

# Three-Dimensional Nonlinear Contact Finite Element Analysis of Mandibular All-on-4 Design

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The All-on-4 design was used successfully for restoring edentulous mandible. This design avoids anatomic cripples such as inferior alveolar nerve by tilting posterior implants. Moreover, tilting posterior implants of All-on-4 design had a mechanical preference than the conventional design. On the other hand, the anterior implants are parallel at the lateral incisor region. Several researches showed favorable results for tilting posterior implants. However, research did not study the influence of the anterior implant position or orientation on the mechanical aspects of this design. This study analyzes the influence of varying anterior implant position and orientation of the All-on-4 design using nonlinear contact 3D finite-element analysis. Three copied 3-dimensional models of the All-on-4 design were classified according to anterior implant position and orientation. The frictional contact between fixtures and bone was the contact type in this finite element analysis. Finally, von Mises stress and strain at implant and bone levels were recorded and analyzed using finite element software. Stress concentrations were detected mainly around the posterior implant at the loaded side. Values of the maximum equivalent stress and strain were around tilted implants of design III followed by design II, then design I. Changing the position or orientation of the anterior implants in All-on-4 design influences stress-strain distribution of the whole design.

**Key Words:** *nonlinear FEA, All-on-4 design, 3D finite element, implant-supported prosthesis*

## INTRODUCTION

Clinicians have been using immediate loading fixed prosthesis successfully for many years to rehabilitate edentulous mandible. However, limited bone volume is frequently found in the posterior region of the mandible; fixed prosthesis is still valid as a line of treatment even without surgical management.<sup>1</sup> Maló et al<sup>2</sup> suggested the use of 4 implants with angled posterior implants in what is called All-on-4 design to solve the problem of reduced bone volume. All-on-4 design not only add mechanical advantage to prosthetic design but also provide simple and cost-effective treatment. Moreover, this technique relied on immediate loading protocol, which is preferable to many patients.<sup>3</sup> All-on-4 implants displayed a high success rate and slight marginal bone resorption (98.1% at 5 years and 94.8% up to 10 years). Therefore, it confirms the use of fixed mandibular full-arch prostheses supported by 4 immediately loaded implants to rehabilitate edentulous mandible.<sup>4</sup>

Despite all of these clinical results, biomechanical questions often arise about load distribution on tilted implants and bone as well as the resulting stress and strain fields.<sup>5</sup> Authors recommended avoiding angled abutments in several articles unless they are needed.<sup>6-8</sup> Clelland et al<sup>7</sup> clarified that angulated implants will subsequently induce more stresses in the design. On the other hand, Zampelis et al<sup>9</sup> detected no difference in cortical bone stress between angled and nonangled implants. Takahashi et al<sup>10</sup> studied the influence of implant number and angulation on stress-strain distribution using 3-dimensional (3D)

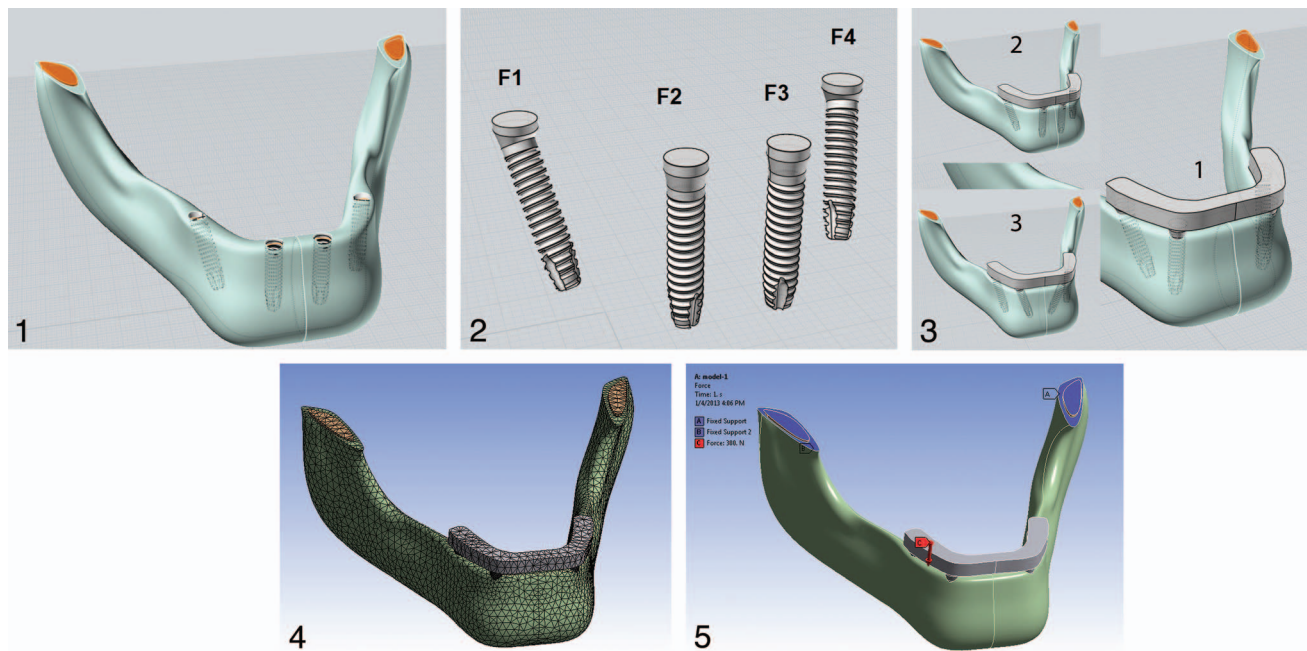
finite element analysis. They concluded that the use of inclined implants in the All-on-4 concept with a short cantilever acts to lessen the peri-implant stress of the cortical bone.

Reflection photoelasticity, mathematical equations, mechanical models, holographic interferometry, and finite element analysis (FEA) are different methods used to evaluate equitable distribution of load and analyze stresses and strains. A 3D finite element analysis has a comparative and descriptive design to evaluate stress distribution.<sup>11</sup> Researchers used finite element analysis extensively to predict the biomechanical performance of various dental implant designs. In addition, this method was valuable to study the effect of clinical reasons on implant success. The principal difficulty in simulating the mechanical behavior of dental implants is to model human bone tissue and its response to applied mechanical force. During analysis, certain assumptions used to be applied to ease modeling and solving process. The complexity of bone structure and its interaction with implant surface have forced authors to make major simplifications.<sup>12</sup>

Recently, several new scanning techniques such as the computerized tomography (CT), cone beam CT, and magnetic resonance imaging enabled producing accurate 3D anatomic models. Combined with these scanning technologies, specific commercial software has been developed to create efficiently finite element meshes. ScanFE (Simpleware), Mimics (Materialise), Amira, and 3D Doctor are some examples that allow processing the images and creating 3D geometric models.<sup>13-16</sup>

The condition of the bone-to-implant contact may, to a certain extent, decide the success of the restorative surgery. Most FEA studies assumed that osseointegration between bone and implant is 100%, and thus bones were bonded to the implant surface. However, this assumption does not appear

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**FIGURES 1–5.** **FIGURE 1.** Isometric view of the 3-dimensional (3D) model of the mandibular bone including cancellous (orange color) and compact bone (light blue color). **FIGURE 2.** 3D models of the 4 implants (angled and paralleled) designed according to manufacturer's dimensions and configurations. Starting from right to left fixture 1 (F1) to fixture 4 (F4). **FIGURE 3.** Assemblies of the 3 designs seen in computer-aided design (CAD) software showing changes in the position and orientation of the anterior implants. **FIGURE 4.** Meshing of the models was performed using 3D tetrahedron element type with 4 node element shape. **FIGURE 5.** Boundary conditions were selected using fixed support constrains (blue areas) and force applied (red arrow and areas).

realistic for the clinical conditions where an imperfect bond could be present, especially in the case of immediate loading. Many existing FEA packages offer various frictional contact algorithms. Nevertheless, experimental recording of the frictional coefficient is necessary to simulate different contact conditions between bone and implants.<sup>17</sup>

Isotropic, transversely isotropic, orthotropic, and anisotropic are different types of material properties used in FEA. An isotropic material describes material properties that are identical in all directions. This is a linear FEA and thus needs only 2 independent constants (Young's modulus  $E$  and Poisson's ratio).<sup>17–22</sup> Naini et al<sup>23</sup> used 3D FEA method to assess the stress-strain distribution of the All-on-4 design through a CT-based 3D model. They applied the oblique isotropic material properties to both cancellous and cortical bone. Further, Fazi et al<sup>24</sup> believed that compact bone is an orthotropic material not a transversely isotropic. On the other hand, some authors<sup>9,10,12,25</sup> preferred using isotropic material properties for both types of bone to ease solving procedures.

The aim of the present study was to examine the influence of different anterior implant positions and angulations of the All-on-4 design on stress-strain distribution. Therefore, values of maximum equivalent stress and strain of implants and bone were observed and analyzed.

#### MATERIALS AND METHODS

The 3D model of the human mandible was created following the general configurations and dimensions of a mandibular CT

scan. Thus, a 64-year-old edentulous patient having healthy alveolar mucosa and well-developed residual ridge was selected. No pathologic conditions were recorded during clinical examination. After examination, a CT scan was obtained (GE Medical System/Bright Speed S, Wauwatosa, Wis). Consequently, all diagnostic procedures confirmed that the patient was a good candidate for the All-on-4 design according to the criteria suggested by Maló et al.<sup>2</sup>

Six CT cut images were selected (at canine areas, angle of the mandible, and ascending ramus bilaterally) then imported into computer-aided design (CAD) software (Mol v2, Triple Squid software design, Seattle, Washington, USA). These images were used as a guide for modeling the mandibular bone. The compact bone was modeled with different thicknesses, ranging from 1 to 2 mm. To simplify the analysis, bones above the mandibular notch were cut off (Figure 1).

A 3D model of a dental implant (Brånemark MKIII Groovy, Nobel Biocare, Göteborg, Sweden) and its abutment was created using the CAD designing software. Both the dimension (4-mm diameter  $\times$  13.5-mm length) and shape of the implant were consistent with the manufacturer's data. To create the specified number of threads and the needed pitch, the body Boolean (software tool) was used. Similarly, this tool designed the cutback parts of the apical region (Figure 2).

As all abutments carried the prosthesis evenly, angled abutments were used to compensate fixture inclinations. The 4 implants of each model were named, for simplification, according to their position from left to right (F1, F2, F3, and F4, respectively).

TABLE 1

Mechanical properties including Young’s modulus and Poisson’s ratio assigned to the model components

	Young’s Modulus, GPa	Poisson’s Ratio
Implant	113.8	0.342
Prosthesis	110	0.3
Compact bone	20	0.3
Cancellous bone	2	0.4

The 3D design of the prosthesis was a simple rod following the arch shape and connecting the implant abutments. This rod included a small cantilever extension distal to the posterior implants.

Three different models of the designs under study were assembled as follows:

- Design I: This model was an All-on-4 design having 2 posterior implants placed just anterior to the foramina and tilted distally about 30° to the occlusal plane. Additionally, 2 parallel implants were placed in the canine area bilaterally (Figure 3).
- Design II: This model included the typical All-on-4 design as suggested by Maló et al.<sup>2</sup> The design included 4 implants: 2 angled implants posteriorly and 2 parallel implants at the lateral incisor areas bilaterally (Figure 3). This group was considered the control group.
- Design III: This model was also an All-on-4 design, but with their anterior implants placed in the canine areas at an angle 20° to the occlusal plane as seen from the labial view (Figure 3).

The model assemblies were transferred as IGS file format from the CAD software to the design modeler module of ANSYS v14 finite element software (ANSYS Inc, Houston, Tex). Using the predefined material library, material properties were assigned to each solid component in the assembly. Regarding material properties assumed, all parts had isotropic behavior.<sup>6,8,9,10</sup> Thus, each part owned 2 values: Young’s modulus and Poisson’s ratio (Table 1). Before meshing the model, areas of refinements were determined at locations of interest. The element type was 3D tetrahedron with 4-node element shape. The total number of elements was about 193 100 with 322 890 nodes (Figure 4). After meshing, boundary conditions were determined to define the relationships between elements. A bonded surface-surface contact was the contact among all assembly parts except at the implant-bone interface. Alternatively, a nonlinear frictional contact, with frictional coefficient

0.3, was a suitable type to simulate immediate loaded implants.<sup>12,17</sup>

A 300 N axial load was applied perpendicular to the left right molar region of the prosthesis. To avoid false stress concentrations, the applied load was directed over a circular area with a diameter of 4 mm and away from the cantilever end by 6.5 mm. The mandible was restrained at the rami ends to limit any movement or rotation in this area (Figure 5).

The stress-strain analysis was performed using equivalent (von Mises) stress and strain. The values of maximum stress and strain, especially for bony tissues and implants, were recorded and analyzed.

RESULTS

The von Mises stress and strain of each part of the 3 design assemblies was collected and tabulated (Table 2 and Figures 6 and 7).

**I: von Mises stress of different components within each design**

*Cancellous Bone*

The maximum equivalent stress was calculated and analyzed in the cancellous bone, especially at the bone-implant contact. The maximum value recorded was 19.14 MPa, which was detected in the distobuccal aspect of the implant F1 of design III as seen in Table 2 and Figure 6. By changing the angulations of the implant F3 and F2 to be parallel as in design I, the value of the maximum stress was reduced to 15.89 MPa. The value of the maximum stress was also decreased to 17.84 MPa by changing implant F3 and F2 positions as in design II (Figure 8).

*Compact Bone*

Maximum equivalent stress in the compact bone showed the highest value at the ascending ramus of the design III (68.79 MPa) followed by design I, where slight reduction of the stress (68.56 MPa) was elicited. The maximum value of the equivalent stress of the second design showed a decrease of stress value to 63.89 MPa (Figure 9).

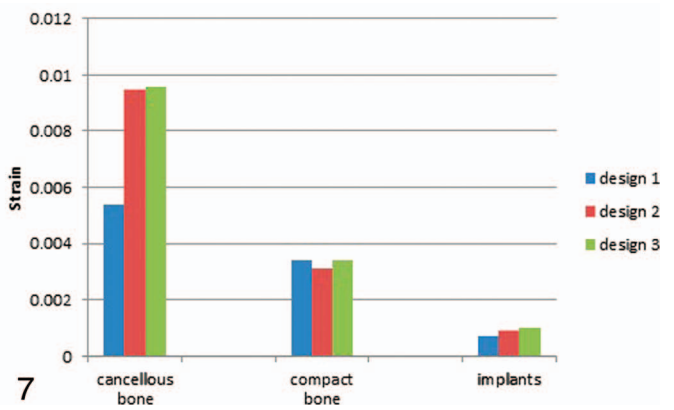
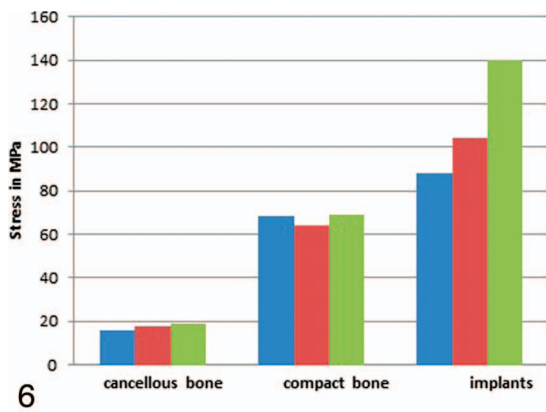
*Implants*

The stress distributions at the implant level could be followed as shown in Figure 10. A marked area of stress concentration was noticed at the implant-prosthesis connection of the F2 implant of the design III (140.01 MPa). Design II showed maximum equivalent stress at the distal surface of F1 implant (104.22 MPa). In the case of design I, stress concentration was recognized in the mesiobuccal aspect of the F1 implant-

TABLE 2

Maximum equivalent (von Mises) stress and strain of the different design components including cancellous bone, compact bone and implants

	Stress	Strain	Stress	Strain	Stress	Strain
I	15.89	$5.4 \times 10^{-3}$	68.56	$3.4 \times 10^{-3}$	88.38	$7 \times 10^{-4}$
II	17.84	$9.5 \times 10^{-3}$	63.89	$3.1 \times 10^{-3}$	104.22	$9 \times 10^{-4}$
III	19.14	$9.6 \times 10^{-3}$	68.79	$3.4 \times 10^{-3}$	140.01	$1 \times 10^{-3}$



FIGURES 6 AND 7. FIGURE 6. Graph showing maximum equivalent stress of components of different designs. FIGURE 7. Graph showing maximum equivalent strain of components of different designs.

prosthesis connection with reduction in the stress value (88.38 MPa).

**II: von Mises strain of different components within each design**

*Cancellous Bone*

Maximum equivalent strains in cancellous bone were recognized in the distal surface of the implant F1 with the highest value recorded in design III ( $9.6 \times 10^{-3}$ ) followed by a slight decrease in design II ( $9.5 \times 10^{-3}$ ). The value of the maximum equivalent strain was reduced dramatically in the design I ( $5.4 \times 10^{-3}$ ) (Table 2 and Figure 11).

*Compact Bone*

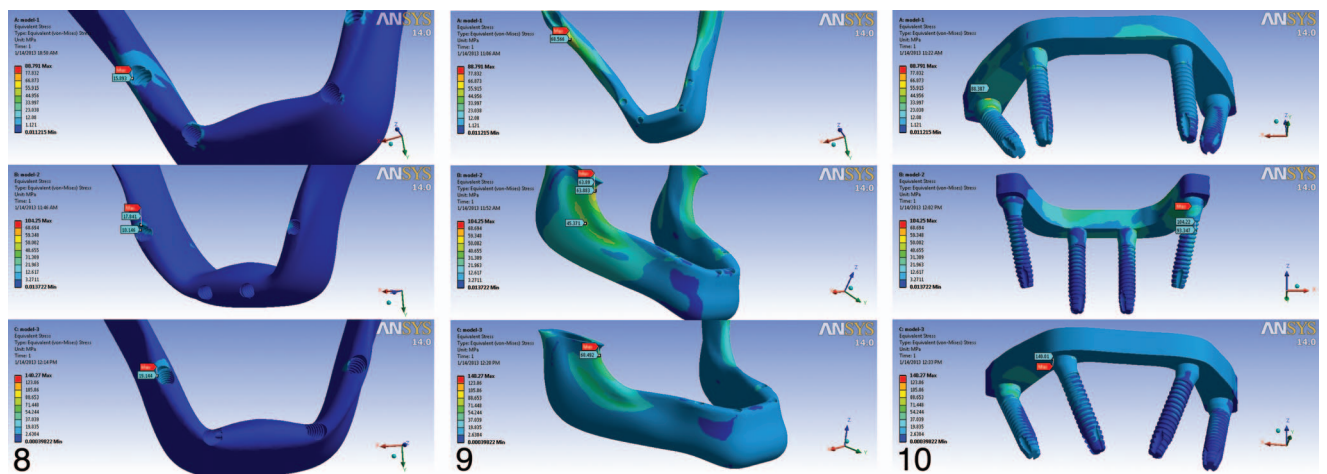
The value of the maximum strain recorded in the compact bone at the ascending ramus with values ranged between  $3.4 \times 10^{-3}$  and  $3.1 \times 10^{-3}$ . Both design I and III showed nearly equal values; however, design II showed reduced value of strain ( $3.1 \times 10^{-3}$ ) (Figure 12).

*Implants*

Maximum equivalent strain was recognized in the same positions of the maximum equivalent stress for all designs. The values of maximum equivalent strain were  $7 \times 10^{-4}$ ,  $9 \times 10^{-4}$ , and  $1 \times 10^{-3}$  for design I, II, and III, respectively (Figure 13).

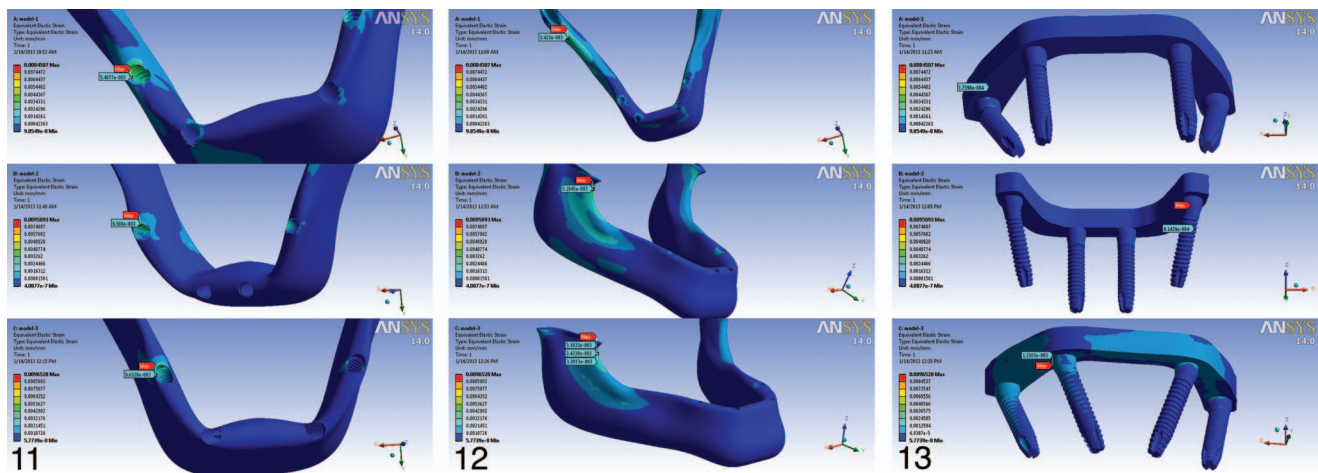
**DISCUSSION**

There is a correlation between biomechanics of bone tissues and masticatory force amount and direction. This force could induce or reduce bone formation. Whenever possible, prosthetic designs aim to perform favorable conditions that keep the force within the physiologic limit of the supporting tissues. Stress transferred from implants to surrounding bones depends on the load direction, the bone-implant interface, and implant size. Further, stress not only was influenced by implant surface treatment and shape, but also with the prosthesis type and the surrounding bone quantity and quality.<sup>12</sup>



FIGURES 8–10. FIGURE 8. Equivalent stress in the cancellous bone in 3 different designs, from top to bottom (design I, II, and III, respectively). FIGURE 9. Equivalent stress in the compact bone in 3 different designs, from top to bottom (design I, II, and III, respectively). FIGURE 10. Equivalent stress in the implants in 3 different designs, from top to bottom (design I, II, and III, respectively).





**FIGURES 11–13.** **FIGURE 11.** Equivalent strain generated in cancellous bone in 3 different designs, from top to bottom (design I, II, and III, respectively). **FIGURE 12.** Equivalent strain generated in compact bone in 3 different designs, from top to bottom (design I, II, and III, respectively). **FIGURE 13.** Equivalent strain generated in implants in 3 different designs, from top to bottom (design I, II, and III, respectively).

Maló et al<sup>2</sup> suggested All-on-4 design to restore an edentulous mandible in 2003. When implants supported by this design were evaluated after 10 years, it shows a 95% survival rate.<sup>4</sup> Several authors studied this design to explain and clarify the key to success of this design from the biomechanical aspect.<sup>9,10,23,24</sup> Logically, implant inclinations in any direction will lead to change in the stress distribution, and may lead to stress concentration. This is true for a single implant, so clinicians avoid inclination, unless other controlling factors impose this.<sup>7,8</sup>

In the present study, bones at the posterior implant (F1) display the highest value of stress compared with other bony areas around other implants. These results agreed with those of Naini et al<sup>23</sup> who studied the stress-strain analysis in All-on-4 design. They realized an increase in the stress values by 9% in bone surrounding posteriorly inclined implants. The current results are also in harmony with the findings of Weinberg and Kruger<sup>26</sup> regarding clinical variants affecting implant loading. They emphasized that for every 10° increase in implant inclination, there is approximately a 5% increase in torque. Moreover, this finding was also confirmed with Kim et al.<sup>27</sup> The great difference between Young's moduli of bone and prosthesis also plays a role to achieve these results. In addition, load that was applied between implant F1 and F2 changed the stress-strain distribution surrounding the implants. Such load was transmitted to 3 implants (F2, F3, and F4) splinted together from one side and only the angled implant F1 from the other side. This may lead to increased stress surrounding this implant. In addition, stress produced from implant F1 position was superimposed by its inclination, leading to increase in lateral vector of force. As a result, the distal aspect of bone around this implant showed higher stress values.

Regarding design I, the anterior 2 implants were placed in the canine area. The results confirmed the hypothesis that placing the anterior implants at a strategic location in the corner of the arch may reduce stress level. The stress and strain values were the least between all other designs. Placing the implants in this position may produce a favorable stress distribution through minimizing the distance between the

anterior implant and the applied load, if compared with design II (typical All-on-4 design). The third design had 2 inclined posterior implants combined with 2 inclined anterior implants. This design showed highest values of both stress and strain at the bone-implant interface among different designs. Unfortunately, anterior implant inclination complicated stress conditions and suppressed benefits from placing implants in the canine area. This could be recognized as a stress concentration at implant F2 with a total stress value of 140 MPa compared to other designs. Thus, the assumption of using inclined implant anteriorly to enable using longer implants (similar to posterior implants of typical All-on-4 design) was unsound. The generated stress looks greater than any compensation gained by proposed lengthening. Therefore, it should be used with caution to avoid stress concentrations.

On the other hand, the present study disagreed with Fazi et al<sup>24</sup> and Takahashi et al.<sup>10</sup> They encouraged the use of inclined posterior implants in All-on-4 design. They observed a preferable effect on reducing the stress value when applied force was on the cantilever area. This stress decreased (17%) by placing tilted implants 34° to the occlusal plane. They also realized that when the implant angulation was doubled, the stress value was further decreased by 8%.<sup>24</sup> These studies relied on applying the force at the cantilever end; thus, the cantilever length was playing a role in stress distribution. Therefore, distal inclination of the posterior implants acted to minimize cantilever, and so decreased the stress values. The main difference between these studies and the present study was the load position. These findings confirmed the role of load position on stress-strain distribution.

Nonlinear frictional contact was the contact of choice at the implant-bone surface. On the other hand, bonded contact was used for contact between other parts. Several authors preferred using bonded contact because it simplifies solving the finite element problem, lessens solving time, and minimizes exhaustion of the hardware resources.<sup>10,11,23,24</sup> In addition, researchers used bonded contact to represent 100% osseointegration with no gaps or change in the bone quality at bone-implant interface. However, in the present study, immediately loaded

implants were assumed to represent the All-on-4 design or its variations.<sup>2</sup> Accordingly, the use of the bonded contact was not suitable to represent immediate loading. Several researchers<sup>12,28,29</sup> used the frictional contact at the bone-implant interface with a frictional coefficient  $\mu = 0.3$ , where 0 displays no friction, and 1 represents a significant amount of friction. Although a slight change in the coefficient of friction will not affect the results, the shift from bonded to frictional contact affects the stress-strain distribution.<sup>30</sup>

Posterior unilateral force was selected because it is a frequent chewing position during mastication. To avoid local stress concentration on point or line, the force was applied on 4-mm area diameter.<sup>23</sup> Instead of using the inferior border of the mandible, the restraint was set to the terminal end of the mandibular rami. This position of restraint enabled mandibular flexion during loading.<sup>24</sup>

There are certain simplifications rather than limitations used to solve the present finite element problems. The rami above the mandibular notch were removed, and thus the simulation of the muscular attachment was ignored.<sup>5,18</sup> The material properties of the bone were isotropic and homogeneous in all directions.<sup>10</sup> Although the applied load was static, a dynamic load could be applied in future studies to simulate real masticatory cycles.<sup>22</sup> In addition, a clinical investigation of the modified designs should be considered to validate this stress-strain analysis study.

### CONCLUSION

Within the limitation of this study, changing either anterior implant positions or angulations affects the stress-strain distribution of the mandibular All-on-4 design. Although All-on-4 is an integrated and well-structured design for mandibular fixed prosthesis, shifting the anterior implants toward canine areas showed more favorable stress and strain values. In addition, anterior implant angulations induced more stress in the studied design and should be avoided.

### ABBREVIATIONS

CAD: computer-aided design  
CT: computerized tomography  
FEA: finite element analysis

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