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Finite element analysis of stress distribution in implant, abutment, screw and bone using different superstructure materials in implant-retained fixed partial dentures

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ABSTRACT

Aims: Aim of this study is to evaluate the stresses of the implant and implant surrounding tissues by using different superstructure materials by using the finite element method.

Methods: In our study, titanium implants with a diameter of 3.3 mm and 4.8 mm were placed in the 2nd premolar and 2nd molar region of mandibula, respectively. Bone modeling was performed in order to analyse stress distribution by finite element method. 4 different 3-member bridge prostheses were designed by using porcelain fused metal, monolithic zirconia, porcelain fused zirconia and PEEK substructures on the titanium abutments. Analyzes were performed by vertical and oblique forces applied to the occlusal surface.

Results: Oblique forces were found more destructive than vertical forces in all study groups. Stresses depending on the effect of the forces in the bone were affected by compression forces in 4 groups. Stresses in implant, abutment and screw were found around neck area. Stresses in crown were found around connector and margin areas.

Conclusion: It is concluded that the stresses that occur as a result of the applied forces occur at least in zirconia generally.

Keywords: PEEK, monolithic zirconia, zirconia, finite element analysis, dental implant

INTRODUCTION

Throughout history, teeth have symbolized youth, prestige, aesthetics, and health. Consequently, tooth loss has not only resulted in functional impairments but also psychological and social challenges, contributing to the advancement of modern dentistry.[1]

Dental implants have become a crucial treatment modality for the replacement of missing teeth. Ongoing research continues to explore optimal implant designs, surface characteristics, load distribution, and appropriate restorative materials to enhance clinical success.[2]

Unlike natural teeth, dental implants lack a periodontal ligament and transmit occlusal forces directly to the alveolar bone.[3] Proper load distribution is critical for implant longevity, as excessive or misdirected forces may lead to marginal bone loss or prosthetic complications.[4-6]

Factors such as the magnitude and direction of occlusal forces, the restorative design, superstructure material, implant design and number, and bone quality significantly influence load transmission and implant longevity.[7-9] Improper force distribution can result in marginal bone resorption and eventual implant failure.[8] Therefore, selecting the most appropriate prosthetic design and material is essential for long-term success.[10,11]

Finite element analysis (FEA) has been widely used in implant dentistry since 1976 to assess biomechanical behavior under various loading conditions.[11,12] Han et al.,[5] and Almeida et al.[6] recently applied 3D FEA to evaluate stress distribution in implants with different superstructure materials, confirming its reliability for simulating clinical scenarios. FEA enables realistic simulations of clinical scenarios by modeling bone, implant, and prosthetic components, allowing for accurate stress localization analysis that is otherwise difficult to measure in vivo.[13,14] With the FEA procedure, bone, implant and over-implant structures can be modeled close to clinical conditions. Thus, with the applied forces, it makes it possible to accurately monitor the localization of stress and bone losses occurring in the implant and surrounding bone.[15-19]

Although metal-ceramic restorations remain the gold standard due to their strength and long-term clinical performance, concerns regarding metal allergies and esthetic limitations have led to the increased use of metal-free alternatives such as zirconia and PEEK.[20] Monolithic zirconia and PEEK offer improved aesthetics, biocompability and stress distibution compared to metal-ceramic restorations.[4,21] Recent developments in monolithic materials aim to overcome veneering ceramic chipping and improve mechanical performance.[20]

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This study aims to evaluate the stress distribution in implants, abutments, screws, and surrounding bone under vertical and oblique loading using various superstructure materials including porcelain fused metal (PFM), monolithic zirconia, porcelain-fused-zirconia, and PEEK through three-dimensional FEA analysis.

METHODS

This study did not require ethical approval as it did not involve any human subjects or animal experiments. All procedures were carried out in accordance with the ethical rules and the principles.

This finite element analysis was conducted in collaboration with Dicle University, Faculty of Dentistry and Ay Tasarım Ltd. Co. The analysis and modeling were performed using the following hardware and software:

- Intel Xeon® R CPU @ 3.30 GHz, 14 GB RAM, Windows 7 Ultimate
- +Optical scanner: Activity 880 (Smart Optics, Germany)
- 3D modeling: Rhinoceros 4.0 (Seattle, WA, USA)
- Mesh generation: VRMesh Studio (VirtualGrid Inc., Bellevue, WA, USA)
- Finite element analysis: Algor Fempro (ALGOR Inc., Pittsburgh, PA, USA)

After the geometrical models were created using VRMesh software, they were exported in STL format and transferred to the Algor Fempro software for analysis and preparation. The STL format is a universal format for 3D modeling programs and stores the coordinate information of nodes, preventing data loss during inter-program transfers. Material properties (elastic modulus and Poisson's ratio) were assigned to each structural component in the model. The bridge restoration model was digitally designed to match anatomical tooth dimensions as defined in Wheeler's anatomy atlas.

Preparation of Models

In this study, four different models were created. All models used titanium implants and abutments, but differed in prosthetic superstructure materials (Metal supported porcelain, Monolithic Zirconia, Zirconia based porcelain, PEEK).

Modeling of mandibular cortical bone: Using Rhinoceros software, a 20x20x2 mm cortical bone block was first modeled. The periodontal ligament (PDL) space was subtracted using the Boolean method to enable implant and cortical bone alignment. A 0.2 mm thick lamina dura was also modeled (**Figure 1**).

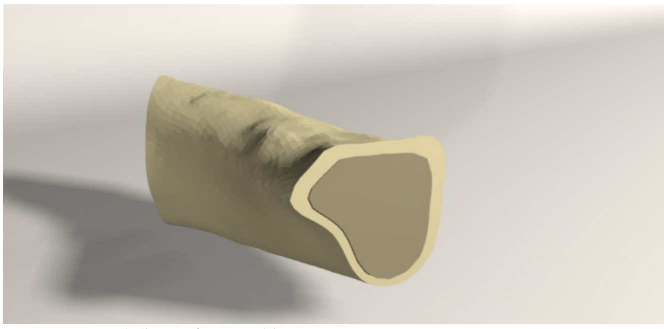

Figure 1. Modelling of cortical bone

Boundary conditions were set by fixing the lower and lateral surfaces of the bone model at 0 degrees of freedom (DOF) (**Figure 2**).

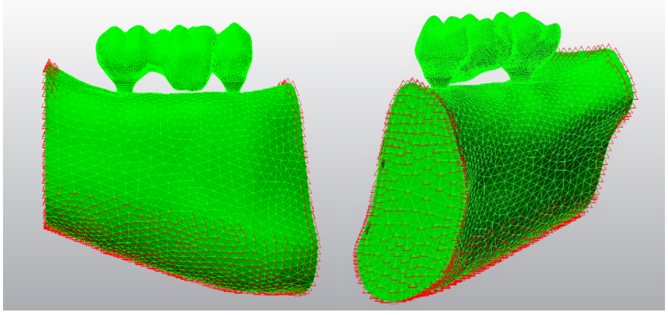

Figure 2. Boundary conditions applied to the models

Modeling of implants, abutments and screws: Implants were positioned at the sites of the second premolars and second molars, guided by anatomical landmarks. The vertical line passing through the mental foramen was accepted as the mesial border of the second premolars. The distal margin was anatomically determined using measurements from Wheeler's atlas.[15]

Titanium implants (Bone Level, Straumann, Basel, Switzerland) with a standard diameter of 3.3 mm and length of 10 mm were placed in the second premolars, while wider implants with a 4.8 mm diameter and the same length were used for the second molars. Titanium abutments (Straumann NC and RC Cementable Abutments) with a 3.5 mm emergence profile were used (**Figure 3-5**).

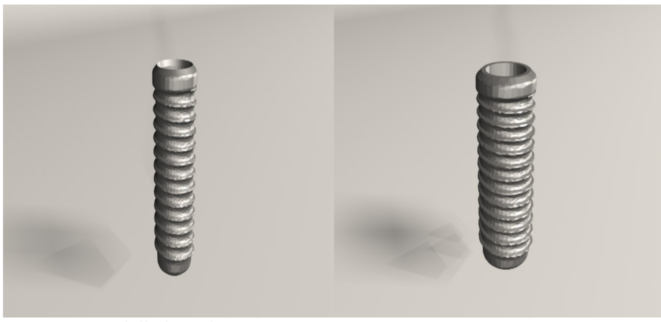

Figure 3. Modelled implants

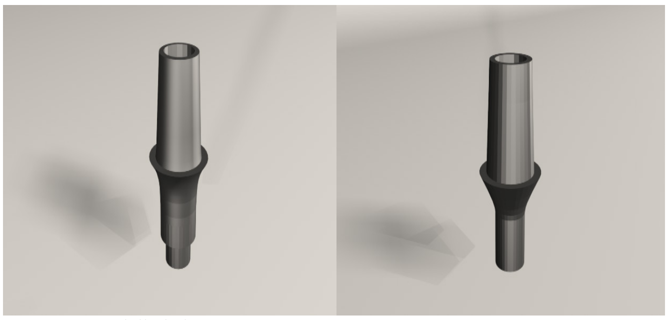

Figure 4. Modelled abutments

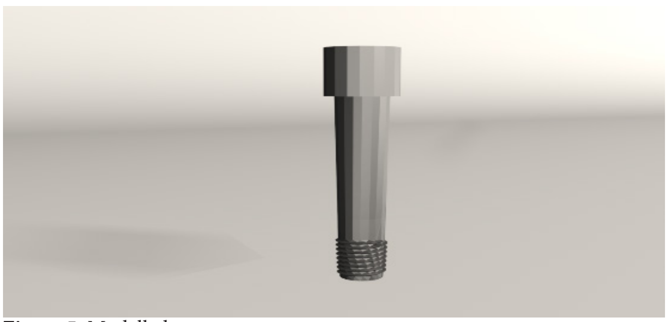

Figure 5. Modelled screws

Modeling of implant supported fixed prosthesis: A three-unit cement-retained implant-supported fixed prosthesis was designed in four different variations. To ensure standardization, all models maintained the same overall prosthesis and substructure thickness.

- ***Group 1:*** Chrome-cobalt alloy (Bego, Bremen, Germany) framework with feldspathic porcelain (Ceramco II; Dentsply, Burlington, USA). The metal framework allowed for 1-2 mm of porcelain covering. Cervical margins were rounded with a minimum thickness of 0.8 mm.
- ***Group 2:*** Monolithic zirconia (BruxZir Solid Zirconia, USA) prosthesis with the same thickness as group 1.
- ***Group 3:*** Zirconia (Y-TZP) infrastructure (NobelProcera, Nobel Biocare AB, Gothenburg, Sweden) veneered with feldspathic porcelain. Cervical margins were also rounded with at least 0.8 mm thickness, and total prosthesis thickness matched the other models.
- ***Group 4:*** Monolithic polyetheretherketone (PEEK) prosthesis (Goodfellow), with the same overall thickness as the other groups.

The number of elements and nodes in each model is summarized in **Table 1**.

Table 1. Number of elements and nodes in the models used

Model	Number of nodes	Number of elements
verticalCrco	75.952	397.952
obliqueCrco	75.952	397.952
verticalMonolithicZr	50.632	248.320
obliqueMonolithicZr	50.632	248.320
verticalZr	75.952	397.952
obliqueZr	75.952	397.952
verticalMonolithicPEEK	50.632	248.320
obliqueMonolithicPEEK	50.632	248.320

In this study, both cortical and trabecular bone were assumed to be isotropic, homogeneous, and linearly elastic. The mechanical properties of the materials used are listed in **Table 2**.

Table 2. Mechanical properties of the materials used in the study

Material	Young's modulus (GPa)	Poisson's ratio	Reference(s)
Titanium (implant, abutment, screw)	110	0.35	(12)
Cortical bone	13.7	0.30	(177)
Trabecular bone	1.37	0.30	(177, 178)
Chrome-cobalt alloy	218	0.33	(14, 179-181)
Feldspathic porcelain	82.8	0.35	(182, 183)
Monolithic zirconia	210	0.35	(184)
Zirconia infrastructure	210	0.30	(184)
PEEK	4.10	0.40	(185)

In total, eight finite element analyses were conducted under two different loading conditions across four groups. As the results of finite element analysis are derived from mathematical modeling without variance, statistical analysis was not applicable. Instead, the focus was placed on the interpretation of cross-sectional images and evaluation of stress magnitudes and distributions at the nodes.

The stresses resulting from applied forces were classified as normal stresses (tensile and compressive, represented by σ) and shear stresses (represented by τ). The greatest stress values in 3D elements occur when all shear components are zero. In the analysis, positive values represent tensile stresses, while negative values indicate compressive stresses. The prevailing type of stress (tensile or compressive) is determined by the greater absolute value and is used for interpretation.

Principal stress values are particularly important for brittle materials. Failure is expected when the maximum principal stress reaches or exceeds the tensile strength, or when the absolute value of the minimum principal stress reaches or exceeds the compressive strength.

RESULTS

In this study, stress distribution was analyzed under vertical and oblique loading in four different prosthetic designs. Von Mises stress values were recorded for implants, abutments, screws, and superstructures. Principal stress values were assessed in the cortical bone.

Vertical Loading Findings

Vertical loads of 1000 N were applied perpendicularly (90°) to the occlusal surfaces at three points across the molar and premolar units.

Von Mises stress values under vertical loading: The stress values in the implant components and prosthetic structures are summarized in **Table 3**.

Table 3. Von Mises stress values under vertical forces

Group	Component	Von Mises stress (N/mm²)
Group 1	Superstructure (connector area)	24.81
	Abutment (cervical region)	361.68
	Titanium screw	131.68
	Titanium implant (neck region)	199.04
Group 2	Superstructure (cervical region)	90.82
	Abutment (cervical region)	356.44
	Titanium screw	125.45
	Titanium implant (neck region)	196.26
Group 3	Superstructure (connector area)	25.11
	Abutment (cervical region)	361.54
	Titanium screw	131.51
	Titanium implant (neck region)	199.16
Group 4	Superstructure (connector area)	90.82
	Abutment (cervical region)	356.44
	Titanium screw	125.45
	Titanium implant (neck region)	196.26

Principal stress values in cortical bone under vertical loading: The stress values in the implant components and prosthetic structures are summarized in **Table 4**.

Table 4. Principal stress values under vertical loading

Group	Maximum (MPa)	Minimum (MPa)
1	3.48	-28.65
2	3.49	-28.17
3	3.48	-28.67
4	3.34	-30.93

Oblique Loading Findings

An oblique force of 500 N at 30° was applied to simulate parafunctional occlusal loading. Stress concentrations were found to be significantly higher compared to vertical loading.

Von Mises stress values under oblique loading: Stress values under oblique forces are shown in **Table 5**.

Table 5. Stress values under oblique forces

Group	Component	Von Mises stress (N/mm²)
Group 1	Superstructure (cervical region)	88.27
	Abutment (cervical region)	1706.35
	Titanium screw	449.20
	Titanium implant (neck region)	940.11
Group 2	Superstructure (cervical region)	415.76
	Abutment (cervical region)	1680.53
	Titanium screw	452.32
	Titanium implant (neck region)	938.00
Group 3	Superstructure (cervical region)	90.53
	Abutment (cervical region)	1707.73
	Titanium screw	449.04
	Titanium implant (neck region)	940.91
Group 4	Superstructure (cervical region)	52.74
	Abutment (cervical region)	2376.10
	Titanium screw	479.78
	Titanium implant (neck region)	1290.96

Principal stress values in cortical bone under oblique loading: Stress values around the implant neck are summarized in **Table 6**.

Table 6. Stress values around the implant neck

Group	Maximum (MPa)	Minimum (MPa)
1	87.81	-108.46
2	86.79	-108.76
3	87.88	-108.43
4	114.00	-126.49

DISCUSSION

According to numerous studies, one of the most important factors affecting the short- and long-term success of implant treatments is biomechanics.[16-18] Oblique loading generates higher stresses than vertical loading, highlighting the clinical importance of load direction in implant planning.[4,5]

Unlike natural teeth, dental implants do not have periodontal ligaments. Therefore, the forces exerted on the prosthesis are directly transmitted to the bone.[19,20] These forces should be considered during implant planning, and a homogeneous distribution of the loads on the bone must be ensured for successful outcomes. It has been reported that the material of the implant and prosthesis, their design, and the crown-to-implant ratio significantly influence these forces. Consequently, selecting appropriate implant and prosthesis options is critical for treatment success.[15,22-26]

Various methods can be used for stress analysis, each with its own advantages and limitations.[27,28] For example, the brittle varnish method does not provide numerical data, and strain gauge methods only measure stress at contact points. Laser beam and radiotelemetry methods are less commonly used due to practical challenges. The FEA method involves dividing the model into numerous small elements, analyzing each part separately, and then combining the results using mathematical equations. Many studies have emphasized the superiority of the FEA method compared to other techniques.[16,18,29-34]

Several FEA methods have been employed in implantology studies to investigate stress distribution.[35-37] Recent studies have utilized both two-dimensional (2D) and three-dimensional (3D) FEA to evaluate the impact of different prosthetic superstructure materials on stress distribution.[35,38] Among these, 3D FEA has been reported to offer more detailed and realistic analyses.[39-41]

Meijer et al.[42] and Clelland et al.[43] noted that the accuracy of FEA results is directly proportional to the number of nodes and elements in the model. For more realistic 3D FEA results, it is recommended to use at least 30,000 nodes and 200,000 elements.[44-46] In our study, a minimum of 50,000 nodes and 200,000 elements were used in each model.

Advanced digital imaging techniques are essential for obtaining accurate results in FEA.[11] These include computed tomography (CT), magnetic resonance imaging (MRI), and laser scanning. Among these, CT has been reported to produce the most detailed images.[47] Therefore, in our study, the mandibular model was created using CT imaging.

With the aid of constantly evolving technology, our models were designed to be as anatomically accurate and detailed as possible in three dimensions. Type 2 bone, as classified by Lekholm,[19] is typically found in the posterior mandible.[48-50] In our model, cortical bone thickness was set at 1.5 mm, and trabecular bone thickness was set at 13 mm, with implants placed in the posterior mandibular region.

Since implants lack the periodontal ligament, they transmit occlusal forces directly to the bone, differing significantly from natural teeth in this regard. Therefore, masticatory forces play a crucial role in maintaining the bone-implant interface.[23] Studies have demonstrated variability in occlusal forces. In this study, 1000 N of vertical force and 500 N of oblique force were applied. An angle of 30° was chosen for the oblique force, consistent with the literature.[48]

Metal-ceramic restorations remain the most commonly used prosthetic materials due to their durability, strong porcelain bonding, affordability, and accessibility. However, they also have disadvantages, including corrosion and potential allergic reactions. Other drawbacks include mismatches in thermal expansion coefficients between the metal and porcelain, discoloration due to silver content, and poor esthetics due to limited light transmission.[51]

Considering these limitations, we included metal-supported porcelain crowns (still regarded as the gold standard) in one of the experimental groups. Recently, zirconia-based ceramics with high durability and improved optical properties have gained popularity. Although zirconia has strong mechanical properties, it is veneered with glass ceramics due to its limited translucency. The most common issue with this type is the frequent detachment between the porcelain and zirconia layers.[52-56] To overcome this, newer monolithic materials have been developed, among which monolithic zirconia has demonstrated the best clinical performance.[52,54] Monolithic zirconia reduces stress concentrations and layer detachment, improving clinical performance in posterior implant restorations.[4]

Hybrid ceramics and PEEK have also been introduced to meet the demand for biocompatible and esthetically pleasing materials.[57] Due to its elastic properties, PEEK is believed to reduce occlusal forces and minimize stress transmission to implants. Thus, it may help solve stress-related problems.[58] Han et al.,[5] Almeida et al.,[6] and Schwitalla et al.,[21] reported that PEEK frameworks reduce prosthetic stress but may increase compressive stress in bone, which should be considered in patients with low bone density. It has been suggested that PEEK can serve as both an abutment and prosthetic material.[59] CAD-CAM-manufactured PEEK fixed prostheses have demonstrated higher fracture resistance compared to lithium disilicate glass ceramics (950 N), alumina (851 N), and zirconia (981-1331 N).[60,61]

Considering all this data, our study compared monolithic zirconia, zirconia-based porcelain, and PEEK materials via finite element stress analysis as potential alternatives to metal-ceramic restorations.

Limitations

This study was conducted using FEA. Although FEA is an effective tool for simulating clinical scenarios, it has certain limitations. The materials were assumed to be homogeneous, isotropic, and linearly elastic, which may not fully represent the complexity of biological tissues. Clinical variables such as bone remodeling, fatigue loading, and patient-specific anatomical variations were not considered. Therefore, the results should be interpreted within the context of the simulation and need to be validated by clinical studies.

CONCLUSION

- Oblique forces generated higher stress values than vertical forces.
- Under vertical forces, the lowest Von Mises stress in implants was observed with monolithic zirconia, and the highest with PEEK.
- In the premolar implant region, the highest abutment stress under vertical force was observed with metal-supported ceramic, and the lowest with monolithic zirconia.
- In the molar implant region, the highest abutment stress under vertical force was seen with PEEK, and the lowest with monolithic zirconia.
- In the premolar region under vertical force, the highest screw stress was observed with monolithic zirconia, and the lowest with PEEK.
- In the molar region under vertical force, the highest screw stress was found with PEEK, and the lowest with monolithic zirconia.
- Maximum stress in bridge restorations under vertical force was observed in monolithic zirconia, and minimum in metal-supported ceramic.
- Under oblique forces, the highest stress values in the premolar region abutments and screws were found with monolithic zirconia, and the lowest with PEEK.
- In the molar region under oblique forces, the highest stress in abutments and screws was observed with PEEK, and the lowest with monolithic zirconia.
- The highest bridge stress under oblique force was found with monolithic zirconia, and the lowest with PEEK.
- In the premolar implant region, bone stresses under both vertical and oblique forces were compressive, with maximum stress observed with PEEK and minimum with monolithic zirconia.
- In the molar implant region, bone stresses were also compressive, with maximum stress observed with monolithic zirconia and minimum with PEEK.

ETHICAL DECLARATIONS

Ethics Committee Approval

This study did not require ethical approval as it did not involve any human subjects or animal experiments.

Informed Consent

Since the study was conducted without the participation of any living being, no written consent form was obtained.

Referee Evaluation Process

Externally peer-reviewed.

Conflict of Interest Statement

The authors have no conflicts of interest to declare.

Financial Disclosure

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Author Contributions

All of the authors declare that they have all participated in the design, execution, and analysis of the paper, and that they have approved the final version.

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