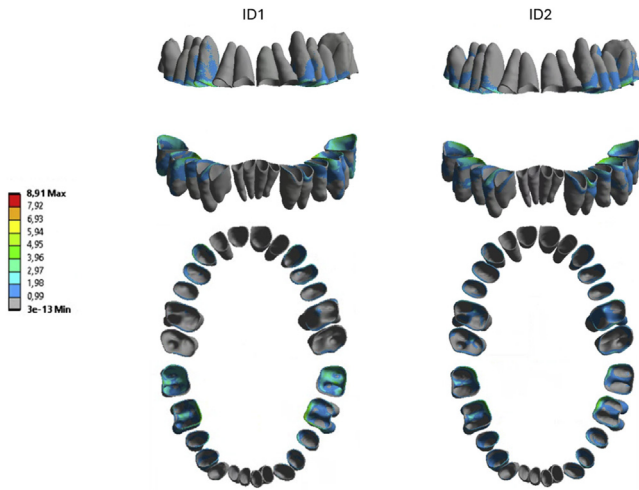


Fig. 13 – Comparison of the maximum stress (kPa) over PDLs of ID1 (left) and ID2 (right) models, according to a frontal view (top) and occlusal view (bottom).

the teeth and the PDL. Therefore, minimizing mechanical stress on the teeth and surrounding tissues can reduce the risk of complications such as tooth mobility, PDL damage, and even bone resorption.^{31,32}

Choosing a more uniform stress-distribution device, such as Panthera D-SAD X3 (ID2), could be especially beneficial for patients with more fragile dental structures or at higher risk of dental and periodontal issues. For instance, patients with pre-existing periodontal disease or reduced bone density could experience a heightened risk of complications if exposed to localized, high-stress concentrations, as seen with devices like ID1. The more uniform distribution of mechanical forces across the dentition, as observed in ID2, may reduce the likelihood of exacerbating these conditions, thereby preserving tooth integrity and periodontal health over the long term.

Additionally, the uniform stress distribution in ID2 can mitigate the risk of appliance-related tooth movement or deformation, which can occur if specific teeth or regions are subjected to excessive forces over extended periods. This situation is particularly relevant when MADs are worn nightly for many months or years, as the cumulative effects of these mechanical forces can lead to undesirable dental shifts or damage to dental tissues. Therefore, the even load distribution offered by ID2 could help maintain occlusal stability and prevent malocclusions that might otherwise arise from the uneven mechanical forces exerted by devices with more localized stress, such as ID1. Further improvements to the ID2 design could be considered, particularly regarding material optimization. For instance, while the device currently exhibits superior performance in stress distribution by reducing material in less mechanically stressed areas, such as the incisor region, this could enhance patient comfort and compliance. Many patients find bulkier devices uncomfortable, especially during long-term use, which affects treatment adherence.³³ Strategically removing or thinning material from areas with minimal mechanical demands may create a more streamlined, comfortable device without compromising mechanical efficiency.

Moreover, a material reduction would not only improve comfort. Still, it could also decrease the device's overall weight, further enhancing patient satisfaction. A lighter, less bulky MAD could be easier for patients to tolerate, especially those with smaller oral cavities or who experience gagging with larger appliances. The balance between material reduction and maintaining structural integrity in high-stress areas, such as around the connection pins, would be key to achieving this improvement.

Although the results of this study are consistent with the actual effects of MADs, the simulated model had several limitations. First, alveolar bone structures were not explicitly modeled, since the primary aim of the study was to compare the relative biomechanical behavior of the 2 MAD configurations rather than to reproduce absolute clinical magnitudes. Thus, while the relative comparison between the 2 devices is not expected to be affected by this simplification, it should be noted that bone structures contribute more substantially to the localization and magnitude of peak stresses. Therefore, the results cannot be quantitatively translated into a real clinical situation.

Indeed, despite prior studies supporting these findings, the use of a single anatomical model limits the generalizability of the results. To enhance clinical relevance, future studies should incorporate multiple patient-specific anatomies to capture inter-individual variability. Moreover, mandibular advancement therapy is typically performed in a gradual, incremental manner, with mandibular protrusion progressively adjusted over time according to the patient's tolerance and therapeutic response. Consequently, the dental and periodontal structures are subjected to repeated, prolonged loading cycles during nightly appliance use, a condition not represented in the present static finite element simulation. The current analysis, therefore, reflects only the immediate mechanical response of the tooth–PDL complex under a single loading condition. Incorporating cyclic and time-dependent loading protocols in future FEM studies would allow a more physiologically realistic evaluation of the cumulative biomechanical effects of MAD therapy on teeth and periodontal tissues. Accordingly, the computed tooth displacements should not be interpreted as long-term orthodontic tooth movement, which depends on biological remodeling processes that were beyond the scope of the present study.

Future research aims to involve a larger patient sample and expand the dataset of results to include other meaningful Panthera configurations that may affect stress distribution in teeth and periodontal structures.

Conclusions

This study employed the finite element method to investigate the biomechanical differences between 2 MADs, Panthera D-SAD Classic (ID1) and Panthera D-SAD X3 (ID2), and their implications for clinical performance and patient outcomes. Despite its larger volume, ID2 demonstrated a more favorable stress distribution profile than ID1, particularly by reducing peak stresses in the mandibular molars and periodontal ligaments (PDLs). The lateral connection design of ID2 contributed to a more uniform transmission of forces across the

dental arch, potentially reducing the risk of localized mechanical overload and associated PDL tissue fatigue.

In contrast, ID1 exhibited higher stress concentrations, especially around its posterior anchorage points, indicating a greater likelihood of strain on specific dental and periodontal regions. These findings suggest that design geometry, especially the placement and orientation of connection elements, plays a critical role in modulating the biomechanical response of the dentoalveolar complex during mandibular advancement. Moreover, although ID2 subjected anterior teeth, such as canines and premolars, to slightly increased stresses, this distribution pattern may contribute to a more balanced load across the arch, reducing the long-term impact on any single tooth or region. Therefore, the enhanced stress management of ID2 may offer a biomechanical advantage in clinical practice, particularly in long-term therapy.

However, it is essential to emphasize that these results are based on a single anatomical model and static loading conditions. Further research involving patient-specific anatomies, dynamic simulations, and clinical follow-up is necessary to fully validate the long-term safety and efficacy of different MAD designs. Understanding how these mechanical differences translate into clinical outcomes will be vital for optimizing treatment planning and improving patient comfort and adherence.

Author contributions

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Data availability statement

Data already presented within the paper.

Conflict of interest

None disclosed.

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Supplementary materials

Supplementary material associated with this article can be found, in the online version, at [doi:10.1016/j.identj.2026.111181](https://doi.org/10.1016/j.identj.2026.111181).

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