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Biomechanical investigation of maxillary implant-supported full-arch prostheses produced with different framework materials: a finite elements study

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PURPOSE. Four and six implant-supported fixed full-arch prostheses with various framework materials were assessed under different loading conditions. **MATERIALS AND METHODS.** In the edentulous maxilla, the implants were positioned in a configuration of four to six implant modalities. CoCr, Ti, ZrO_2, and PEEK materials were used to produce the prosthetic structure. Using finite element stress analysis, the first molar was subjected to a 200 N axial and 45° oblique force. Stresses were measured on the bone, implants, abutment screw, abutment, and prosthetic screw. The Von Mises, maximum, and minimum principal stress values were calculated and compared. **RESULTS.** The maximum and minimum principal stresses in bone were determined as $\text{CoCr} < \text{ZrO}_2 < \text{Ti} < \text{PEEK}$. The Von Mises stresses on the implant, implant screw, abutment, and prosthetic screws were determined as $\text{CoCr} < \text{ZrO}_2 < \text{Ti} < \text{PEEK}$. The highest Von Mises stress was 9584.4 Mpa in PEEK material on the prosthetic screw under 4 implant-oblique loading. The highest maximum principal stress value in bone was found to be 120.89 Mpa, for PEEK in 4 implant-oblique loading. **CONCLUSION.** For four and six implant-supported structures, and depending on the loading condition, the system accumulated different stresses. The distribution of stress was reduced in materials with a high elastic modulus. When choosing materials for implant-supported fixed prostheses, it is essential to consider both the number of implants and the mechanical and physical attributes of the framework material. [J Adv Prosthodont 2022;14:346-59]

KEYWORDS
Biomechanics; Finite element analysis; Polyetheretherketone; Zirconia; Titanium

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INTRODUCTION

Tooth loss is an irreversible condition that severely impacts an individual's overall quality of life. Edentulism leads to the loss of function as well as physical, mental, and social impairments for the patient.[1] Implant therapy has superceded traditional treatments as the treatment of choice for the replacement of single and multiple tooth loss. Full mouth implant treatments with fixed prostheses are especially popular among edentulous patients and have become common in contemporary dentistry.[2,3]

A concept of four implant-supported fixed prostheses, often known as 'All-on-4', developed by Malo *et al*., is an alternative to Dr. Misch's conventional fixed implant treatments.[4,5] By deliberately placing a limited number of implants and avoiding complicated surgical procedures, this method aimed to save recovery time and expenses (such as bone grafting and maxillary sinus augmentation). Two axial implants and two distally tilted (30° - 45°) implants are positioned anteriorly and posteriorly, respectively.[5-7] In addition, a modest number of implants may be used to provide prosthetic rehabilitation for patients who are unable to get fixed prosthetic therapy. Despite research comparing the biomechanics of full-arch prosthesis supported by four implants, the lack of studies comparing six- and four-implant modalities over a wide range of materials is significant.[8-13] Previous research has shown that designs with 4-6 implants provide the desired functional, biological, and cosmetic outcomes.[14,15] However, they must be complemented with other materials.

For optimal clinical performance, dental materials must possess the necessary physical, chemical, and biological characteristics. The material for the permanent restoration must be chosen depending on the area being restored, aesthetic standards, the dentist's suggestions, and the patient's budget. For implant-supported fixed prostheses, there are a variety of framework material alternatives. Resins, polymeric materials, metal alloys, and ceramics are commonly used in therapeutic applications due to their aesthetic and mechanical properties. When choosing a material for implant-supported prostheses, it is essential to examine the biomechanical properties. Due to their enhanced physical and mechanical features, including durability, biocompatibility, corrosion resistance, and bond strength with ceramics, cobalt-chromium alloys have a wide variety of applications.[16] Titanium alloys, which have a high melting temperature, resistance to deformation, and low modulus of elasticity, are preferred for implants, abutments, and prosthetic frameworks.[17,18] In comparison to other ceramics, zirconia provides improved mechanical and biological features, including a reduced retention of bacterial plaque and excellent shade compatibility.[19] High-performance polymer polyetheretherketone (PEEK) is a non-toxic, biocompatible product. It has a high level of abrasion resistance, is chemically stable, and supports implants, interim abutments, gingiva formers, and frameworks for implant-supported prostheses in the field of dental implantology.[20,21] Regarding CoCr, ZrO_2, Ti, and current PEEK material, whose *in vitro* testing for clinical outcomes are still being conducted, it is essential to assess alternative implant configurations in terms of clinical implications, in comparison to the literature. The use of distally tilted implants is significant when considering the clinical indications from a wider perspective and improving the quality of life with long-term usage of 4 and 6 implant modalities.[9,10,22,23]

Various *in vitro* experiments are performed on dental materials to clarify their physical and mechanical behaviour. Finite element analysis (FEA) is a method for digitally simulating complex structures by using mathematical principles. FEA provides quantitative data for the thorough examination of complex structures, such as bones, implants, and prostheses.[24]

This study aimed to assess the biomechanical properties of the 4 and 6 implant configurations with distally tilted implants and different framework materials used in the fabrication of implant-supported full-arch fixed maxillary prostheses by using 3-dimensional finite element analysis. The null hypothesis was that implant configurations and framework materials affect stresses in the bone, implant, implant screw, framework and prosthetic screw.

MATERIALS AND METHODS

The upper jaw model with teeth (Frasaco, Tettnang, Germany) was scanned using a 3D desktop scanner (Dental Wings Inc., Montreal, Canada) and transmitted to 3D design software (Exocad, Darmstadt, Germany). In the software, tooth components on the model are eliminated and rendered implant-compatible (Fig. 1A). The edentulous models were saved in .stl format, and implant placement was accomplished using a 3D modelling tool (Ansys, Canonsburg, PA, USA). Implants with internal conical connection (Bioinfinity Dental Implant System, Istanbul, Turkey), a diameter of 4.2 mm and a length of 14 mm are appropriately positioned at bone level on 3D edentulous models (Fig. 1B). Mucosa was not included in the model since its effect on the distribution of stress in the peri-implant bone is minimal.

For the four implant-supported designs, implant placement is bilateral, axial to the lateral incisors, and 30° distal inclination in the second premolars (Fig. 1C). Implants were installed bilaterally, axially to the lateral incisors and first premolars, and 30° distal inclination to the first molars in six implant-supported configurations. As indicated in earlier research, the accumulation of stress on implants with more than 30° demonstrates a considerable increase, hence this angulation was selected. The trial version of the program Rhino 7.0 (Robert McNeel & Associates, Seattle, WA, USA) was used to insert straight and angled multi-unit abutments with a 2 mm gingival height on implants. Using the appropriate dental program Exocad (Darmstadt, Germany), the one-piece hybrid structure was designed. Model is created for four implant-supported designs with a 10 mm cantilever length and six implant-supported designs without cantilevers (Fig. 1D). The connector thickness for both designs was fixed at 5 mm fasciolingually and occlusoapically. By attaching prosthetic screws to the framework design, the main model was created.

Geometric models were constructed and solid meshes were used for 3D static analysis (Fig. 2A). Models were created using 10-node tetrahedral elements in order to achieve the highest-quality mesh structure with the maximum number of nodes possible. In locations near the center of the structures in the models, fewer nodal elements are employed

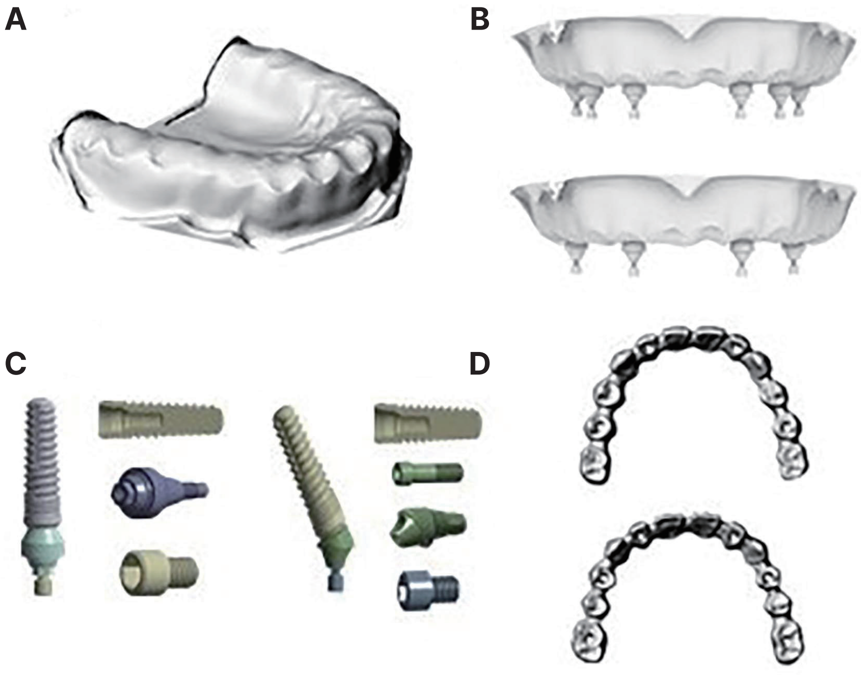

Fig. 1. (A) Raw image of edentulous model, (B) Six and four implant placement, (C) Implant, abutment (straight and 30° angled) and prosthetic screw, (D) Six and four implant supported framework design.

when required to complete the structure. In order to assist the computation, the finest quality mesh structure with the greatest number of nodal elements has been created using this modeling approach. To improve the analysis's validity, elements as many as feasible were created. In addition, the mesh structure was increased and compressed in areas where force would be applied (Fig. 2B). The mathematical model employs 492409 elements and 778014 nodes for the four-implant design and 484334 elements and 758370 nodes for the six-implant design. Convergence test is the process of approximating the real result to equations solved using the Newton-Raphson method in the background while Ansys is running. Ansys does numerical analysis instead of analytical analysis. The result is always shown in numerical analysis; that is, if the result converges, a solution is achieved. This signifies that the analysis has reached a point of convergence. The non-converging analysis indicates that the boundary conditions are not correctly constructed or that the model contains an inaccuracy. Since the results of our study were compatible with and comparable to those of the present literature, no further tests were conducted beyond the software's convergence test. According to earlier research, the thickness of the cortical bone in the maxilla was 1 mm and the rest was cancellous bone to simulate Type 3 bone, to make the results comparable.[8,9,25] All materials evaluated were considered homogeneous, isotropic, and linearly elastic. Table 1 contains the elastic modulus and Poisson's ratio values, which define the physical properties of the structures necessary for model construction.[9,26,27] On the model, implant and prosthetic structures are specified to match the real morphology.

The created model is fixed by holding it in all degrees of freedom. As stated in previous research, cortical and cancellous bone, bone implants, and all connected components such as abutments, frameworks, and screws were thought to be perfectly attached along their contact surfaces, with no relative movement throughout junctions. It was also assumed that osseointegration at the bone-implant contact was completed.[9,10] The following describes the applied forces into the system: 1. assuming centric occlusion, 200 N applied vertically on the palatal tuberosity of the first molar (Fig. 2C). 2. 200 N applied at a 45° angle

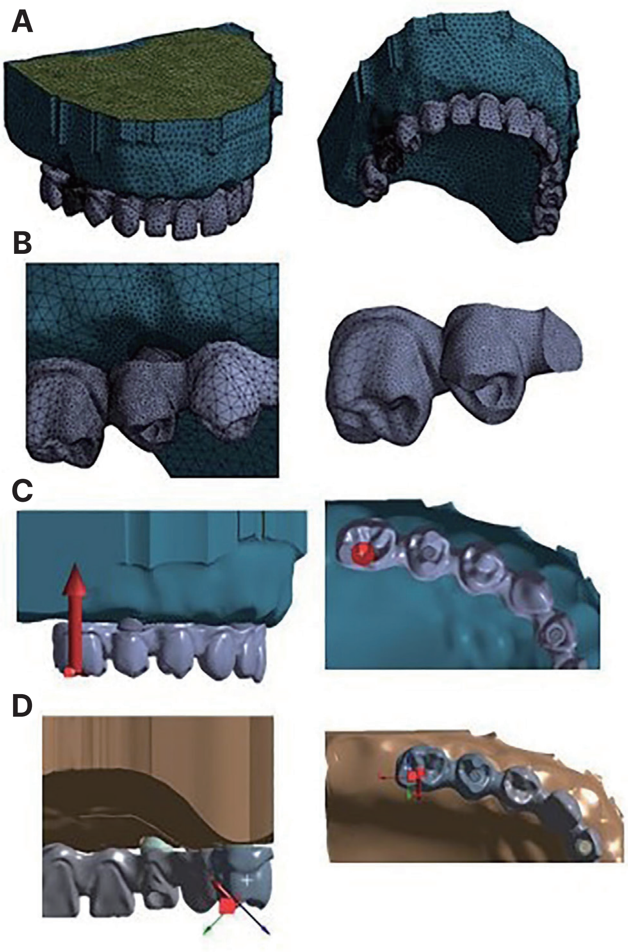

Fig. 2. (A) Finite element model with mesh structure, (B) Element density in the force applied region, Definition of vertical (C) and angled (D) forces.

Table 1. Elasticity modulus and Poisson's ratio of materials

Material	Elasticity Modulus	Poisson's Ratio
Cortical bone[9]	13.700	0.30
Trabecular bone[9]	1.370	0.30
Titanium (Implant, Abutment, Screw)[26]	110.000	0.30
CoCr (Framework) [27]	218,000	0.30
Titanium (Framework)[26]	110.000	0.28
Zirconia (Framework)[26]	210.000	0.30
PEEK (Framework)[26]	4.200	0.36

*PEEK, polyetheretherketone.

to the palatal tuberosity of the first molar (Fig. 2D). As a result, a total of 16 finite element analyses were performed for 2 designs, 4 framework materials, and 2 loading conditions (4 implant-axial loading = 4A, 4 implant-oblique loading = 4O, 6 implant-axial loading = 6A, 6 implant-oblique loading = 6O).

Ansys was utilized for the FEA. For ductile materials, the Von Mises stress (σ_{VM}) was determined, whereas for non-ductile materials, the maximum principal stress (σ_{Max}) and minimum principal stress (σ_{Min}) were determined for each framework material. Additionally, the total deformation of the system was examined.

RESULTS

The stress values obtained in all groups are shown in Figure 3A-F. Regardless of the number of implants, the PEEK framework showed the highest values when the σ_{Max} was measured in bone. Similar results were found in CoCr, ZrO_2, and Ti frameworks. In both loading circumstances, the bone area corresponding to the implant neck showed the highest values (Fig. 4). The highest σ_{Max} value for the bone was obtained at 120.89 MPa in the PEEK framework in 4O. The lowest σ_{Min} value was obtained as -121,7 MPa in 6O. In addition, the σ_{Max} values in the 6-implant design are lower than the 4-implant design.

The PEEK framework provided the highest values for all scenarios when the σ_{VM} stresses on the implant were investigated. The highest σ_{VM} value was 2140.2 MPa on 4O. The lowest σ_{VM} value was 358.1 MPa in CoCr on 6A. The neck of the implant received higher stresses in all conditions. Comparing the 4 and 6-implant designs, the stresses were significantly reduced in the 6-implant design under both loading conditions (Fig. 5).

The highest σ_{VM} on the abutment in PEEK was 4900.5 MPa on 4O and the lowest value was 151.9 MPa in CoCr at 6A design. It was also found that the stress for PEEK decreases by 94.6% for axial loading and 74.11% for oblique loading. The stresses were concentrated on the implant-abutment connection side (Fig. 6).

The highest σ_{VM} achieved on an implant screw was 462.1 MPa in PEEK under 4O conditions. The lowest value measured for CoCr under 6A conditions was 38.3 MPa (Fig. 7).

PEEK had the highest σ_{VM} stress value at 554.9 MPa on 4O. Ti had the lowest value at 143.5 MPa on 6A. The stress in 4O and 4A is localized in the second premolar region (Fig. 8), whereas the stress in 6A and 6O is concentrated in the molar and premolar regions, respectively. On 6O, only PEEK accumulates stresses in the molar area. Lower stress accumulation was observed for all materials in 6 implant-supported frameworks under both loading scenarios

The highest σ_{VM} stresses on the prosthetic screw were found on 4O, 9584.4 MPa in PEEK (Fig. 8). CoCr showed the lowest values, 57.6 MPa on 6A. Under all conditions, the neck of the screw is the area where the force is concentrated. However, in 6 implant-supported structures, lower stress values were achieved.

The maximum deformation of the system is 2,514 mm at 4O in the PEEK framework. The lowest total deformation was 0.027 mm in the CoCr framework at 6A (Fig. 3G).

DISCUSSION

Compared to implant-supported overdentures, treatment with implant-supported fixed prostheses is preferred by clinicians and patients for the rehabilitation of edentulous patients because it improves masticatory function and patient satisfaction. However, in order to distribute forces and offer a biomechanical benefit, this kind of fixed prosthesis necessitates the insertion of multiple implants. One of the most significant objectives of implant dentistry practice is to provide successful treatment while employing a small number of implants and avoiding complex surgical methods. Planning is the key to successful prosthetic treatments, and it’s critical to anticipate the impacts of potential requirement on the system. Finite element analysis can be used to examine objects with complex geometries and a variety of materials.[24] Since FEA is a simulation, the accuracy of the data is improved by increasing the mesh density in the regions to be evaluated on the model. Like previous finite element analysis-based implantology research, the current work makes the assumption that all modelled structures are in constant contact with each other.[28-30] Clinically, however, complete bone-implant

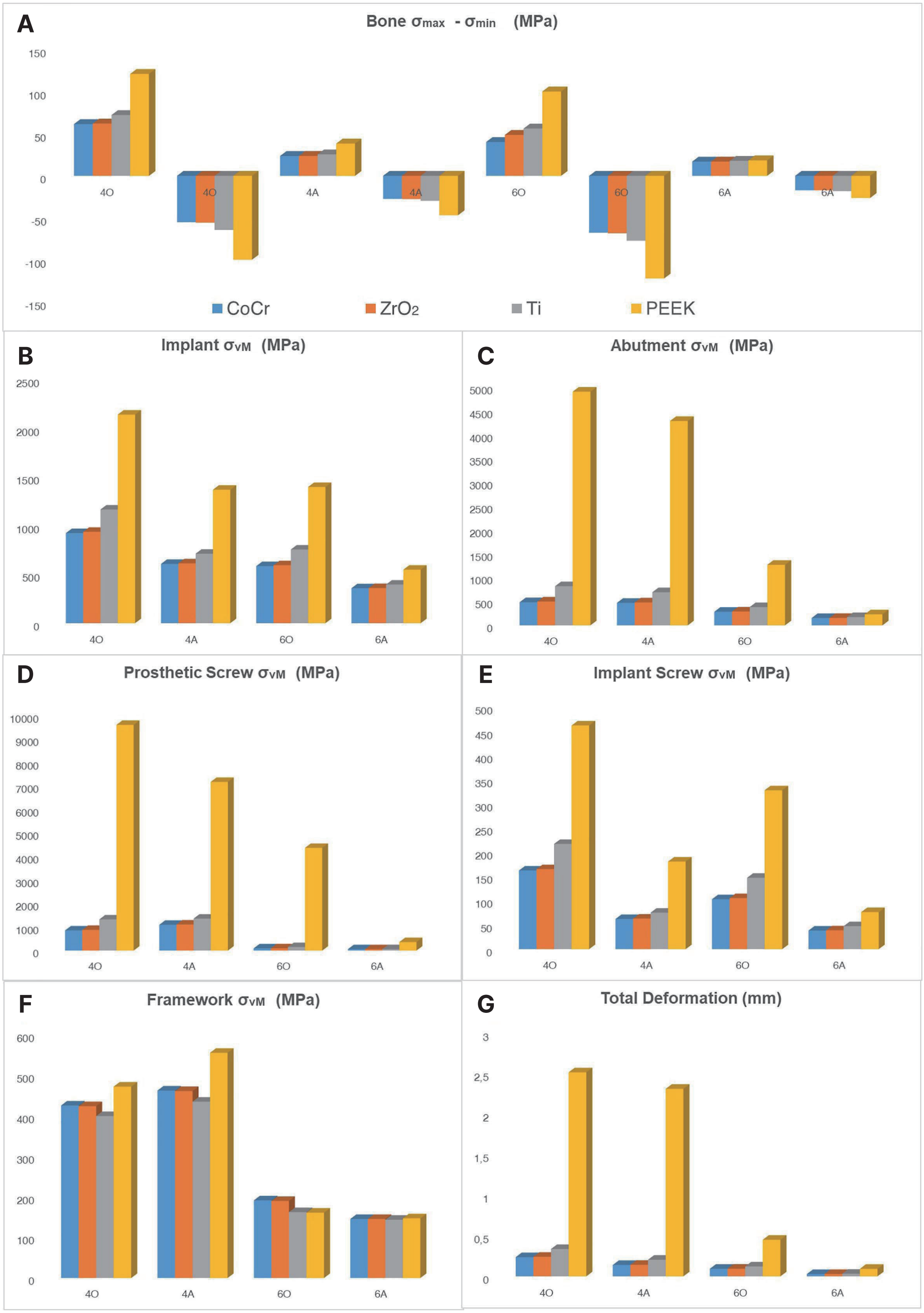

Fig. 3. σ_{max}, σ_{min} values (MPa) in bone (A) and σ_{VM} values (MPa) occurring on the implant (B), abutment (C), prosthetic screw (D), implant screw (E), framework (F) and total deformation (G) under all boundary conditions on the models.

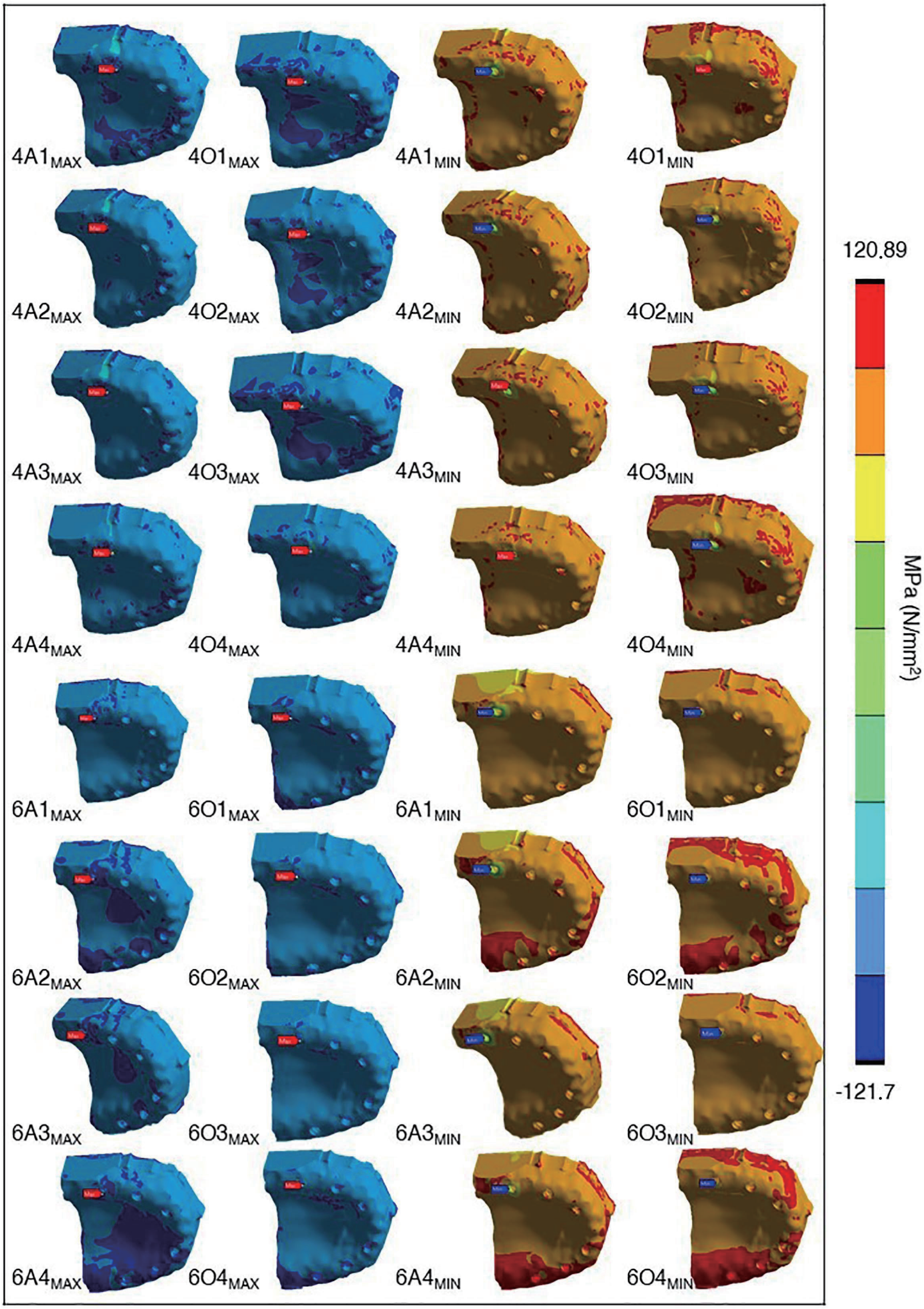

Fig. 4. σ_{max}, σ_{min} stress maps for bone. 1: CoCr, 2: Ti, 3: ZrO_2, 4: PEEK for different loading conditions.

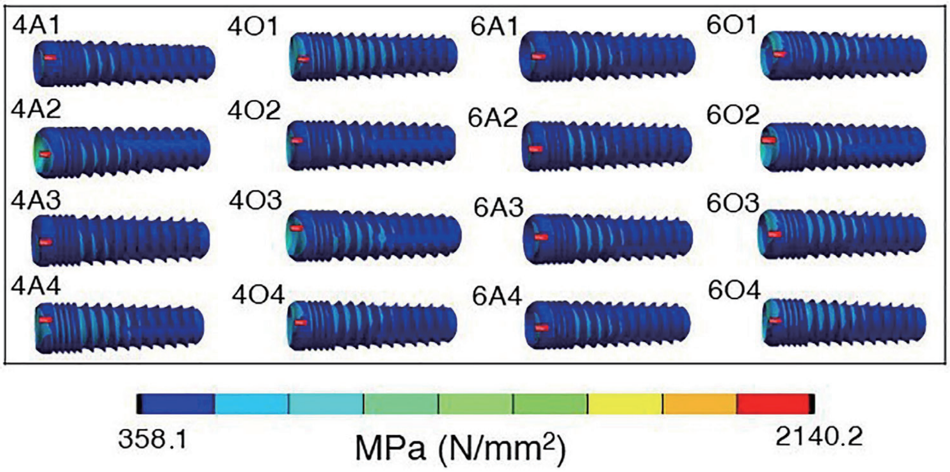

Fig. 5. σ_{VM} stress maps for implant. 1: CoCr, 2: Ti, 3: ZrO_2, 4: PEEK for different loading conditions.

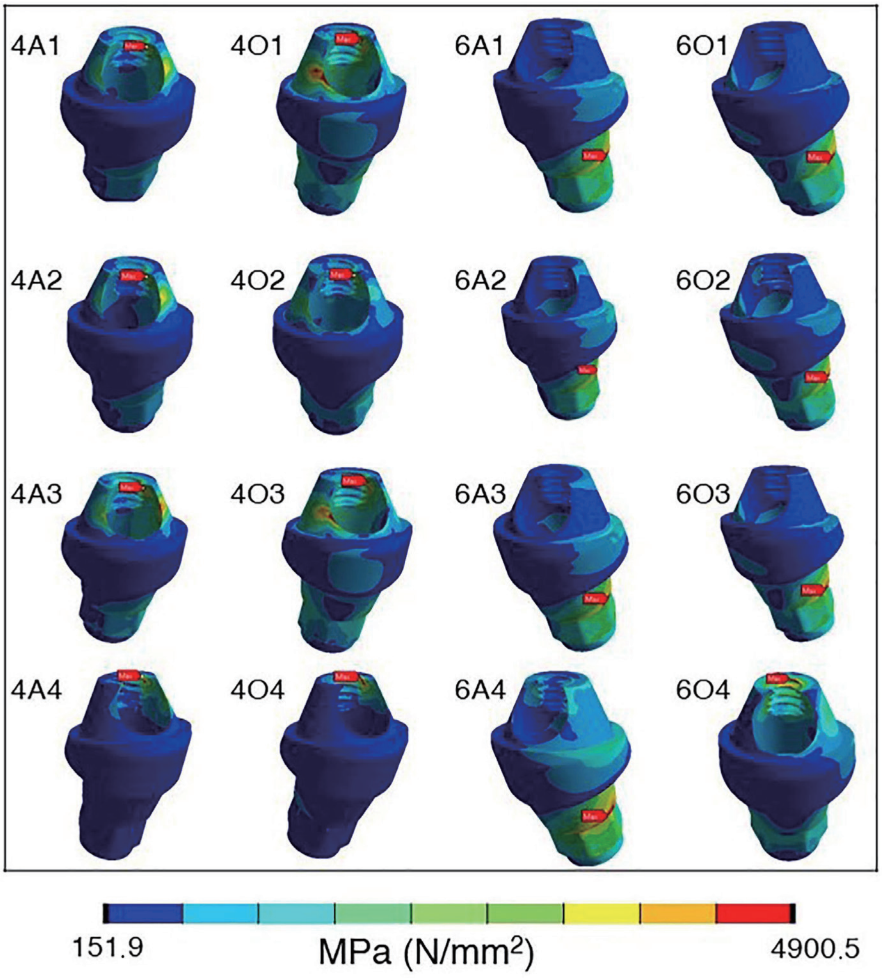

Fig. 6. σ_{VM} stress maps for abutment. 1: CoCr, 2: Ti, 3: ZrO_2, 4: PEEK for different loading conditions.

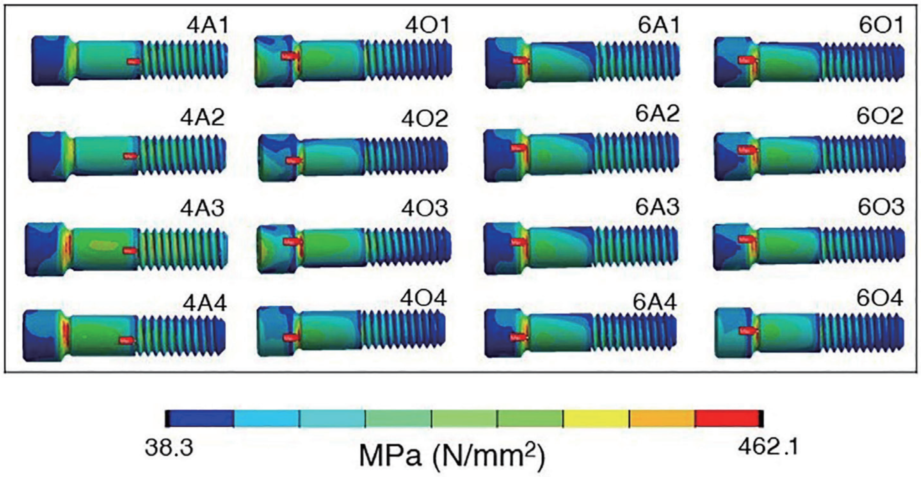

Fig. 7. σ_{VM} stress maps for implant screw. 1: CoCr, 2: Ti, 3: ZrO_2, 4: PEEK for different loading conditions.

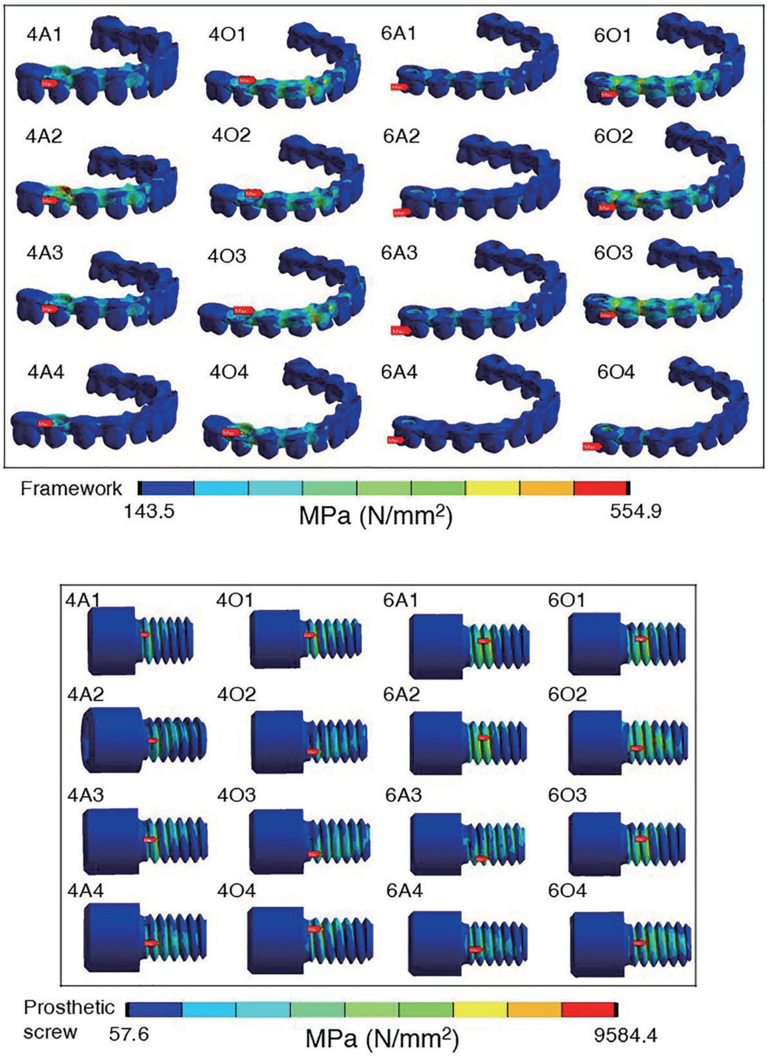

Fig. 8. σ_{VM} stress maps for framework and prosthetic screw. 1: CoCr, 2: Ti, 3: ZrO_2, 4: PEEK for different loading conditions.

contact is not achievable. Therefore, when interpreting the results, this restriction should be considered. In these treatment modalities, it is advised to select implants with a minimum width of 4 mm and a length of 10 - 18 mm for both posterior and anterior position.[5,6] As indicated in earlier research, the accumulation of stress on implants with more than 30° demonstrates a dramatic increase, so the posterior implants were positioned at this angle.[12,31,32] There is currently no standard method for delivering stress to fixed full-arch restorations, despite of previous research using a wide variety of loading factors and locations. However, in the present study, force was applied to molar area where mastication is mostly active in order to observe the effect of cantilever on the system.[28,33,34] Therefore, the loading condition of 200 N on molar area applied unilaterally was designated. Based on the FEA results, the σ_{VM}, σ_{Max} and σ_{Min} were assessed. The structure's tensile stresses were represented by the σ_{Max}, whereas the compressive stresses are σ_{Min}. Von Mises stresses are recommended for the interpretation of ductile materials and principal stresses for brittle materials.[35]

Many different materials such as CoCr, titanium, zirconia, PEEK, PEKK, reinforced plastic, and chromium-nickel can be used as framework materials in implant-supported fixed prosthesis.[34,36-38] These four framework materials, CoCr, titanium, zirconia, and PEEK, which are the most commonly used and developed, were used in our research.

σ_{Max} and σ_{Min} are compared when evaluating the forces exerted on the bone. When the compressive force exceeds 170-190 MPa and the tensile force exceeds 100-130 MPa in the cortical bone, it is considered that the risk of resorption owing to excessive stress build-up.[39] In all analysis, the alveolar crest experienced greater stress. The σ_{Max} and σ_{Min} are higher in PEEK under all conditions, whereas the CoCr material transmits the least force to bone. The values obtained are lower than those reported for bone resorption, which is also consistent with earlier research.[9,10] The σ_{VM} was concentrated at the implant's neck in each tested location, framework, and loading scenario.[8,40-42]

σ_{VM} accumulated at the neck of the abutment in 4 implant-supported designs. However, in 6 implant-supported designs, they were concentrated on the implant-abutment junction for all materials in all loading conditions, except on the cervical area in PEEK under oblique loading. Under oblique loading, PEEK with the lowest modulus of elasticity exhibited the same behaviour. For both loading scenarios and all materials, σ_{VM} on the 4-implant supported design accumulated on the 2nd premolar and posterior implant, and these findings are consistent with previous research.[8,9] For oblique loading, only PEEK with the 6O condition showed concentration in the molar area, whereas all other materials deposited force in the premolar region. PEEK accumulated stress during oblique loading rather than distributing it, in contrast to the stiffer CoCr, ZrO_2 and Ti. As a result, the stresses on the CoCr were the highest under oblique loading while those on the PEEK were the highest under vertical loading. Stiffer materials also contributed to a more uniform distribution of stresses throughout the abutment, implant, and framework. PEEK with a low elastic modulus transmits more stress to implant structures and bone. However, it absorbs stress in the presence of a cantilever. This indicates that it protects the prosthetic structure and is characterized by its polymeric structure. However, high stress levels may result in bone resorption and implant fracture with long-term usage.

In comparing the loading of CoCr and Ag-Pd frameworks, Rubo *et al*. found that CoCr transferred less stress to the implant and abutment.[28] However, in our study, the PEEK experienced increased stress on the implant-abutment complex. Similarly, PEEK crowns caused greater stress accumulation on the abutment, according to the study by Manchikalapudi *et al*.[43] Similarly, according to Kelkar *et al*.,[22] PEEK full-arc prosthetic framework showed higher stress values than zirconia and titanium. In accordance with our findings, Yu *et al*. found that zirconia and metals caused lower stresses on bone and implant, and greater stresses on framework, as compared to polymeric frameworks.[44,45] According to the results of the study in which full-arch implant supported prosthesis were evaluated with different frameworks by strain gauge analysis, PEEK showed higher deformation values than CoCr and ZrO_2 in the presence of cantilever.[11]

Under both loading parameters, the majority of the stress on the prosthetic screw was concentrated in the cervical region. The load on the screw was concentrated in the cervical region because it was unable to dissipate due to the force applied to the prosthesis' terminal location. In the study by Dayan *et al*.,[9] which examined the force distribution on the system employing Ti, ZrO_2, PEEK, and PEKK frameworks in 4 implant-supported designs, the force on the prosthesis screw was larger with PEEK and PEKK materials, and the load was concentrated in the cervical area of the screw. Additionally, Bhering *et al*.[8] examined the impact of material differences on the system in 4 and 6 implant supported structures, and they observed that the load on the prosthetic screw was higher in Ti and the force on the prosthetic screw was higher in 6-implant supported structures. In our study, the force on the prosthetic screw was determined in decreasing order as PEEK, Ti, ZrO_2, and CoCr in all scenarios. However, six implant-supported designs had lower stresses on the prosthesis screws, and these findings varied. PEEK and Ni-Cr bars were used in the study of 4 implant-supported fixed prostheses by Jaros *et al*. They concluded that stresses are concentrated on PEEK and, similar to our study, the stress on bone, implant, and implant components is higher.[46]

In the present study, all of the scenarios with the presence of a cantilever on the system cause excessive stress.[22, 33] Considering the influence of the number of implants on the structure, it is evident that six implants distribute stress more uniformly. In the study of Fazi *et al*.,[33] which investigated stress distribution in 3-4-5 implant-supported prosthesis, it was also shown that load accumulation on the system decreased with 5 implant-supported prosthesis. Similar to the present study, Almeida *et al*.[34] examined maxillary prostheses supported by 4 and 6 implants. They concluded that shortening the cantilever length provides a reduction in the total stress on the system.

It is also interesting that PEEK can respond differently under oblique and axial load among the results of a study. This is based on the material's capability for shock absorption and low elastic modulus. In the study by Sirandoni *et al*.,[10] in which they investigated the loading of Ti, CoCr, ZrO_2, PEEK and PMMA materials in implant-supported prostheses, PEEK and PMMA frameworks with a low modulus of elasticity had the highest total deformation, while CoCr and ZrO_2 had the lowest.

Current dental materials have a continually expanding range of applications. Numerous variables influence the long-term clinical success of a novel material. Due to the low elastic modulus in clinical applications within the distance between implants and biomechanical rules, these characteristics must be taken into account for the long-term clinical success of PEEK frameworks when evaluating the results of this study and applying PEEK, which has a wide range of applications. Higher confidence intervals exist for conventional materials such as CoCr and ZrO_2. With the increase in lifetime and longevity of the produced system, 6 implant model is considered a more reliable system in terms of stress distribution and implant survival than 4 implant structures. Due to anatomical variances and constraints, implants placed up to the posterior area offer a secure foundation for therapeutically applicable frameworks.

The outputs of FEA studies rely on the initial data provided in the system, which are considered as limitations. All materials are introduced to the system as homogenous, isotropic, and linear elastic throughout the creation of the model. Despite the use of this method in FEA investigations, the real reaction of oral tissues cannot be adequately examined. Further *in vitro* testing of the mechanical behavior of the current framework materials and implant modalities under dynamic loads, followed by prospective clinical observational studies, is necessary to confirm the results.

CONCLUSION

The following conclusions were drawn from the results of the study: the distribution of stress on a structure is influenced by implant quantity and framework material. The existence of a cantilever increases the accumulation of stress on the bone, the implant, the abutment, the framework, and the prosthetic screw. As the material's elastic modulus increases, the forces transferred to the bone, implant, abutment, and prosthetic screw decrease. The PEEK material that had the lowest elastic modulus displayed varied biomechanics depending on the loading conditions. For the long-term success of implant-supported fixed prostheses, it is important to have a good understanding of the mechanical and physical properties of the implant design, implant localization, and framework material, and needs to be supported by further more clinical and laboratory studies.

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