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Effects of Prosthetic Material and Framework Design on Stress Distribution in Dental Implants and Peripheral Bone: A Three-Dimensional Finite Element Analysis

Authors' Contribution:
Study Design A
Data Collection B
Statistical Analysis C
Data Interpretation D
Manuscript Preparation E
Literature Search F
Funds Collection G

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Background: The purpose of this study was to evaluate the effects of prosthetic material and framework design on the stress within dental implants and peripheral bone using finite element analysis (FEA).

Material/Methods: A mandibular implant-supported fixed dental prosthesis with different prosthetic materials [cobalt-chromium-supported ceramic (C), zirconia-supported ceramic (Z), and zirconia-reinforced polymethyl methacrylate (ZRPMMA)-supported resin (ZP)] and different connector widths (2, 3, and 4 mm) within the framework were used to evaluate stress via FEA under oblique loading conditions. Maximum principal (σmax), minimum principal (σmin), and von Mises (σvM) stress values were obtained.

Results: Minimum stress values were observed in the model with a 2-mm connector width for C and ZP. The models with 3-mm and 4-mm connector widths showed higher stress values than the model with a 2-mm connector width for C (48–50%) and ZP (50–52%). Similar stress values were observed in the 3- and 4-mm models. There was no significant difference in the amount of stress with Z regardless of connector width. The Z and ZP models showed similar stress values in the 3- and 4-mm models and higher stress values than in the C model. Z, ZP, and C showed the highest stress values for the model with a 2-mm connector width.

Conclusions: Changes in the material and width of connectors may influence stress on cortical bone, cancellous bone, and implants. C was associated with the lowest stress values. Higher maximum and minimum principal stress values were seen in cortical bone compared to cancellous bone.

MeSH Keywords: **Chromium Alloys • Dental Implants • Finite Element Analysis • Resins, Synthetic • Zirconium**

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Background

Dental-implant-supported dental restorations are common clinical approaches to edentulism cases because of their high success rates [1–4] and their biological and biomechanical advantages, such as preservation of adjacent and opposite teeth, simulation of supporting bone, and production of higher mastication force compared to removable prostheses [3,5]. Dental implants can fail because of biological or biomechanical factors [3,6,7]. To maintain high success rates after loading, biomechanical factors – such as the design and material of the superstructure [5,8], occlusal forces, occlusion, type of abutment connection, number of dental implant supports, and density of the supporting bone – should be considered [8–11].

Good planning and application of a prosthesis is essential to prevent bone and dental implants from excessive, unnecessary forces [11]. Because of the high success rates of all-ceramic crowns [12], all-ceramic systems for three-unit fixed dental prostheses (FDPs) may become a viable treatment option [13]. These systems must achieve the biomechanical requirements of restorations, provide longevity similar to metal-ceramic restorations, and feature enhanced aesthetics [13,14]. Recently, zirconia-supported ceramic restorations were proposed as an alternative to metal-supported ceramic in posterior FDP [5,6,15]. Unfortunately, because of its high-elasticity modulus, cracks may occur under high-mastication force. Chipping of the ceramic veneer is the most common type of failure in these restorations [12,16,17]. In spite of the increase in the use of all-ceramic restorative systems, metal-ceramic systems continue to be used because of their clinical longevity and biocompatibility. Metal-ceramic FDPs are advantageous because of their predictable structural performance, versatility, and low cost [14]. Recently, resin-based materials have been increasingly used in dental practices because of their desirable properties, such as their aesthetic appearance, ease of repair, affordability, and low-elastic modulus (similar to that of dentin). However, not only the composition of the materials used, but also the design, can affect the success of the restoration in terms of stress distribution and magnitude [16]. According to Möllers et al. [16], the framework design and material properties of the superstructure play a significant role in stress distribution. Another study [17] also concluded that stress distribution was affected by the type of ceramic used for the infrastructure.

Clinical studies have provided reliable information regarding the longevity and failure rate of dental restorations, but they are not easy to carry out [15,17]. FEA is an easy and inexpensive way to evaluate the mechanical behavior of complex structures [5,6,18,19]. Therefore, it is suitable for this study [1,15,20–22].

The present study evaluated the effects of prosthetic material and framework design on the biomechanical behavior of a posterior dental-implant-supported three-unit fixed partial denture using three-dimensional FEA. The first null hypothesis was that the material of the prosthesis would not affect the stress, and the second null hypothesis was that the change in the dimensions of the connector area of a three-unit restoration would not affect the stress within the dental implants and peripheral bone.

Material and Methods

Three-dimensional finite element models (3D FEMs) were constructed, homogenized, and meshed using a computer (Intel Xeon® R CPU 3.30-GHz processor, 500 GB hard disk, 14 GB RAM, Windows 7 Ultimate Version Service Pack 1), 3D scanner [Activity 880 (Smart Optics Sensortechnik GmbH, Bochum, Germany)], and software, including Rhinoceros 4.0 (Robert McNeel & Associates, Seattle, WA, USA), 3D-Doctor (Able Software Corp., Lexington, MA, USA), VRMesh (VirtualGrid, Bellevue, WA, USA), and Algor Fempro (ALGOR, Inc., Pittsburgh, PA, USA).

Construction of 3D FEMs

3D-Doctor and Rhinoceros were used to generate a mandibular model using data from a 1-mm slice of human cadaver obtained using computerized tomography (3M Imtec Corporation, Ardmore, OK, USA).

Dental implants, abutments, and gypsum anatomical crown models were scanned with an Activity 880 scanner in macro mode to generate 3D FEMs. Two standard dental implant 3D FEMs with a diameter of 4 mm and a length of 10 mm were located in the mandibular first premolar and first molar regions with a distance of 15.5 mm from the center of the dental implants. Dental 3D FEMs were reformed according to the Wheeler Dental Anatomy Atlas using Rhinoceros, and a three-unit fixed partial restoration with a modified ridge-lap second premolar pontic was created (Figure 1). A veneered restoration design was standard for all models. Three different thicknesses (2, 3, and 4 mm) at the connector area of the framework were designed via Boolean processes.

Models

Three different restorations were generated with different frameworks and connector widths of 2, 3, and 4 mm (Figure 1). Three different materials [ceramic veneered cobalt-chromium alloy (C), ceramic veneered zirconium dioxide (Z), and composite veneer (ZP)] were used as superstructure materials.

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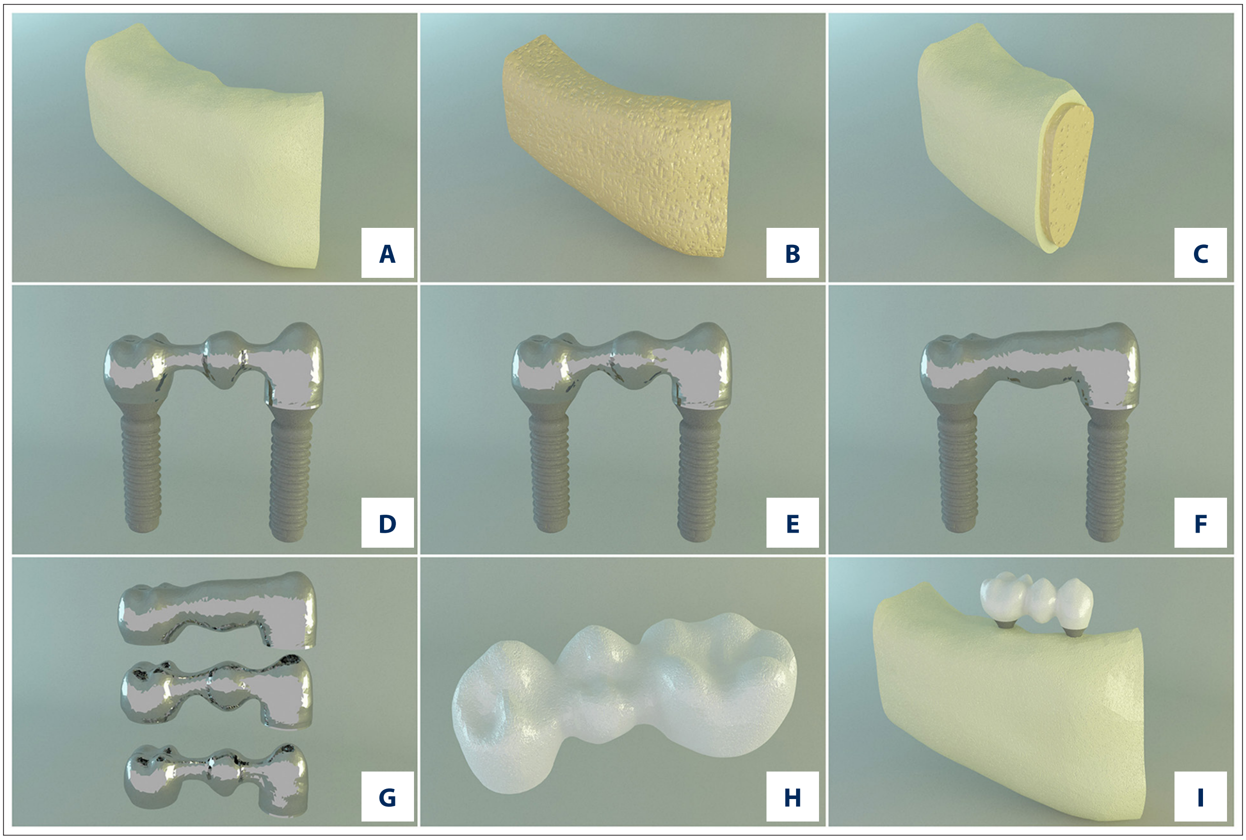

Figure 1. (**A**) Cortical bone model. (**B**) Cancellous bone model. (**C**) Bone model. (**D**) Framework design with 2-mm connector width. (**E**) Framework design with 3-mm connector width. (**F**) Framework design with 4-mm connector width. (**G**) Different framework models. (**H**) Veneering model. (**I**) Final model.

Table 1. Material properties.

Material	Elasticity modulus (GPa) (*E*)	Poisson rate (υ)
Cancellous bone	1.37	0.3
Cortical bone	13.7	0.3
Titanium alloy (Ti-6Al-4V)	110.0	0.35
Cobalt-chromium alloy	218.0	0.33
Zirconia	269.0	0.25
Zirconia-reinforced polymethyl methacrylate	3.05	0.3
Feldspathic porcelain	61.2	0.19
Veneering composite (Variolink occlusal)	10.0	0.3

Mesh creation

Fempro was used to generate the nodal points and meshes of the models. Ten-noded brick-type elements were used as frequently as possible. The number of nodes and elements in each model is shown in Table 1. The properties of the materials used in this study were derived from the literature.

Loading

The base and the anterior and posterior edges of the mandible were fixed in all directions with zero displacement (Figure 2). The models featured linear elastic characteristics. The materials were assumed to be homogenous and isotropic. The dental implants were considered to be fully osseointegrated. Perfect fit was assumed for the bone, dental implants, abutment, and restoration.

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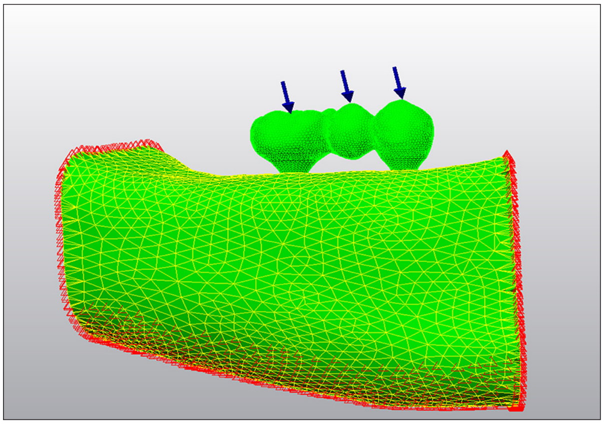

Figure 2. Meshed model.

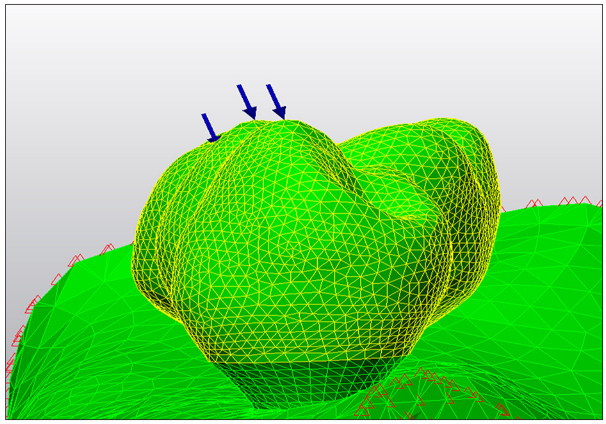

Figure 3. Loading condition.

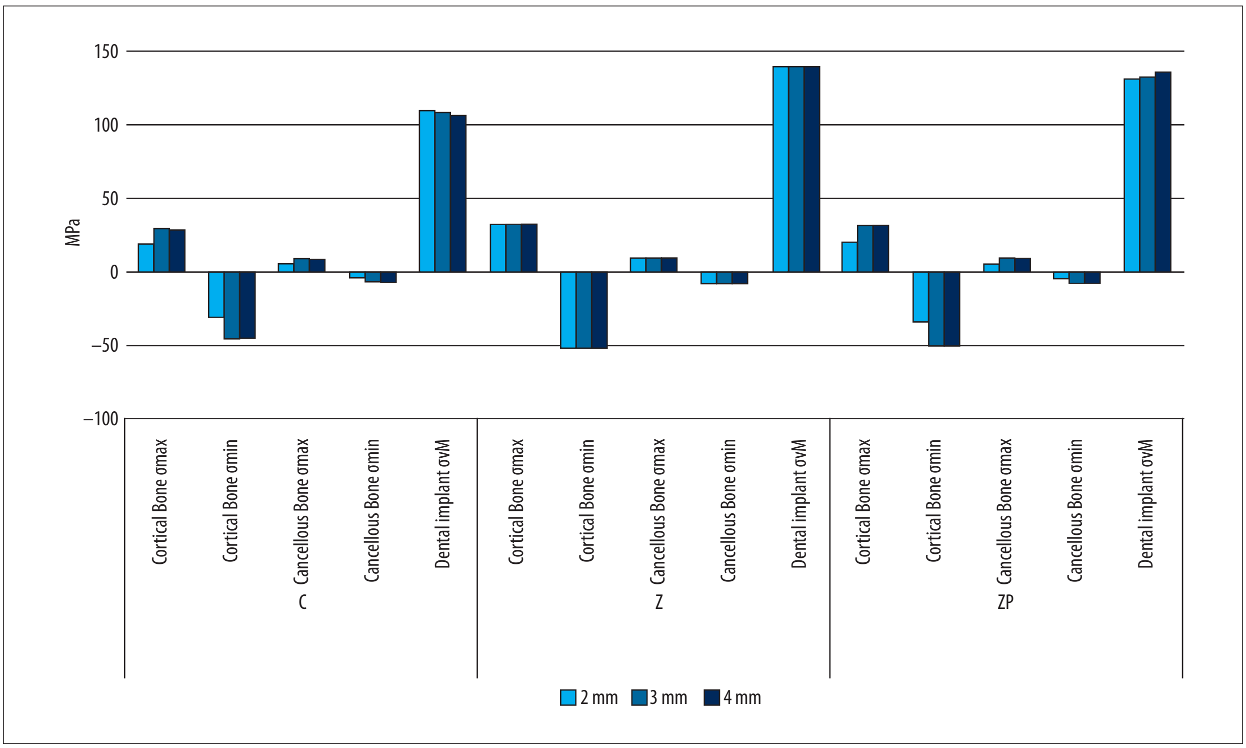

Figure 4. Graphical illustration of results.

To evaluate and compare the stresses, 300 N oblique (30°) loads were applied to the model. In total, 3 models and 9 scenarios were generated with different materials and connector widths (Figure 3).

The results of the mathematical solutions were converted into visual results. Maximum principal stress values (σmax), minimum principal stress values (σmin), and von Mises stress values (σvM) were obtained. The difference in the values exceeding 5% were considered as important.

Results

The results of the von Mises stress analysis and the maximum principal and minimum principal stress analysis are presented in Figures 4–9. The positive values in the illustration represent σmax for bone and σvM for dental implants, whereas negative values represent σmin for bone. The σmax values were in the range of 20–33 MPa for cortical bone and 6–10 MPa for cancellous bone. The σmin values were between –31 and –52 for cortical bone and –4 and –8 MPa for cancellous bone. The σvM values for dental implants varied from 107 to 140 MPa.

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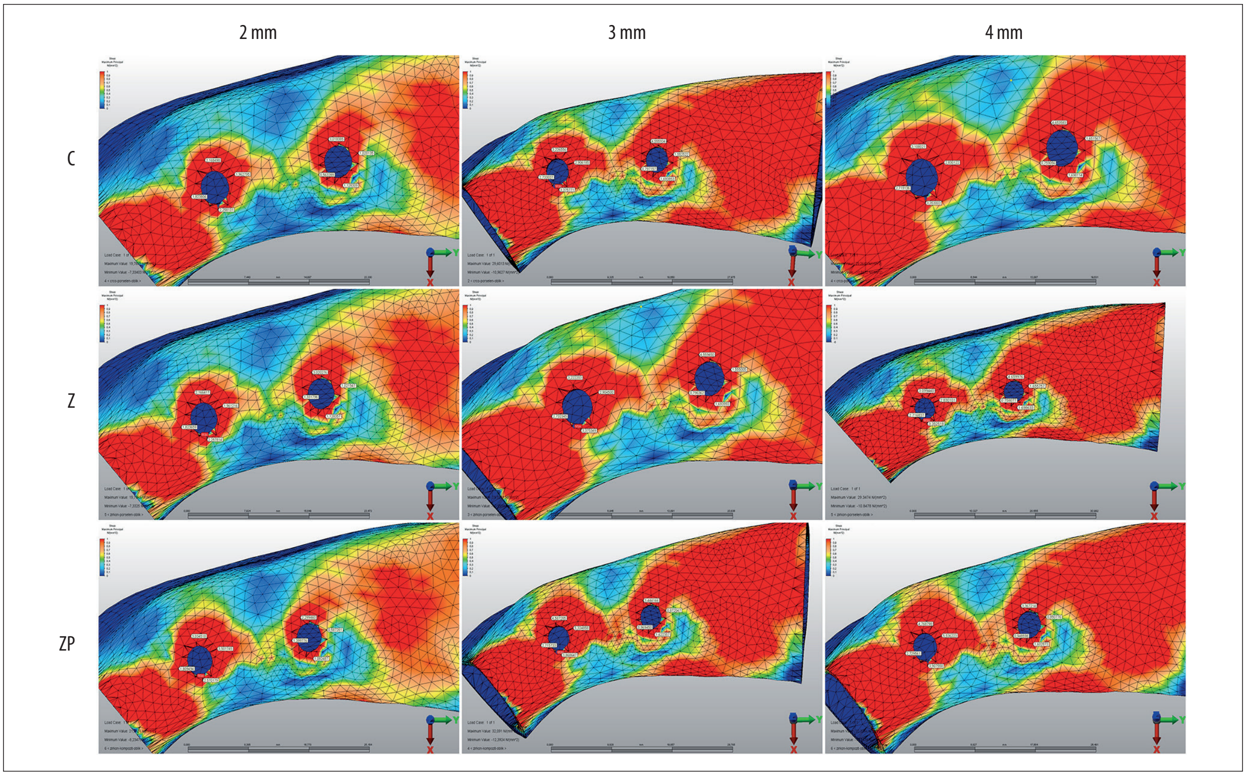

Figure 5. Maximum principal stress results in cortical bone.

Effect of prosthetic material

Prosthetic material affected the stress values in the peripheral bone. The stress values in the dental implants were similar.

Models with 2-mm connector width

C showed the lowest σmax and σmin values for cortical (20 to 30 MPa for σmax and –31 to –46 MPa for σmin) and cancellous bone (6 to 9 MPa for σmax and –4 to –7 MPa for σmin) and the lowest σvM stress value for dental implants (107 to 109 MPa). Z had the highest stress values for all structures.

Models with 3-mm and 4-mm connector widths

C showed the lowest σmax and σmin values for cortical and cancellous bone and the lowest σvM stress value for dental implants. Z and ZP had similar stress values for all structures and higher stress values than C.

Effect of connector width

The σvM values of dental implants were similar in all models, regardless of connector width. A change in the width of the connector did not affect stress in Z models. The σmax and σmin values for cortical bone were similar in the models with 3- and 4-mm connector widths, and were higher than the 2-mm model for C (50% and 48% higher for σmax and 50% and 49% higher for σmin, respectively) and for ZP (52% and 50% higher for σmax and 51% and 52% higher for σmin, respectively). The σmax and σmin stress values for cancellous bone were similar in the models, with 3- and 4-mm connector widths, and were higher than in the 2-mm model for C (50% and 49% higher for σmax and 50% and 52% higher for σmin, respectively) and ZP (51% and 52% higher for σmax and 50% and 52% higher for σmin, respectively).

Discussion

The first null hypothesis of this study was rejected based on the results. C models showed the lowest stress values in dental implants and peripheral bone independent of connector width. According to some studies [1,5,15–17] the stiffer a material, the more load it will attract. Similarly, in the present study, the stiffest prosthetic material (Z) attracted more stress, and the most elastic prosthetic material (ZP) attracted the least stress; however, they transferred the loads independently of the stiffness. According to a study [23], the ceramic veneer materials that have similar elastic modulus values to enamel supported the enamel better than resin composites. In the present study, titanium dental implant abutments were used, and the least stress values in the dental implant-abutment complex were observed in C models, which had more material properties similar

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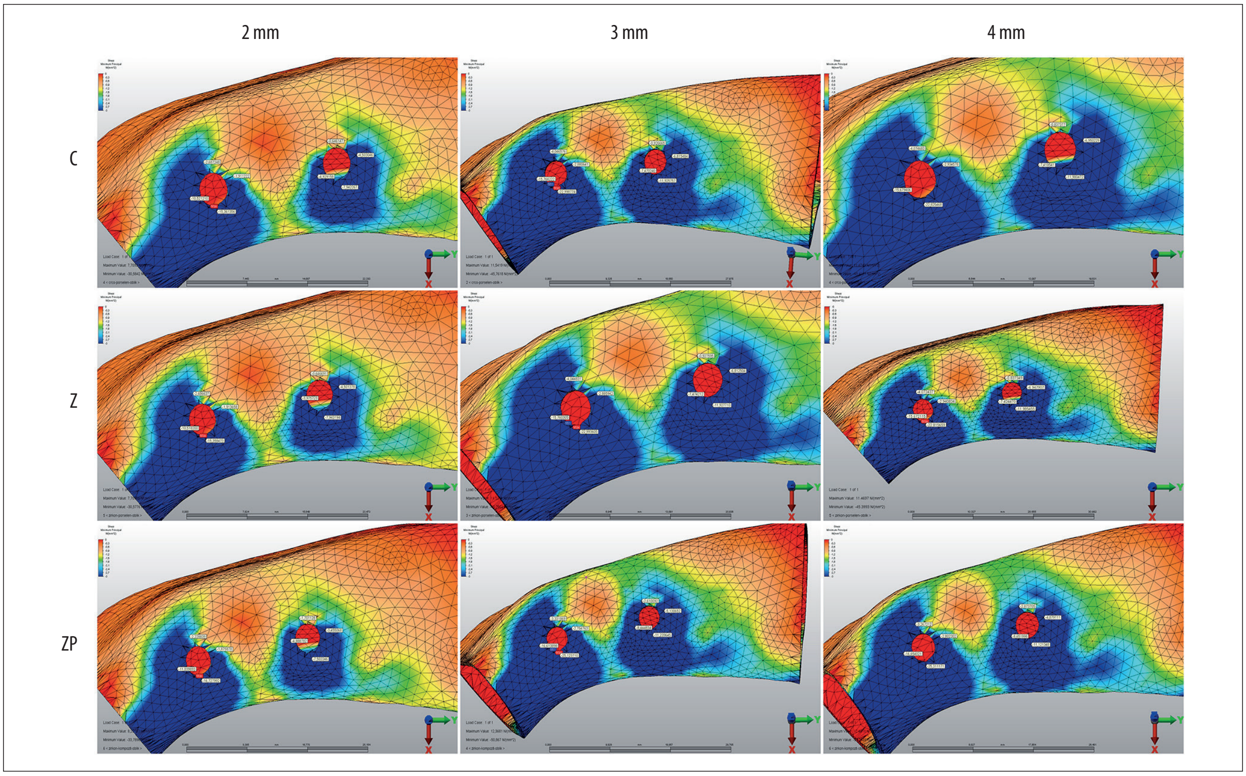

Figure 6. Minimum principal stress results in cortical bone.

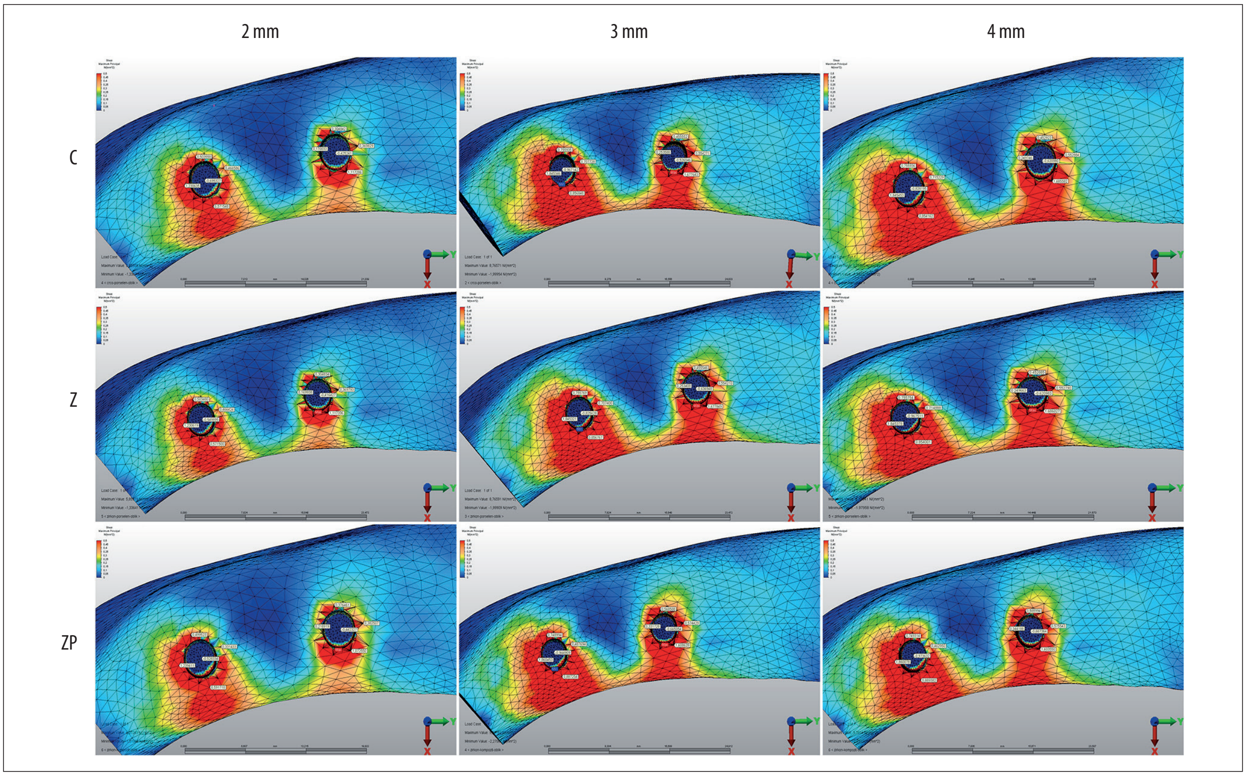

Figure 7. Maximum principal stress results in cancellous bone.

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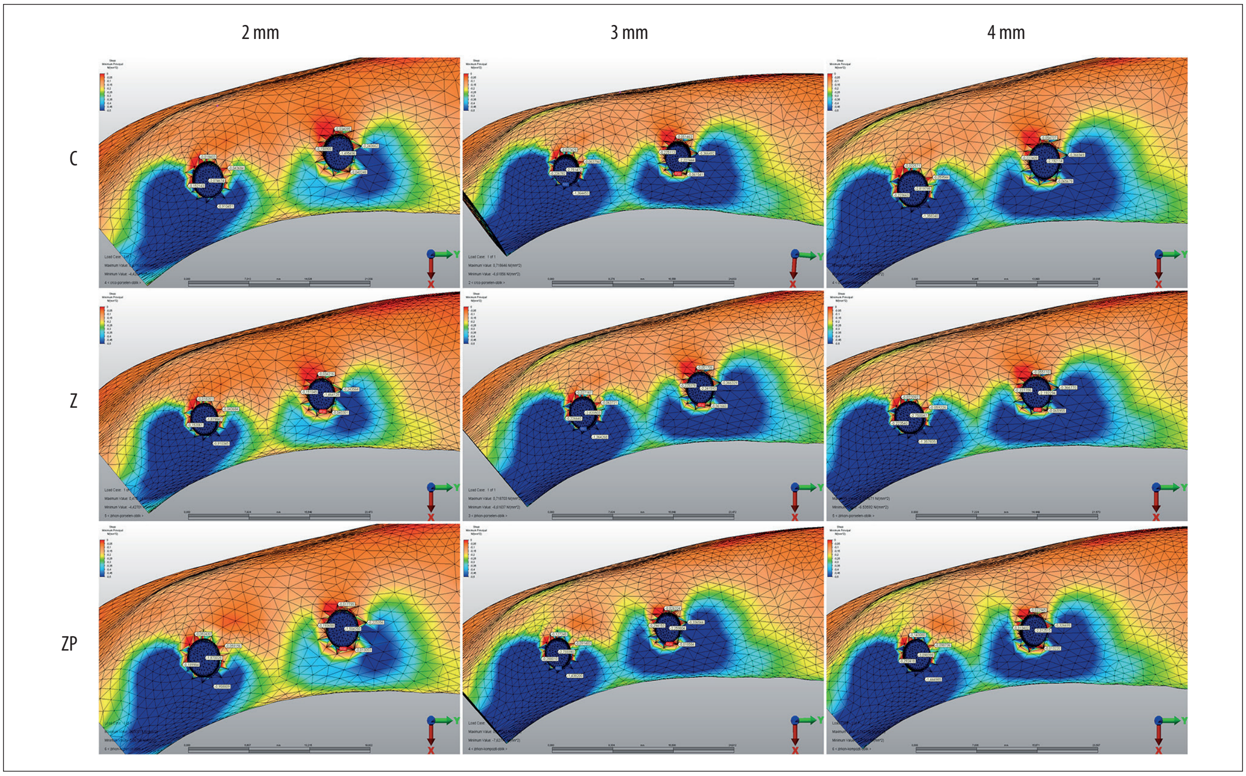

Figure 8. Minimum principal stress results in cancellous bone.

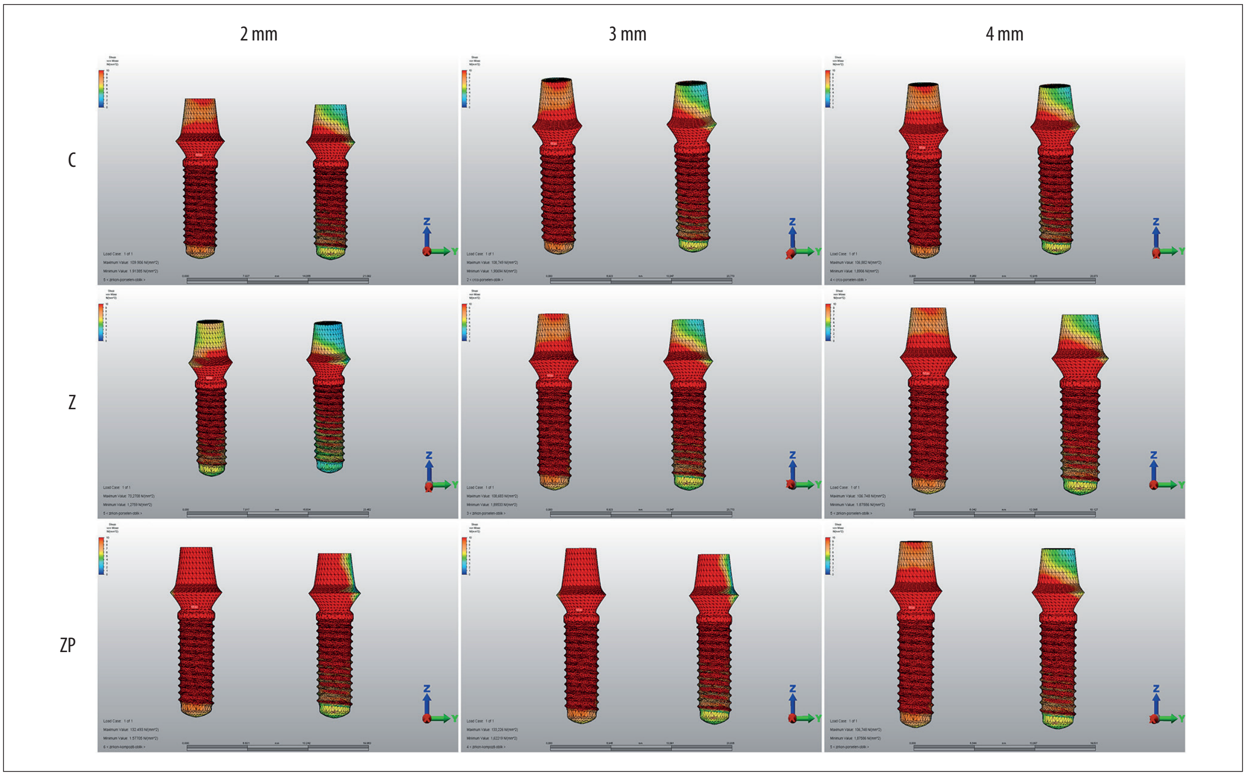

Figure 9. von Mises stress results in implant-abutment complex.

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Table 2. Number of elements and nodes.

	2 mm connector width	3 mm connector width	4 mm connector width
Number of elements	258599	256692	253599
Number of nodes	48717	48218	47977

to titanium in comparison with other materials used. It may be concluded that the stress distribution is affected not only by the material properties, but also by the harmony between the material properties supporting structures and the materials used.

The second null hypothesis was partially rejected based on the differences in the stress of peripheral bone. The lowest stress values in peripheral bone were observed with 2-mm connector models for C and ZP. In the Z model, the second null hypothesis was accepted. The reason behind it might be because the high elastic modulus of Z could make it more rigid in the low dimensions of the connector width.

The shape of an FDP is not uniform; its contour features multiple convexities and concavities [24]. Additionally, the dimensions of the connectors have major impacts on the stress concentration in FDP [21], and the design of the connectors affects the FDP's fracture resistance [13]. The failure rate of three-unit ceramic FDPs around connector areas has been reported to be relatively high [13,14]. Thus, the lifetime of bridge restorations can be significantly increased by improving the design in the connector area [21]. Johanson et al. [25] pointed out that the vertical dimensions of connectors were greater in the anterior region (mean 4.4 mm) than in the posterior region (mean 3.6 mm) in their study. Ridwaan et al. [26] found lower values in the posterior region (2.7–2.9 mm). According to Bahat et al. [27], the recommended dimensions for connectors of zirconia vary from 2 to 4 mm in occluso-gingival height and from 2 to 4 mm in bucco-lingual width. In the present study, 3 different connector widths and heights were used in these ranges (2, 3, and 4 mm).

The highest tensile strength of the cortical bone was reported as 121 MPa, and the maximum compression strength was reported as 167 MPa [28]. In this study, the highest tensile (σmax= 33 MPa in Z model) and compression (σmin=52 MPa in Z model) stress values were lower than the ultimate strength values of cortical bone.

There should be at least 30 000 elements and 200 000 nodes [1]. The numbers of the elements and nodes in this study are given in Table 2.

Similar to other research [2,15,22], we determined that stress was concentrated at the cortical bone around the neck of dental implants. The probable reason for this is that the difference in the elastic modulus of the cortical and cancellous bone, and the rotational center in the dental implant, is at the cortical bone level [29].

The maximum occlusal force during mastication varies in natural dentition and dental implants because of muscle size, bone shape, the temporomandibular joint tissues, and the amount of jaw separation [30,31]. Furthermore, the change in bite direction and the sex of the patient affect the maximum bite force [31]. When applying FEA, the loading conditions are important factors [32]. Some studies suggested that oblique loading was associated with realistic loading [1,22]. In the present study, the nodal points of load application were on the buccal (functional) cusp for oblique loads (300 N) with 30°.

The outcomes of this study must be evaluated within the limitations of the FEA approach. All materials in this study were assumed to be homogenous, isotropic, and linearly elastic. Three-dimensional data from a mandible were used to generate the bone model; however, cortical bone was assumed to be in the same thickness all over the cancellous bone. The occlusal forces were applied from one point and in one direction. In addition, dental implants were assumed to be 100% osseointegrated. The material properties and thicknesses of dental cements were not included in this study. Abutment and dental implants were created as one body. The dental implant material, diameter, length angulation, and surface treatment of dental implants can affect the stress in peripheral bone; however, these variables were constant in the present study [30,33–36]. Nonetheless, FE models and analyses are only an approximation of the clinical situation. Further studies are needed to better understand the biomechanical results of different designs and materials of the prosthesis supported by dental implants.

Conclusions

Within the limitations of this FEA study, it is concluded that the changes in prosthetic material affected the stress within the dental implant-abutment complex and peripheral bone. The changes in connector width may affect the stress distribution within the peripheral bone in more elastic materials. The lowest stress values in peripheral bone occurred in the cobalt-chromium-supported ceramic model and the zirconia-reinforced polymethyl methacrylate model with a 2-mm connector width. In the zirconia-ceramic model, the dimensions of the connector did not affect the stress in dental implants and

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peripheral bone. Cortical bone showed higher stress values than cancellous bone, regardless of material and connector design.

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Conflicts of interest

None.

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