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Biomechanical evaluation of cantilever length and arch form in the All-on-4 implant concept: a finite element analysis

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Abstract

Background This finite element analysis (FEA) was performed to evaluate the stress distribution and biomechanical performance of implants, cortical and trabecular bone tissues, and the titanium prosthetic framework in the All-on-4 implant concept. The aim of this study was to investigate the effects of different cantilever lengths and arch forms on the biomechanical behavior of these components.

Methods Digital models representing an edentulous mandible were created in three different arch forms: square, oval, and triangular. In each model, two anterior parallel implants and two posterior tilted implants were placed within the All-on-4 concept. Full arch fixed prostheses with a titanium framework were designed on the implants. For each arch form, cantilever lengths were applied as follows: 5, 10, and 15 mm. To physiologically simulate masticatory forces, the foodstuff loading method was employed. Occlusal loads consisting of a vertical component of 150 N and an oblique component of 100 N were applied to the posterior region. Maximum von Mises stress values were calculated for the implants and titanium framework, whereas maximum and minimum principal stresses were analyzed in the cortical and trabecular bone tissues.

Results In all arch forms, stress values obtained in the implant and titanium framework components under vertical loading were lower compared with those under oblique loading. In contrast, in the cortical and trabecular bone tissues, stress values observed under oblique loading conditions were lower than those under vertical loading. With an increase in cantilever length from 5 mm to 15 mm, stress values in the implant, framework, and bone tissues generally increased across all arch forms. Regardless of loading type, stress values were consistently lowest in the square arch form, followed by the oval and triangular arch forms. However, in the square arch form, higher maximum principal stress values were observed in the cortical bone in the 5 mm cantilever configuration compared with the 10 mm and 15 mm cantilever lengths.

Conclusions The biomechanical performance of the All-on-4 implant concept is influenced by arch form, cantilever length, and loading direction. Clinically, considering patient specific arch form and limiting cantilever length may reduce stress concentrations and support long term success.

Keywords All-on-4, Arch form, Cantilever length, Finite element analysis, Stress distribution

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Background

Complete edentulism is a common clinical condition that occurs when an individual loses all natural teeth in the maxilla and/or mandible, leading to both functional and aesthetic problems. Removable and fixed prosthetic options are commonly preferred for its management. However, conventional complete dentures may be insufficient in terms of retention and stability in the long term due to progressive bone resorption, as they are supported solely by the mucosa and the residual alveolar ridge [1, 2]. In contrast, implant supported fixed prostheses have been shown to reduce bone loss and improve patient satisfaction and functional outcomes [3, 4].

The All-on-4 treatment concept, which has gained significant popularity in recent years, is considered a biomechanically optimized and cost effective solution that simplifies both surgical and prosthetic procedures in patients with complete edentulism [5]. This approach involves the placement of a fixed full arch prosthetic restoration supported by four implants: two parallel implants in the anterior region and two posterior implants placed at an angle to preserve critical anatomic structures. Angled posterior implant placement helps avoid anatomical structures such as the maxillary sinus, nasal cavity, and mandibular canal, thereby reducing the need for advanced surgical interventions including sinus lift procedures, grafting, and bone augmentation [6, 7]. Consequently, this technique shortens surgical time, minimizes postoperative morbidity, and enables patients to achieve functional and aesthetic rehabilitation within a reduced treatment period.

Due to the limited number of implants used in the All-on-4 system, cantilevers are often required, particularly in the posterior regions. These cantilevers directly influence stress distribution in the implants and surrounding bone tissues, depending on their length, position on the prosthesis, and direction of extension [8, 9]. Excessively long cantilevers, in particular, may lead to stress accumulation in the cortical bone under posterior occlusal loading, thereby increasing the risk of undesirable complications such as marginal bone loss, loosening of the prosthetic screw, and fracture of implant or prosthetic components [10]. Therefore, in the All-on-4 system, cantilever length, implant angulation, arch form, and the magnitude and direction of occlusal forces should be carefully planned by considering multiple biomechanical parameters.

Another important factor in treatment planning is the arch form. In All-on-4 prosthetic applications, implant placement and the distribution of cantilever loads may vary depending on the arch form [11]. Clinically, the most common arch forms are classified as square, oval, and triangular. The square arch form allows for a shorter and straighter implant alignment, resulting in a reduced cantilever length and a more homogeneous stress distribution. In contrast, the oval arch form requires implant positioning along a longer and more curved arch, which may lead to an increased cantilever length. In the triangular arch form, narrowing of the anterior region may result in asymmetrical implant placement, potentially causing uneven load distribution and localized stress accumulation in specific implants [12]. Therefore, evaluation of the arch form represents a critical parameter that directly influences the biomechanical success of prosthetic design.

In complex implant prosthetic rehabilitations, such as the All-on-4 treatment concept, detailed and accurate analysis of biomechanical behavior is essential for improving clinical outcomes. In this context, FEA is a powerful method that enables numerical and visual evaluation of the effects of implant placement parameters, cantilever length, arch form, and occlusal forces on implants and the surrounding alveolar bone tissue. Through FEA, regions subjected to critical stress concentrations can be identified in advance, allowing treatment planning to be optimized based on scientific evidence [13]. This is particularly important in All-on-4 applications, where a limited number of implants are placed at specific angulations, as FEA derived data enhance the biomechanical rationale of the treatment plan and contribute significantly to the prediction of long term clinical success.

This study is one of the first to systematically evaluate the biomechanical effects of different arch forms (square, oval, triangular) on of implants, cortical and trabecular bone tissues, and the titanium prosthetic framework within the All-on-4 treatment concept using FEA. While the existing literature generally focuses on the effects of implant placement angles, cantilever lengths, and various materials, arch form has often been considered a standard. In this study, however, the effects of arch form on stress distribution have been compared from a perspective that has not been thoroughly addressed before. The All-on-4 concept presents a biomechanically challenging approach by providing fixed prosthetic treatment with a limited number of implants. Therefore, in addition to optimizing implant placement angles and cantilever lengths, the arch form plays a critical role in the system. Proper planning based on the appropriate arch form can optimize the loads on implants and surrounding bone tissues by balancing the stress distribution in the prosthesis. This innovation provides important guidance for enhancing the success of All-on-4 fixed prostheses in clinical practice and shortening treatment durations. Furthermore, with accurate prosthetic design, a reduction in postoperative complications accelerates the recovery process, allowing patients to achieve functional and aesthetic rehabilitation more quickly. Additionally, minimizing errors during the prosthesis manufacturing process enhances the overall efficiency of the treatment, contributing significantly to both time and cost savings in clinical practice. The null hypothesis of this study is that arch form and cantilever length do not affect stress distribution in implants, cortical and trabecular bone tissues, and the prosthetic framework. Testing this hypothesis is crucial for the biomechanical optimization of treatment planning and for improving the long term success of implant supported fixed prostheses using the All-on-4 concept.

Methods

Study design, hardware and software infrastructure

This study was conducted through a collaboration between Recep Tayyip Erdoğan University Faculty of Dentistry and Tinus Technologies. Three dimensional finite element model generation, mesh optimization, and stress analyses were performed on high performance workstations. Reverse engineering and three dimensional computer assisted design (CAD) procedures were carried out using Blender v4.3 (Blender Foundation, Amsterdam, Netherlands). Solid model preparation for finite element analysis and mesh optimization were performed using Altair HyperMesh v2024 (Altair, Troy, MI, USA), and the finite element simulations were solved using the Altair OptiStruct v2024 solver (Altair, Troy, MI, USA).

Creation of the mandible bone model

The mandibular computed tomography (CT) data used for generating the virtual model were obtained from the Department of Oral and Maxillofacial Radiology at the Faculty of Dentistry, Recep Tayyip Erdoğan University. The scans were acquired using a Planmeca ProMax 3D Classic unit (Planmeca, Helsinki, Finland) at 90 kVp, 4–10 mA, with a voxel size of 200 µm. Ethics approval for this study was granted by the Recep Tayyip Erdoğan University Non Interventional Clinical Research Ethics Committee (Decision No: 2025/499).

The tomography data were imported into 3D Slicer v5.4 (MIT AI Lab, Cambridge, MA, USA) in DICOM format, where segmentation was performed using threshold based methods to isolate mandibular bone structures, consistent with previously reported workflows in dental finite element analyses [14]. Following segmentation, three dimensional surface models of the mandible were generated and exported in STL format.

The resulting surface models were subsequently imported into Blender v4.3, where the cortical bone layer was created by applying a uniform 2 mm inward offset to the outer surface of the mandible model, consistent with previous finite element studies that assumed a 2 mm mandibular cortical thickness [11, 15]. The trabecular bone model was then generated with reference to the inner surface of the cortical bone.

Modeling and scaling of alveolar arch forms

In this study, mandibular models with three different alveolar arch forms (square, oval, and triangular) were modeled using Blender v4.3 software based on reference measurements obtained from two orthodontic studies conducted on young dentate individuals in the Turkish population [16, 17] (Fig. 1). To eliminate the effect of dimensional differences in comparisons among the arch forms, the intercanine distance was standardized. The original measurement values for the different arch forms reported in the literature are presented in Table 1.

The reference intercanine distance was determined as 28.313 mm by calculating the arithmetic mean of the intercanine width values of the different arch forms. Each arch form was scaled to this reference value using a scale factor, calculated by dividing the target intercanine

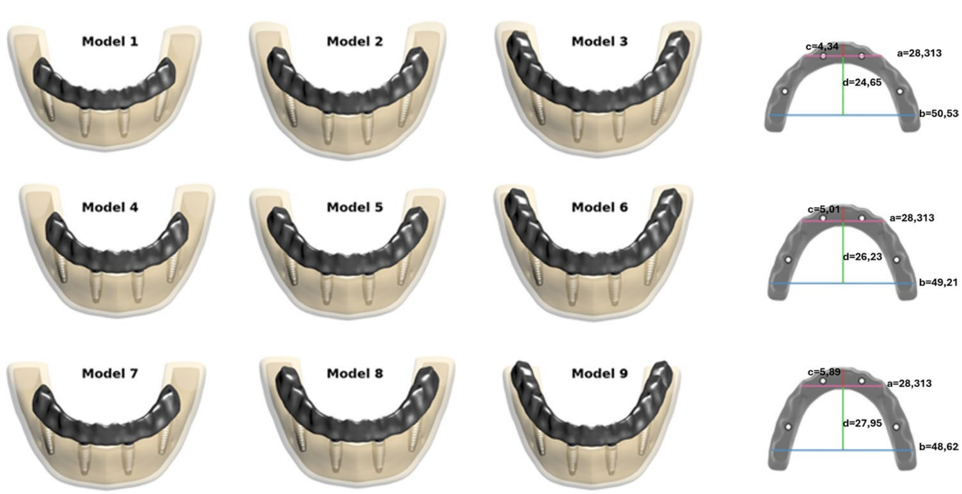

Fig. 1 Models 1–9 show the square (Models 1–3), oval (Models 4–6), and triangular (Models 7–9) arch forms with cantilever lengths of 5, 10, and 15 mm. **a** = intercanine width, **b** = intermolar width, **c** = canine depth, **d** = molar depth

Table 1 Original measurement values for different arch forms reported in the literature

Arch Form	Intercanin Width (mm)	Intermolar Width (mm)	Canine Depth (mm)	Molar Depth (mm)	Canine W/D Ratio	Molar W/D Ratio
Square	28.84 ± 1.71	51.47 ± 1.96	4.42 ± 0.91	25.11 ± 2.20	6.84 ± 1.76	2.07 ± 0.19
Oval	28.58 ± 1.87	49.67 ± 2.04	5.12 ± 0.95	26.48 ± 2.03	5.80 ± 1.34	1.89 ± 0.14
Triangular	27.52 ± 1.68	47.26 ± 2.30	5.72 ± 1.07	27.17 ± 2.13	4.99 ± 1.18	1.75 ± 0.14

Table 2 Scale factors applied to the respective arch forms

Arch Form	Original intercanine width (mm)	Desired intercanine width (mm)	Scale factor(mm)
Square	28.84	28.313	0.9827
Oval	28.58	28.313	0.9916
Triangular	27.52	28.313	1.0288

distance by the original measurement. The scale factors applied to the respective arch forms are presented in Table 2.

The calculated scale factor was applied uniformly to all linear measurements of the corresponding arch form, including intermolar width, canine depth, and molar depth. Consequently, the intercanine distance was standardized across all models, and the final arch form measurements used for the FEA are presented in Table 3. By preserving the characteristic geometric properties of each arch form, this approach enabled the evaluation of biomechanical differences attributable solely to variations in arch form.

Modeling of the implant, framework, and loading conditions

The three dimensional CAD model of the 4.1 × 12 mm Straumann BLT implant (Straumann, Basel, Switzerland) used in this study was generated in Blender v4.3 software based on the manufacturer's catalog data. Multi unit abutments with 0° and 30° angulations, along with corresponding 30° angled abutment screws and occlusal screws, were modeled using the same approach (Fig. 2.) Prosthetic framework models were designed with cantilever lengths of 5, 10, and 15 mm.

For force application, a 10 mm diameter foodstuff sphere was positioned in the molar region, and the mucosal thickness (abutment–gingiva height) was modeled as 1.5 mm. Mesh compatibility between the components

Table 3 Final arch form measurements used for the finite element analysis

Arch Form	Intercanine Width (mm)	Intermolar Width (mm)	Canine Depth (mm)	Molar Depth (mm)	Canine W/D Ratio	Molar W/D Ratio
Square	28.313 ± 1.68	50.53 ± 1.92	4.34 ± 0.89	24.65 ± 2.16	6.53	2.05
Oval	28.313 ± 1.85	49.21 ± 2.00	5.01 ± 0.94	26.23 ± 2.01	5.58	1.88
Triangular	28.313 ± 1.73	48.62 ± 2.34	5.89 ± 1.10	27.95 ± 2.19	4.81	1.74

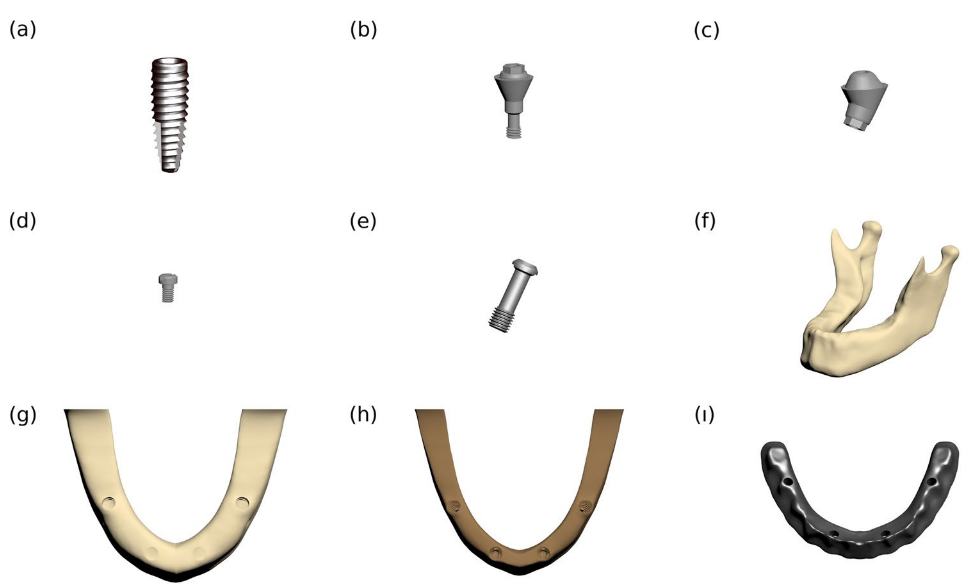

Fig. 2 Components used in the finite element analysis. **a** Implant, **b** Straight abutment, **c** Angled abutment, **d** Occlusal screw, **e** Base screw, **f** Mandible, **g** Cortical bone, **h** Trabecular bone, **i** Titanium framework

Table 4 Finite element model configurations based on arch form and cantilever length

Model	Description
Model 1	Square arch form with 5 mm cantilever
Model 2	Square arch form with 10 mm cantilever
Model 3	Square arch form with 15 mm cantilever
Model 4	Oval arch form with 5 mm cantilever
Model 5	Oval arch form with 10 mm cantilever
Model 6	Oval arch form with 15 mm cantilever
Model 7	Triangular arch form with 5 mm cantilever
Model 8	Triangular arch form with 10 mm cantilever
Model 9	Triangular arch form with 15 mm cantilever

Table 5 Mechanical properties of the materials used in the FEA

Material	Elastic modulus (MPa)	Poisson's ratio
Cortical bone	13,700 [18]	0.30
Trabecular bone	1,370 [19]	0.30
Titanium	114,000 [20]	0.35

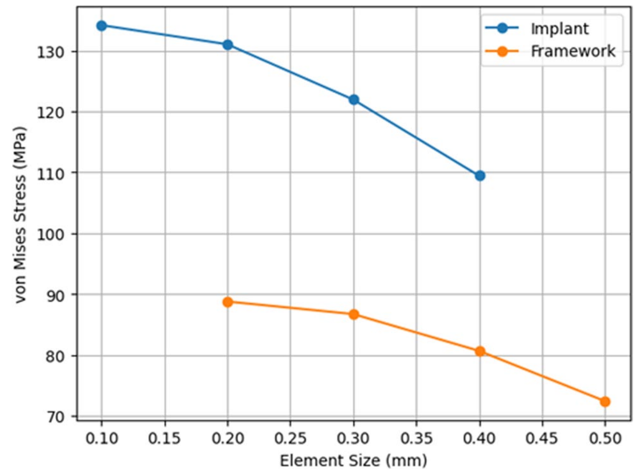

Fig. 3 Mesh convergence analysis showing the relationship between element size and maximum von Mises stress values in the implant and titanium framework

was ensured using ALTAIR HyperMesh software to allow accurate force transfer across the models.

Model configurations

A total of nine finite element models were constructed by combining three different alveolar arch forms (square, oval, and triangular) with three cantilever lengths (5, 10, and 15 mm), as detailed in Table 4.

Material definitions

All materials used in the analysis were assumed to be linear, homogeneous, and isotropic, and were characterized based on their elastic modulus and Poisson's ratio. The mechanical properties of the materials used in the FEA are presented in Table 5.

It was assumed that all models used in this study exhibit homogeneous and isotropic material properties [21]. The interfaces between the implants, abutments and the frameworks were defined as bonded contacts, assuming perfect adhesion and no relative micromotions between the surfaces [22, 23]. This simplification is commonly adopted in comparable FEA studies to represent a rigid and stable connection under static loading conditions [24–26].

Finite element model development and mesh convergence analysis

To enable finite element analysis, the geometric models were discretized into finite elements, thereby converting them into mathematical models. Following the completion of geometric modeling in Blender software (Blender Foundation, Amsterdam, The Netherlands), all models were imported into ALTAIR HyperMesh for mesh generation. Triangular (tria) surface mesh elements with element sizes ranging from 0.10 to 0.25 mm were applied to all models. After surface meshing, three dimensional solid models were generated using tetrahedral solid elements. The resulting finite element models were subsequently transferred to the ALTAIR OptiStruct (Altair Engineering Inc., Troy, MI, USA) solver for numerical analysis.

To ensure the accuracy and reliability of the finite element model, a mesh convergence test was performed. The test was conducted on the model with an oval arch form and a 10 mm distal cantilever, as it was considered geometrically representative. Mesh quality was evaluated for all models; elements with skewness values greater than 80° and those with a minimum edge length below 0.001 mm were corrected prior to analysis. Following successive mesh refinements, the stress values calculated for the implant and framework components exhibited a stable plateau trend (Fig. 3). When the relative error between two consecutive mesh densities decreased below 3%, the solution was assumed to be mesh independent. Accordingly, mesh sizes of 0.1 mm for the implant and 0.2 mm for the framework were determined to be optimal, and the resulting mesh density was applied consistently across all analyses.

Loading protocols and boundary conditions

In the finite element analyses, loading locations were defined according to the cantilever length. In models with cantilever lengths of 10 and 15 mm, a foodstuff sphere was positioned in the central fossa–distal region of tooth number 6 to represent the most distal support area. A vertical load of 150 N and an oblique load of 100 N applied at a 30° angle from buccal to lingual were used in these models [27–31] (Fig. 4a, b).

In models with a cantilever length of 5 mm, the loading was applied to the posterior region encompassing

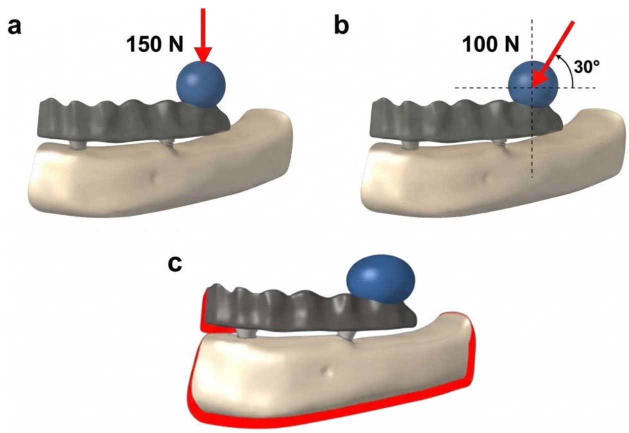

Fig. 4 Loading protocols and boundary conditions applied in the finite element models. (**a**) Vertical loading of 150 N, (**b**) oblique loading of 100 N at a 30° angle, and (**c**) boundary conditions showing fully fixed distal and basal regions of the cortical and trabecular bone

Table 6 Total number of nodes and elements for each finite element model

Model	Total number of nodes	Total number of elements
Model 1	344,179	1,385,805
Model 2	349,544	1,407,769
Model 3	354,463	1,427,390
Model 4	343,304	1,383,205
Model 5	347,978	1,402,229
Model 6	352,836	1,422,017
Model 7	336,316	1,350,338
Model 8	340,923	1,369,082
Model 9	345,799	1,388,808

teeth numbers 5 and 6 in order to accurately simulate distal loading effects associated with the shorter cantilever. In these models, a total vertical load of 150 N and an oblique load of 100 N at a 30° angle were defined.

Cyclic and dynamic occlusal loads more closely represent physiological mastication; however, static linear finite element analysis (FEA) remains a widely adopted approach in implant biomechanics research [22, 32] Although dynamic loading may increase stress magnitudes compared to static conditions, static analyses provide valid and fundamental insights for comparing relative stress patterns among different models [22, 33].

The applied forces were distributed across circumferential nodal points to prevent stress singularities at the loading sites. Regarding boundary conditions, the models were fixed at the nodal points located in the distal and basal regions of the cortical and trabecular bone, with all degrees of freedom constrained in three axes (Fig. 4c). Under the defined loading and boundary conditions, a total of 18 linear static analyses were performed for nine different analysis models.

Mesh characteristics of the finite element models

The number of nodes and elements of the nine models used in FEA is presented in Table 6. Similar mesh density was provided in all models, ensuring the comparability of the results.

Contact definitions and connection assumptions

In the finite element analyses, contact definitions were applied to accurately represent the mechanical interactions between the model components. A FREEZE (bonded) contact condition was assigned between all contacting components, assuming no relative displacement at the interfaces of the parts. This approach was adopted to model ideal osseointegration at the implant–bone interface and rigid connections between the prosthetic components.

Results

The maximum von Mises stress values observed in the anterior and posterior implants are summarized in Fig. 5, while the corresponding von Mises stress distributions in the implants under vertical and oblique loading conditions are illustrated in Figs. 6 and 7, respectively. Evaluation of the maximum von Mises stress distributions in the implants revealed that, irrespective of the loading direction, the lowest stress values were consistently observed in models with a square arch form, followed by the oval and triangular arch forms. Posterior implants located closer to the load application area exhibited higher stress values than the anterior implants.

Overall, an increase in cantilever length from 5 mm to 15 mm resulted in a progressive increase in maximum von Mises stress values in both the implant and titanium framework components across all arch forms. The maximum von Mises stress values calculated in the titanium framework are summarized in Fig. 8, while the corresponding stress distributions under vertical and oblique loading conditions are presented in Figs. 9 and 10, respectively.

Stress concentrations in the titanium framework were predominantly observed in the posterior cantilever regions and in areas adjacent to the implant–framework connection sites, and the stress distribution patterns were found to be similar to those observed in the implants.

Stress values generated in the cortical bone were higher than those observed in the trabecular bone; however, both bone types demonstrated comparable stress distribution patterns. The maximum principal stress values in the cortical bone are summarized in Fig. 11, while the corresponding principal stress distributions under vertical and oblique loading conditions are illustrated in Figs. 12 and 13, respectively. Evaluation of the

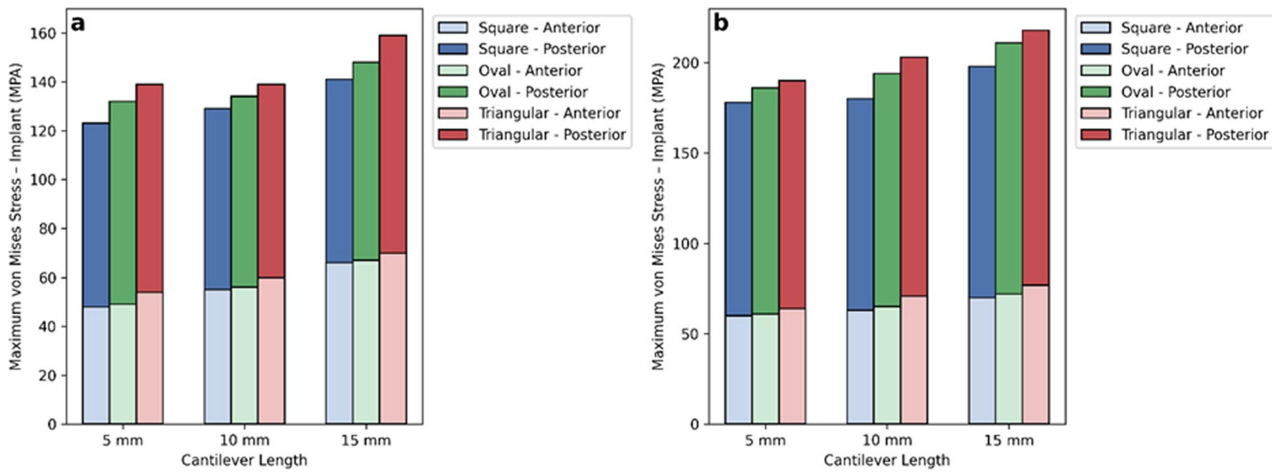

Fig. 5 Maximum von Mises stress values (MPa) in anterior and posterior implants for square, oval, and triangular arch forms at different cantilever lengths (5, 10, and 15 mm) under (**a**) vertical and (**b**) oblique loading conditions

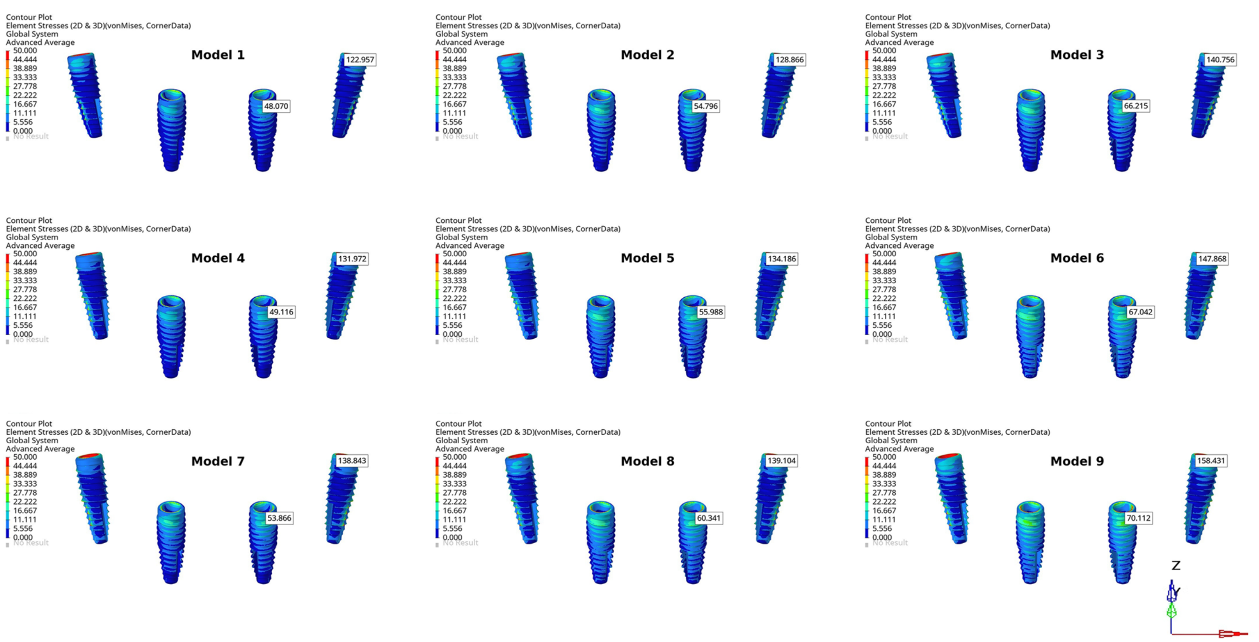

Fig. 6 Maximum von Mises stress distributions in the implants for all finite element models under vertical loading conditions

principal stresses in the bone tissue demonstrated that, consistent with the findings for the implant and framework components, the square arch form exhibited the lowest stress values, whereas the triangular arch form showed the highest values. In the cortical bone, principal stress concentrations were primarily localized in the crestal regions adjacent to the implant necks and extended over wider posterior areas with increasing cantilever length.

With respect to loading direction, higher stress values were observed in the implant and framework components under oblique loading compared with vertical loading. In the framework components, oblique loading resulted in stress distributions extending over broader areas and concentrating predominantly in the posterior regions. In contrast, higher maximum principal stress values were observed in both cortical and trabecular bone tissues under vertical loading conditions. The maximum principal stress values in the trabecular bone are summarized in Fig. 14, while the corresponding principal stress distributions under vertical and oblique loading conditions are shown in Figs. 15 and 16, respectively. In the trabecular bone, principal stress distributions remained localized around the implants, particularly in regions adjacent to the apical portions of the implants; however, the area of stress distribution gradually expanded in the posterior regions as the cantilever length increased.

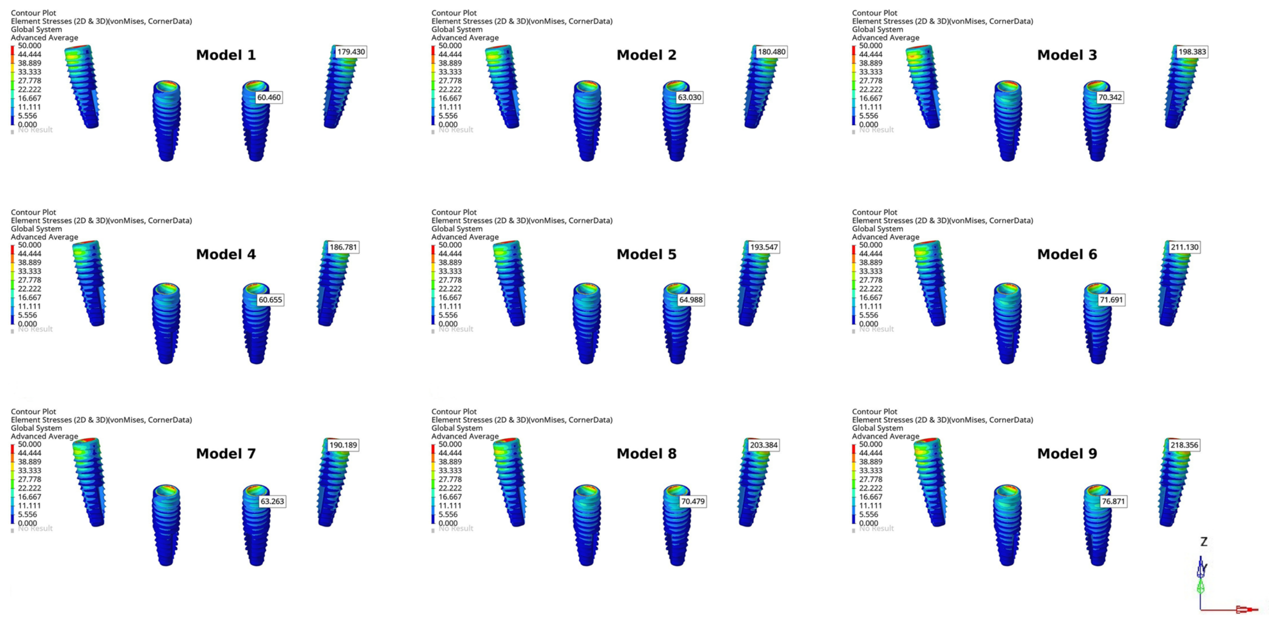

Fig. 7 Maximum von Mises stress distributions in the implants for all finite element models under oblique loading conditions

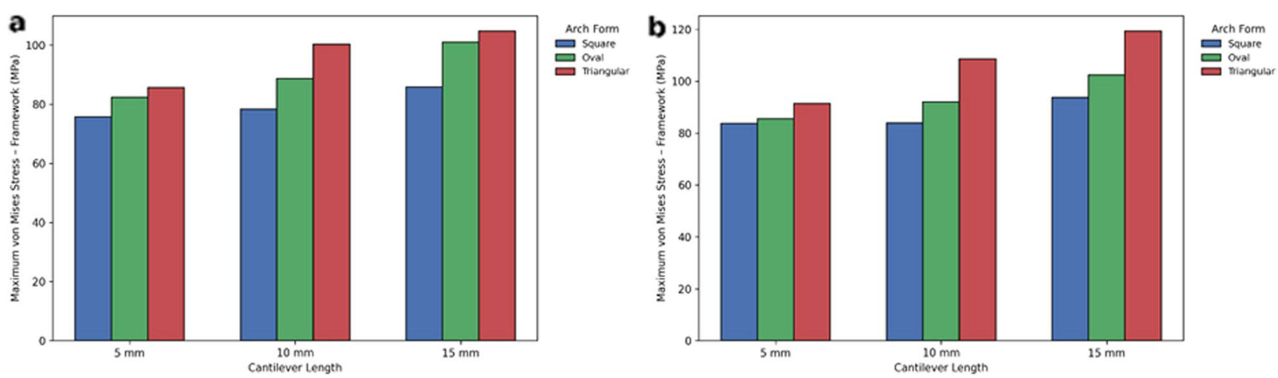

Fig. 8 Maximum von Mises stress values (MPa) in the framework for square, oval, and triangular arch forms at different cantilever lengths (5, 10, and 15 mm) under (**a**) vertical and (**b**) oblique loading conditions

Although increasing cantilever length was generally associated with elevated stress values in all components, an exception was observed in the square arch form. In this configuration, particularly under oblique loading conditions, the maximum principal stress values in the cortical bone were higher in the 5 mm cantilever model (14.149 MPa) than in the 10 mm (13.010 MPa) and 15 mm (13.303 MPa) cantilever models.

Across both vertical and oblique loading scenarios, stress values followed a consistent hierarchical pattern among the different arch forms, with the square arch form exhibiting the lowest stress values, followed by the oval and triangular arch forms. Finally, the maximum von Mises stress values calculated in the titanium implant and framework components remained below the yield strength of titanium (890 MPa), indicating that no material failure was expected under the simulated loading conditions.

Discussion

This study evaluated the effects of different cantilever lengths (5 mm, 10 mm, and 15 mm) and mandibular arch forms (square, oval, and triangular) on stress distribution in the implant, titanium framework, cortical and trabecular bone tissues within the All-on-4 concept. The null hypothesis, which assumed that cantilever length and arch form would have no effect on stress distribution, was rejected. The findings demonstrated that stress distribution was influenced by cantilever length, mandibular arch morphology, and loading type. Notably, the square arch form demonstrated a more favorable biomechanical behavior, as increases in cantilever length did not result in a monotonic increase in stress levels, particularly under oblique loading conditions. In general, increasing cantilever length resulted in higher stress concentrations around the posterior implants and cortical bone, whereas shorter cantilever configurations exhibited lower stress

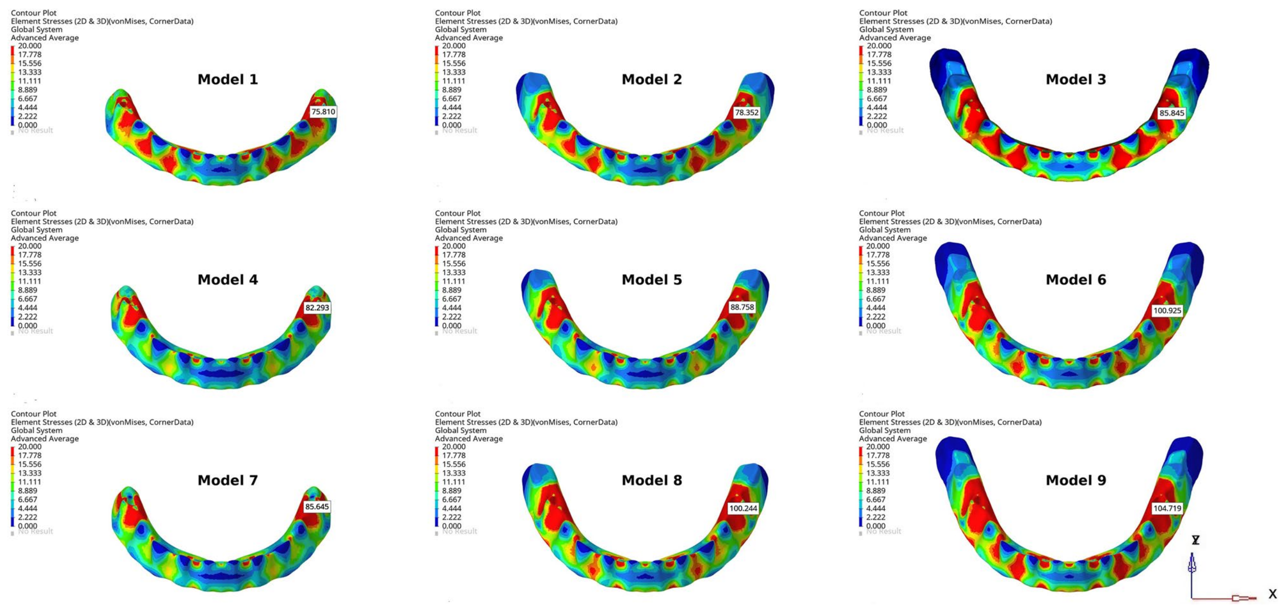

Fig. 9 Maximum von Mises stress distributions in the framework for all finite element models under vertical loading conditions

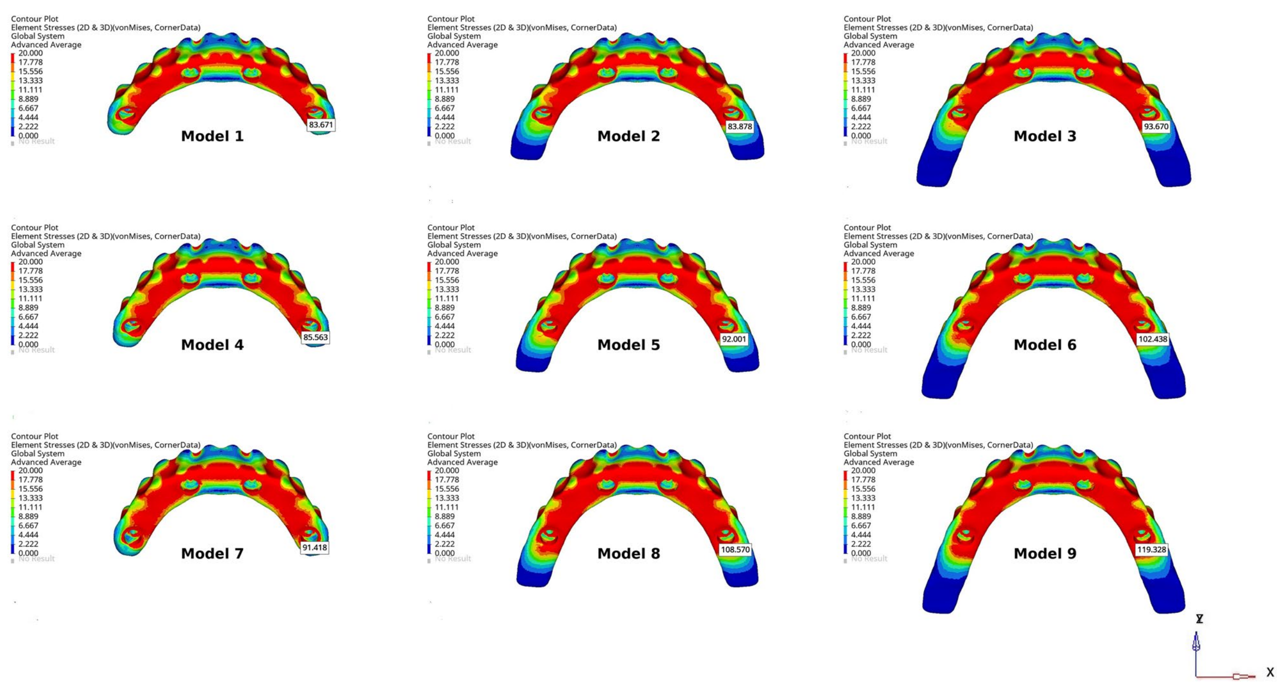

Fig. 10 Maximum von Mises stress distributions in the framework for all finite element models under oblique loading conditions

values. Regarding arch form, stress distribution was relatively more homogeneous in the square arch form, while localized stress concentrations were observed in the triangular arch form, likely due to geometric limitations and differences in implant placement. The oval arch form exhibited an intermediate biomechanical behavior between these two configurations.

The All-on-4 treatment concept was developed by Dr. Paulo Malo and colleagues and has been widely and successfully applied in clinical practice [34]. This approach reduces the need for advanced surgical interventions such as bone grafting and sinus lift procedures, thereby decreasing treatment duration and cost. In addition, supporting fixed restorations with a limited number of implants enhances patient comfort and enables the achievement of functionally and esthetically satisfactory outcomes within a short period of time. Owing to these advantages, the All-on-4 concept has become an effective

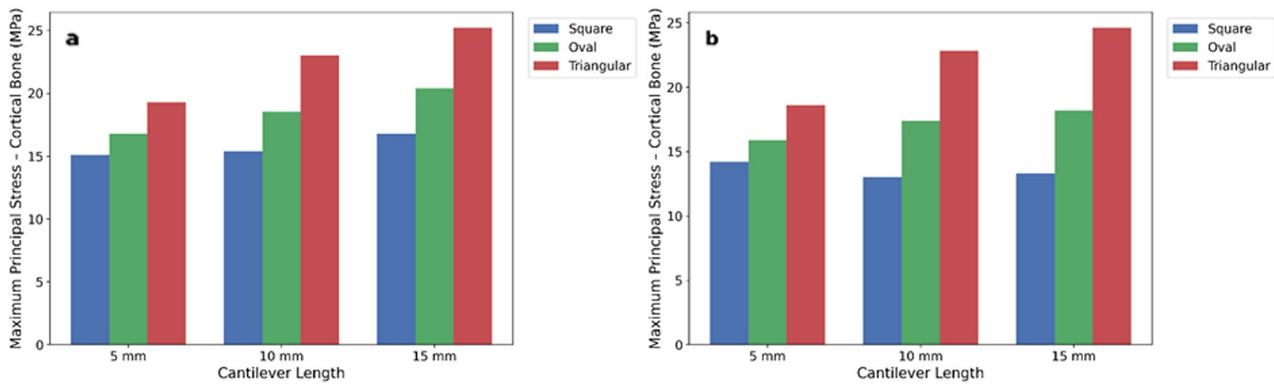

Fig. 11 Maximum principal stress values (MPa) in the cortical bone for square, oval, and triangular arch forms at different cantilever lengths (5, 10, and 15 mm) under (**a**) vertical and (**b**) oblique loading conditions

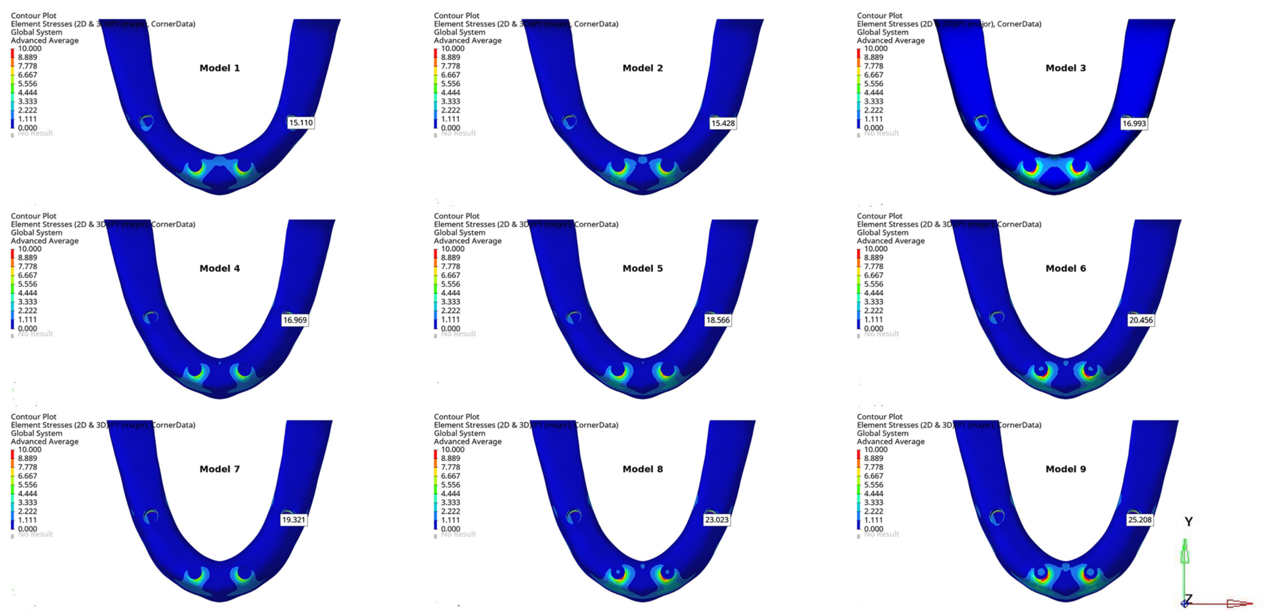

Fig. 12 Maximum principal stress values (MPa) in the cortical bone for all finite element models under vertical loading conditions

treatment option, particularly for elderly patients and individuals with systemic conditions [35]. Due to the complication risks associated with surgical procedures aimed at increasing bone volume, alternative treatment strategies such as the use of short implants, angled implants, as well as pterygoid and zygomatic implants have been recommended in recent years [36–38].

Within the All-on-4 concept, angled placement of posterior implants is a fundamental principle to overcome anatomical limitations and to reduce distal cantilever length [39]. The angulation of posterior implants is determined based on residual bone volume, the anteroposterior implant distance, and prosthetic planning. Clinical studies have reported comparable success rates between axial and angled implants [40]. However, some biomechanical studies have demonstrated that increasing implant angulation may lead to elevated stress levels in the surrounding bone [41–43]; nevertheless, other studies have reported that angled implants and abutments do not adversely affect clinical success [40, 44]. Accordingly, in the present study, posterior implants were placed at an angle in accordance with the defined protocol of the All-on-4 concept.

In the literature, the biomechanical effects of inclined implants have predominantly been investigated at angulation ranges between 15° and 45°. Liu et al. [45] reported that more balanced stress distribution and reduced micromovement at implant–bone interfaces were observed at implant angulations of 30° and 45°. Similarly, Güçlü et al. [46] demonstrated that a 30° inclination reduced the maximum stress levels in the cortical bone. In contrast, Tribst et al. [47] reported that a 45° angulation provided a biomechanical advantage only in the presence of a short cantilever, whereas longer cantilevers

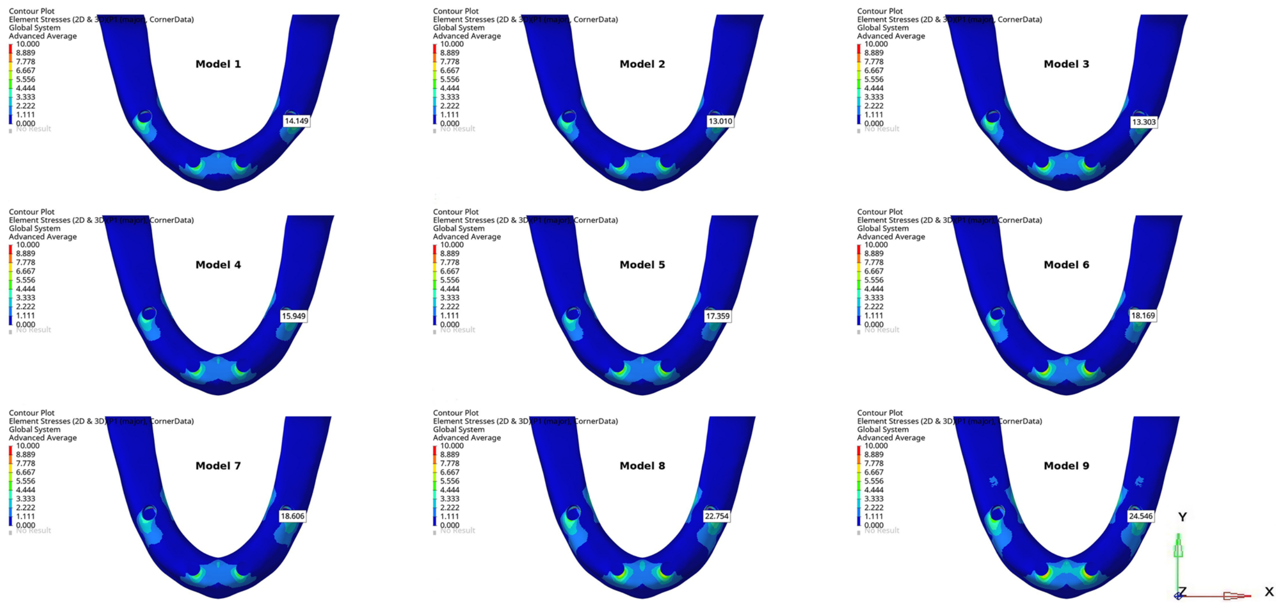

Fig. 13 Maximum principal stress values (MPa) in the cortical bone for all finite element models under oblique loading conditions

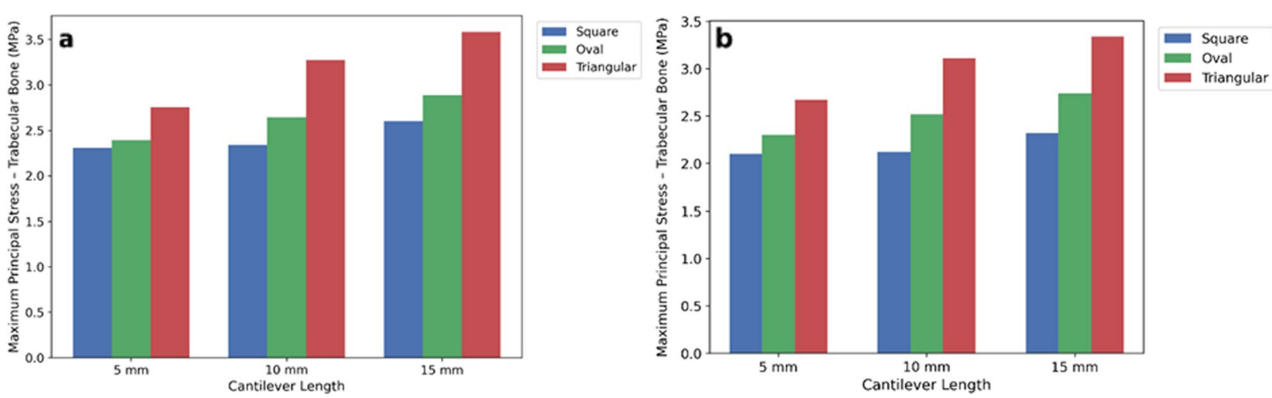

Fig. 14 Maximum principal stress values (MPa) in the trabecular bone for square, oval, and triangular arch forms at different cantilever lengths (5, 10, and 15 mm) under (**a**) vertical and (**b**) oblique loading conditions

resulted in increased stress concentration at the implant neck. In accordance with these findings, a posterior implant angulation of 30° was selected in the present study.

The All-on-4 concept aims to provide fixed prosthetic rehabilitation using inclined posterior implants in order to reduce the need for advanced surgical procedures. Due to limited posterior support, the presence of a distal cantilever is unavoidable in most cases and must be carefully planned, as occlusal loads are transferred to the anterior implants. The literature suggests that cantilever length should not exceed 1.5 times the anteroposterior (AP) distance between implants [48]. Finite element analyses reported in the literature consistently demonstrate that the presence of a cantilever introduces significant biomechanical disadvantages. Wang et al. [6] reported that distal cantilevers increased marginal bone stress, whereas elimination of the cantilever through 45° implant angulation reduced stress levels. Ileri et al. [8] showed that stresses in the implant and peri implant bone increased linearly with increasing cantilever length. Similarly, Bevilacqua et al. [49] demonstrated that shortening the cantilever by means of angled implant placement significantly reduced stress values. Recent studies further support these findings, identifying posterior implant angulation and cantilever length as primary determinants of stress distribution. Deste and Durkan [50] reported that increasing posterior implant angulation elevated cortical bone stress and that the presence of a cantilever amplified this effect. Chen et al. [51] indicated that the influence of framework material on stress distribution was limited, whereas implant angulation and cantilever length were the dominant factors. Wang et al. [6] reported that stress concentration at the neck of angled

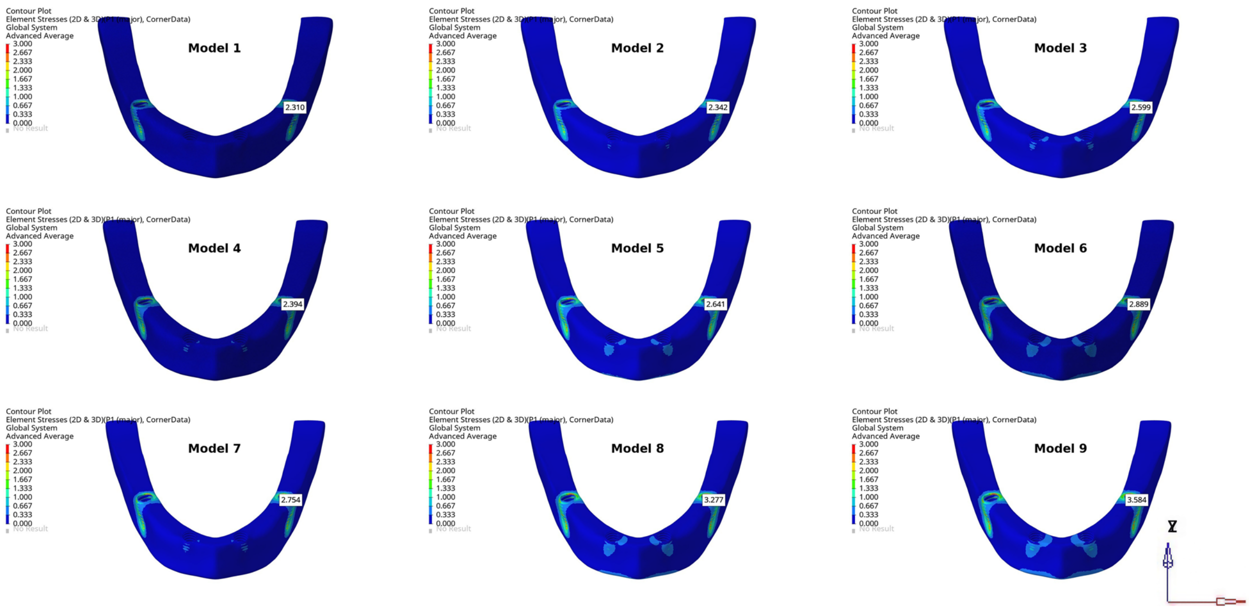

Fig. 15 Maximum principal stress values (MPa) in the trabecular bone for all finite element models under vertical loading conditions

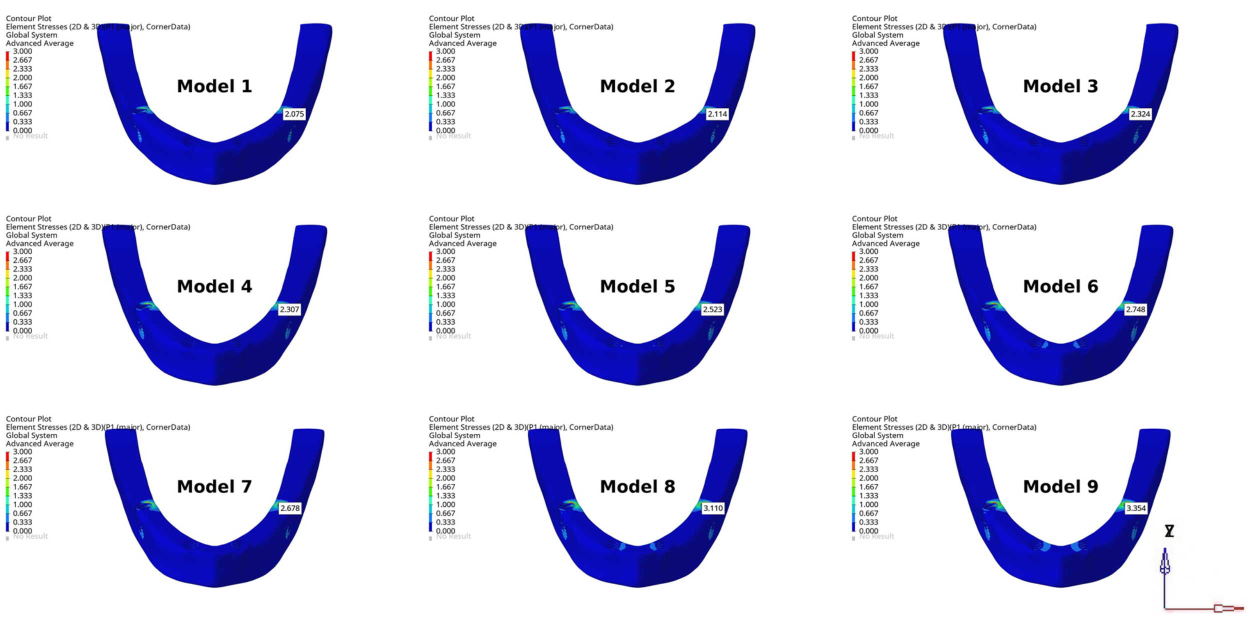

Fig. 16 Maximum principal stress values (MPa) in the trabecular bone for all finite element models under oblique loading conditions

implants became more pronounced as cantilever length increased. Collectively, these studies indicate that distal cantilevers substantially increase stress accumulation in the posterior region, whereas shortening or eliminating the cantilever through angled distal implant placement results in a more favorable biomechanical stress distribution. In line with these findings, the present study demonstrated that increasing distal extension led to greater stress accumulation in most models, while the influence of arch form was preserved, showing general agreement with previously reported biomechanical data.

There are also studies in the literature reporting a decrease in stress levels with increasing cantilever length [9, 10]. Durkan et al. [9] suggested that altering the direction of bending moments through the use of rigid superstructure materials and implant inclination allows load distribution over a wider area, thereby reducing stresses on posterior implants and framework components. Similarly, Almjaddr and Saker [10] reported reduced stress values in models with long cantilevers when PEEK frameworks with a low elastic modulus were used. The stress reduction observed in these studies has been attributed to the elastic properties of the framework materials, the higher load magnitudes applied, and the more flexible mechanical behavior of the system under different loading scenarios. In the present study, the framework was fabricated from titanium, a material characterized by a high elastic modulus and rigid structural behavior. Consequently, increasing cantilever length resulted in elevated stress values in the implant and framework components. This finding can be explained by the limited ability of the rigid titanium framework to dissipate bending moments generated by longer cantilevers, leading to stress concentration within a more localized region.

When the stress values of the anterior and posterior implants and the prosthetic framework were evaluated collectively, it was evident that the effect of increasing cantilever length on stress distribution was non linear. Across all arch forms and under both vertical and oblique loading conditions, increasing the cantilever length from 5 mm to 10 mm generally resulted in a limited increase in stress; however, extending the cantilever from 10 mm to 15 mm led to a pronounced and disproportionate increase in stress, particularly in the posterior implants. This behavior may be attributed to the posterior implant–framework junction acting as the primary pivot point of the system as the cantilever lengthens, resulting in the concentration of increasing bending moments in this region. The similar stress increase pattern observed in the posterior implants across all arch forms suggests that this response reflects a cantilever mechanics related threshold effect that is largely independent of loading direction.

In contrast, analysis of prosthetic framework stresses revealed that this threshold behavior varied according to arch form. In the square and oval arch forms, increasing the cantilever length from 10 mm to 15 mm resulted in a more pronounced increase in framework stresses, whereas in the triangular arch form, framework stresses increased more sharply between 5 mm and 10 mm and exhibited a plateau like trend between 10 mm and 15 mm. This finding may be explained by the more limited posterior support width of the triangular arch form, which leads to the attainment of critical stress levels at shorter cantilever lengths, while increased system rigidity at greater cantilever extensions partially constrains further stress escalation. Consequently, differences in absolute stress magnitudes and threshold points appear to be closely associated with the load sharing capacity determined by arch morphology.

Moreover, across all arch forms, oblique loading resulted in lower stress values in the cortical and trabecular bone compared with vertical loading. Although oblique loading is generally associated with increased shear forces in peri implant bone, this finding may be explained by the combined effects of posterior implant angulation and framework rigidity, which may have promoted load transfer toward the implant–framework complex rather than the surrounding bone tissue [52]. As a result, higher stress accumulation was observed primarily within the implant and framework components, while bone stresses remained comparatively lower under oblique loading conditions.

Cantilever length and implant inclination are important factors influencing stress distribution; however, mandibular arch morphology also plays a critical role in determining biomechanical behavior. Arch form directly affects the anteroposterior (A–P) distance, and increasing the A–P distance—particularly through the addition of an anterior implant—may improve stress distribution [14]. More distal positioning of implants effectively shortens the posterior cantilever, thereby reducing stress levels in both the peri implant bone and the prosthetic framework [53].

Gül et al. [11] evaluated All-on-4 and All-on-5 configurations in different mandibular arch types (square, U, and V) and reported that increasing the number of implants reduced cortical bone stress, whereas stress levels were significantly higher in All-on-4 systems, particularly in V type arches, due to the cantilever effect. Similarly, Ulu and Ercan [12] investigated the influence of maxillary arch morphology on stress distribution and demonstrated that the triangular arch form with the addition of an anterior implant resulted in the greatest reduction in cortical and trabecular bone stress; however, this effect was less pronounced in square and oval arch types. Collectively, these findings highlight the critical biomechanical importance of optimizing implant number and placement according to patient specific arch morphology. In the present study, higher stress concentrations were observed in the triangular arch form, consistent with the findings reported by Gül et al. [11] and Ulu and Ercan [12]. The literature emphasizes that stress reduction is more effectively achieved by increasing the number of implants or optimizing implant positioning rather than extending cantilever length. In this context, considering the greater bending moments and associated stress accumulation that occur with increasing cantilever length in triangular arch forms, increasing the number of implants appears to be a more biomechanically advantageous strategy in clinical practice. Such an approach may reduce stress levels in both the peri implant bone and implant components by enabling implant placement compatible with arch morphology.

The literature on the All-on-4 treatment concept includes numerous biomechanical studies investigating various loading conditions, cantilever lengths, framework materials, and implant numbers. However, the majority of these studies assume a fixed mandibular arch morphology and address the influence of different arch forms on stress distribution only to a limited extent. In this context, the present study provides an original contribution to the literature by being among the limited number of investigations that comparatively evaluate stress accumulation associated with different cantilever lengths across varying mandibular arch forms.

In line with the findings of the present study, particularly in the square arch form, maximum principal stresses in the cortical bone did not show a monotonic increase with increasing cantilever length under oblique loading conditions. This behavior suggests that the square arch form distributes occlusal loads more effectively due to its wider posterior support area, resulting in a more favorable biomechanical response to cantilever elongation. These findings indicate that arch morphology plays a determining role in stress distribution and should be considered a critical parameter in cantilever design. Accordingly, incorporating individual arch morphology into treatment planning appears to be essential for optimizing the biomechanical performance of All-on-4 prosthetic rehabilitations.

The findings of this study demonstrate that increasing the cantilever length from 5 mm to 10 mm results in a smaller increase in implant and framework stress values, particularly in square and oval arch forms, compared to the transition from 10 mm to 15 mm. In the triangular arch form, with the exception of the stresses in the implants, an increase from 5 mm to 10 mm caused a more significant rise in framework, cortical, and trabecular bone stress values than the increase from 10 mm to 15 mm. These findings emphasize the critical role of arch morphology and cantilever length in prosthetic design and the optimization of biomechanical performance.

In biomechanically disadvantageous designs like All-on-4, where the number of implants is limited and posterior implants are angulated, prosthetic design and patient specific factors must be carefully considered. In patients with square arch forms, increasing the distal extension may provide a safer biomechanical response, while in patients with triangular arch forms, limiting the distal extension and increasing the number of implants may be an important strategy for minimizing stress accumulation. This study highlights the critical role of mandibular arch morphology and cantilever design in the success of All-on-4 prosthetic rehabilitations.

These findings underscore the significance of arch morphology and cantilever length as essential parameters in implant design and clinical outcomes. Customizing the cantilever design according to the arch form, optimizing implant placement and number, can enhance the biomechanical performance and longevity of the prosthesis.

This study has several limitations. The finite element models represent idealized anatomical conditions and therefore do not fully account for inter individual variations in bone density, cortical thickness, or anatomical morphology. In addition, the assumption of materials as homogeneous, isotropic, and linearly elastic does not reflect the viscoelastic and heterogeneous behavior of biological tissues. Chewing forces were applied with fixed magnitudes and directions and thus did not simulate dynamic, multi directional, or time dependent clinical loading conditions. Moreover, prosthetic attachment details, screw pre tensioning, parafunctional habits, and soft tissue effects were not incorporated into the model. Despite these limitations, the present study provides valuable biomechanical insight by comparatively evaluating the effects of cantilever length and arch morphology, and may serve as a reference for future experimental and clinical investigations.

Conclusions

This study demonstrated that both cantilever length and mandibular arch form play an important role in biomechanical stress distribution within the All-on-4 concept. Shorter cantilever configurations and the square arch form were generally associated with lower stress values in the implants, cortical bone, and prosthetic framework. In contrast, higher stress concentrations were observed in models combining a triangular arch form with a 15 mm cantilever, particularly around posteriorly angled implants. These findings underscore the biomechanical importance of evaluating arch morphology in conjunction with cantilever design. Therefore, incorporating individual arch morphology into clinical treatment planning and limiting cantilever length whenever possible may contribute to a more favorable and balanced stress distribution in All-on-4 rehabilitations.

Further studies incorporating different framework materials, implant dimensions, dynamic loading conditions, and long term clinical data are required to achieve a more comprehensive understanding of the biomechanical behavior of All-on-4 systems.

Abbreviations

MPa Megapascal
FEA Finite element analysis
CT Computed tomography

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Not applicable.

Authors' contributions

HOB and ZY contributed to the conceptualization of the study, writing, and preparation of the original draft. ZY contributed to data curation, interpretation of the results, formal analysis, and critical revision of the manuscript. HOB contributed to methodology design, interpretation of the results data collection, literature review, and figure preparation. Both authors contributed to the review, editing, and final approval of the manuscript. All authors have read and approved the published version of the manuscript.

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Data availability
All data generated or analysed during this study are included in this published article.

Declarations

Ethics approval and consent to participate
This study was reviewed and approved by the Non-Interventional Clinical Research Ethics Committee of Recep Tayyip Erdoğan University during the meeting held on December 25, 2025 (Decision No: 2025/499). Written informed consent was obtained from the participant included in the study. All procedures were carried out in accordance with the ethical principles of the Declaration of Helsinki.

Consent for publication
Not applicable.

Competing interests
The authors declare no competing interests.

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