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Biomechanical effects of two different collar implant structures on stress distribution under cantilever fixed partial dentures

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Abstract

Objective. The purpose of the study was to compare the effects of two distinct collar geometries of implants on stress distribution in the bone around the implants supporting cantilever fixed partial dentures (CFPDs) as well as in the implant-abutment complex and superstructures. ***Materials and methods.*** The three-dimensional finite element method was selected to evaluate the stress distribution. CFPDs which was supported by microthread collar structured (MCS) and non-microthread collar structured (NMCS) implants was modeled; 300 N vertical, 150 N oblique and 60 N horizontal forces were applied to the models separately. The stress values in the bone, implant-abutment complex and superstructures were calculated. ***Results.*** In the MCS model, higher stresses were located in the cortical bone and implant-abutment complex in the case of vertical load while decreased stresses in cortical bone and implant-abutment complex were noted within horizontal and oblique loading. In the case of vertical load, decreased stresses have been noted in cancellous bone and framework. Upon horizontal and oblique loading, a MCS model had higher stress in cancellous bone and framework than the NMCS model. Higher von Mises stresses have been noted in veneering material for NMCS models. ***Conclusion.*** It has been concluded that stress distribution in implant-supported CFPDs correlated with the macro design of the implant collar and the direction of applied force.

Key Words: *dental stress analysis, implant collar, cantilever fixed partial denture*

Introduction

The use of osseointegrated implants as abutments for fixed partial dentures (FPDs) has become a fashionable treatment option for partially edentulous patients. The long-term success of conventional end-abutment supported FPDs have been evaluated by several researchers [1,2]. However, this therapy is often limited by clinical situations such as bone deficiencies and/or the presence of anatomical structures (i.e. mental foramen, mandibular canal or maxillary sinus) in areas where implants have to be ideally (prosthetically) placed. Thus, sometimes it is not possible to use two implants at each end of the edentulous area to support 3-unit FPDs. Unfavorable local conditions of the residual edentulous ridges may lead to two therapeutic options for a rehabilitation by means of implant supported FPDs. One of the alternatives is the pre-implant reconstructive and/or regenerative procedures such as guided bone regeneration, mental nerve repositioning and sinus floor augmentation [1–4]. The other option is the treatment of a partially edentulous site with cantilever fixed partial dentures (CFPD) [5–7]. The increasing cost and morbidity of complex dental treatment using such advanced approaches together with a number of related complications may limit the choice of complex treatment in daily practice [8–10]. As a simple and economical procedure, edentulous ridges next to implants may be reconstructed by means of cantilevers or pontics [2,11,12].

CFPD is a restoration with one or more abutments at one end and unsupported at the other end. A class I lever system is created if vertical and oblique forces

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directed to the pontic results in forces on the abutment teeth greater than the applied stress [13]. Previous mechanical studies have demonstrated that the CFPDs supported by dental implants could induce excessive stress concentration in the supporting alveolar bone that may facilitate bone resorption under occlusal loads, especially in the cervical region of implants [14]. Higher strain concentrations at the implant sites, particularly at the level of the implant bone interface adjacent to the extensions, were noted [1].

The relationship between biology and mechanics and the stress distribution within tolerable limits to the supporting bone and related structures is critical with CFPDs. Analyzing force transfer at the bone–implant interface is an essential step in the overall analysis of loading, which determines the success or failure of implants and consequently the success of the prosthetic superstructures [15].

The clinical success of osseointegrated implants are largely influenced by the manner mechanical stresses are transferred from the implant to the surrounding bone, without generating forces of a magnitude that would jeopardize the longevity of implants and prostheses [16]. The transfer of functional loads and accompanying stress distribution in a bone-implant-denture assembly depend on the physical properties and special geometry of each component [17–19]. To accommodate the new clinical applications of modern implant dentistry, manufacturers have modified both the macro- and microstructures, e.g. shape, type of implant abutment connection, presence of thread, thread design and surface treatment [20]. Most notably, among the factors proposed as possible causes for bone resorption were implant surface characteristics and the placement of the smooth neck portion of the implant in contact with bone [21]. Some studies have shown that certain implant designs may contribute to bone loss (the geometry of the coronal collar has been implicated) while other studies have indicated that such bone loss could be prevented by incorporating a biomechanically stable joint and retention elements such as microthreads within the collar of the implant [22–24]. Norton [25] stated a low amount of crestal bone loss for dental implants with microthreads at the implant neck. The principles of retention elements in the terms of smaller sized minute threads on the implant neck was introduced in implant systems in the 1990s and it was observed that the retention elements at the endosseous neck portion of the implant would affect the stress distribution [20].

Although many studies have been made regarding the implant supported CFPDs, including the influence of implant diameter and length, the biomechanical effects of different collar implant structures on stress distribution with CFPDs has not been determined yet. The hypothesis of the study was based on a system of the effect of different collar geometries upon the stresses in the bone and the prosthetic components under 3-unit cantilever FPDs. The modeling of these stresses with computer graphics programs and the biomechanical analysis rendered by the three-dimensional finite element method (3D-FEA) are promising alternatives for addressing the subject [26–28].

The aim of the present study was not only to investigate the differences between the stress transfer properties of two different implants that differ significantly in macroscopic collar geometry which support 3-unit cantilever FPDs, but also to evaluate the effects of collar architectural changes on the implant–abutment complex, in bone and the prosthetic superstructures itself.

Materials and methods

To evaluate the stress distribution around the bone, implant–abutment complex and 3-unit cantilever FPD as well as the biomechanical behavior of the above-mentioned components, finite element analysis (FEA) was conducted using mathematical models of unilateral posterior 3-unit cantilever FPDs supported with two osseointegrated implants which contain different collar geometries.

Finite element model

Two three-dimensional finite element models were generated; each representing a 3-unit cantilever FPD supported by two implants. A 4.2 cm long representative posterior section of an edentulous human mandibular bone was used as the basis of a mandibular finite element model in this study. The implants embedded in the second premolar and first molar sites were used. Different associations of implants' collar geometries were tested. Totally edentulous mandibular bone was used as the basis of a mandibular finite element model in this study.

The modeled section of the mandible was composed as a dense cancellous core surrounded by a thick layer of cortical bone. The average thickness of the cortical bone in the crestal area was 2.0 mm. Serial axial sections in every 0.5 mm level of an edentulous mandible were obtained from a Cone-Beam CT (CBCT) imaging System (NewTom 3G; QR, Verona, Italy). The CBCT images were stored using DICOM 3.0 as a medical image file format and imported into 3D medical image processing software, version 2.2.2 (Maxilim Software; Medicim Company, Mechelen, Belgium). The 3D image of the mandible imported with the.stl file format into pre-processing software, version 2005 (MSC Mentat; MSC Software Corporation, CA, USA) for pre-processing and modeling.

In the current study, a model of 14.0 mm long and 4.1 mm in diameter solid screw bone level Straumann implant (ITI; Straumann AG, Waldenburg, Switzerland) for non-microthread collar structure (NMCS) and a model of 13.0 mm long and 4.0 mm in diameter solid screw Astra Tech implant for the microthread collar structure (MCS) (Astra-Tech; Astra-Tech AB, Molndal, Sweden) additionally cementable bone level system abutment RC for Strauman (5 mm in diameter with 5.5 mm abutment height) and cementable, direct abutment (5 mm in diameter with 6 mm abutment height) for the Astra implants were selected. The geometry of the implants as well as the abutments was modeled according to engineering drawings by using a specialized software program (MSC Mentat; MSC Software Corporation). Cobalt–Chromium (Bego; Bremen, Germany) was used as crown framework material and feldsphathic porcelain was used for the veneering material. The thickness of the metal frameworks was 0.5 mm and the veneering material varied from 0.8–1.5 mm in thickness from the cervical to occlusal surface. The cement thickness was ignored. All the final solid meshes were constituted by tetrahedral elements with four nodes by using a finite element solver program (MSC Marc; MSC Software Corporation). The model used in this study is shown in Figure 1.

Material properties

All materials used in the models were considered to be isotropic, homogenous and linearly elastic. The elastic properties were taken from the literature, as shown in Table I [29].

Interface condition

The current study simulated osseointegrated screw implants with different collar geometry. For simulation such an osseointegrated condition fixed bond was set to its interface with the bone. Similarly, the type of bond was provided at the implant-abutment and abutment–prosthesis interfaces.

Constrains and loads

Models were constrained in all directions at the nodes on the mesial and distal bone surface; 300 N static vertical loads were applied along the central fossa of each superstructure. To simulate the oblique load condition, an oblique static load of 150 N in total was applied to the buccal cusps of each crown with an inclination of 60° buccally from the vertical; 60 N static loads were applied in mesiodistal direction horizontally to mimic the parafunctional movement of the jaw. The maximum equivalent von Mises stress in implant-abutment complex as well as in the metal framework and ceramics and the maximum principal stress and minimum principal stress in the jaw bone were set as output variables to evaluate the effect of different implant collar designs under 3-unit cantilever FPD.

Convergence analysis

As there are no appropriate experimental results in literature to compare our results to validate the finite element model, six different models for each implant system with changing numbers of elements were compared and the convergence of the results was examined using the result of the maximum tensile stress in cortical bone under vertical load condition (Figure 2). The results and the number of the elements are shown in Table II. From the table the convergence for the two models was reached for

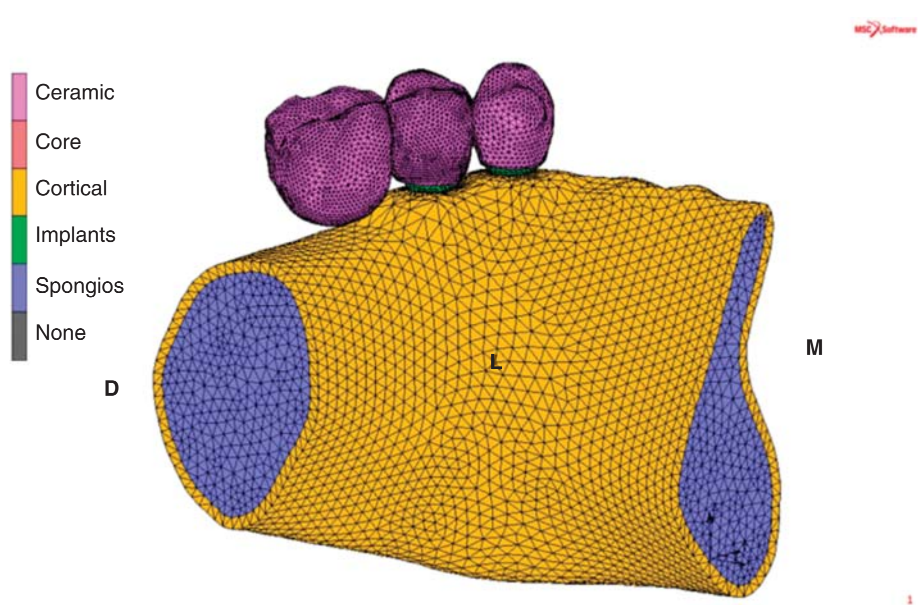

Figure 1. The representation of the cantilever FPD model.

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

Material	Young modulus (GPa)	Poisson ratio (v)
Cortical bone	14.8	0.30
Cancellous bone	1.85	0.30
Ti-6Al-4V	110	0.32
Cobalt–Chromium alloy	220	0.30
Feldisphatic porcelain	61.2	0.19

the following element numbers and the result of those models were considered in the rest of the study. Thus, our 3D models consist of 247,342 elements and 45,817 nodes for the NMCS model and 250,166 elements and 47,736 nodes for the MCS model.

Results

In the current study, maximum (tensile stress) and minimum (compressive stress) principle stresses (P_{max} and P_{min}) on bony structures and von Mises stresses on implants, frameworks and veneering materials were calculated. The highest values of P_{max} and P_{min} in the cortical and in the cancellous bone are given in Tables III and IV, respectively, while the von Mises stress values are shown in Table V. The stress patterns recorded in both models were similar, but the magnitude of the stresses was found to be different.

A color scale with 12 stress values served to measure quantitatively the stress distribution in the model components. The scaling was selected not to represent the yield strength but rather to provide clear visualization of the region of stress.

Cortical bone

In the cortical bone, the highest P_{max} stress was observed in the MCS model in the case of vertical load buccally and mesiobuccally around both the mesial and distal implant neck (Figure 3A). In the horizontal load case the P_{max} stresses were concentrated in the models adjacent to both implant necks lingually. In the case of oblique load, it was observed that distal and distolingual cortical surroundings have been affected more in the distal implant than the mesial one (Figure 3B).

In the vertical load case, the P_{min} stresses were concentrated in the distal and distolingual cortical surroundings of distal implants (Figure 3C), whereas stresses were concentrated in the distobuccal cortical surroundings of the distal implant and the mesial aspect of mesial implant housing within the oblique load. The P_{min} stresses have been determined at the buccal collar area of both mesial and distal housings in the case of horizontal load.

Cancellous bone

In cancellous bone, the highest P_{max} stress was found in the MCS model lingually within the horizontal load case (Figure 4A). In the vertical loading condition the P_{max} stress has been concentrated at the collar area extended all around the mesial and distal housings in the direction of mesially and mesiobuccally. P_{max} stress was located on the distal aspect of the distal implant under oblique loads (Figure 4B). The P_{min} values in the case of vertical load have been determined on the distal region of the distal implant, whereas the horizontal and oblique loads generated the stress at the buccal sides of both housings (Figure 4C).

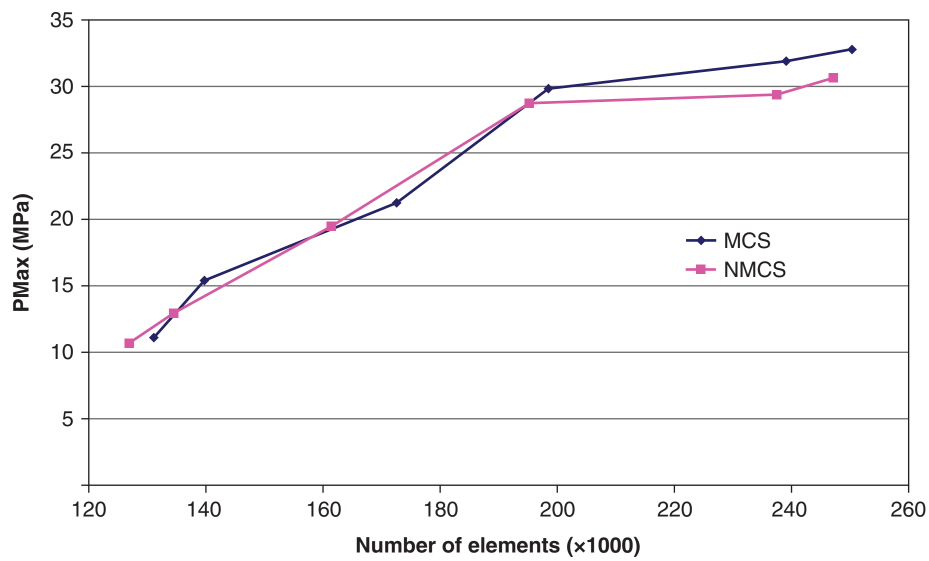

Figure 2. Converge of the two models with the changing numbers of elements and nodes.

Table II. Principal maximum stress values for vertical load in cortical bone with a changing number of elements for convergence and final numbers of elements and nodes.

# Of elements for MCS model	130,864	139,821	172,576	198,341	238,987	250,166
# Of elements for NMCS model	126,746	134,578	161,512	195,216	237,441	247,342
	Principal maximum stress (Pmax) (maximum tensile stress) MPa					
Microthread vertical load case	11.1	15.4	21.2	29.8	31.9	32.8
Relative error		28%	27%	28%	7%	3%
Non-microthread vertical load case	10.7	12.9	19.4	28.6	29.3	30.6
Relative error		17%	33%	32%	2%	4%
	MCS (Microthread) model			NMCS (Non-microthread) model		
Final # of elements	250,166			247,342		
Final # of nodes	47,736			45,817		

Implant–abutment complex

Under vertical load, higher stress values in implant–abutment complex were recorded in the MCS model, whereas lower stress values were recorded under horizontal and oblique loads when compared with the NMCS model. The highest stress value has been noted in the implant–abutment complex in MCS models under vertical loads. The stress values observed in the distal implant–abutment complex were higher than those seen in the mesial implant–abutment complex in the case of all loading conditions in both models. Moreover, the highest stress value was determined around the implant–abutment junction and on the collar threads under the three loading cases (Figure 5).

Framework

Comparison of the NMCS and MCS models from the standpoint of frameworks has revealed that higher von Mises stresses were found in the MCS models within the horizontal and oblique loads, respectively. The highest stress values were observed in the NMCS model under the vertical load.

Dispersion of the stress in the frameworks in both models seemed very similar. Under the vertical load, higher stress was observed on the second premolar's lingual surface, while lesser stress concentrated on the first premolar and first molar in the MCS (Figure 6A). Stress was also concentrated at the mesial and distal connectors under the vertical load. Within the horizontally loaded models, stress was mainly concentrated around the collar area of both first and second premolars and the connectors (Figure 6B), while within the obliquely loaded models, stress was only concentrated around the distal connectors.

Veneering part

When comparing the veneering parts, lesser von Mises stresses have been noted in the MCS models in all three loading conditions. Dispersion of the stress

Table III. Highest principal maximum and principal minimum (tensile maximum and compression maximum) values (MPa) recorded in the cortical bone.

	Principal stress maximum (Maximum tensile stress) (P_{max})			Principal stress minimum (Maximum compression stress) (P_{min})		
Model	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)
MCS (Microthread)	32.8	26.7	8.07	89.5	29.1	20.5
NMCS (Non-microthread)	30.6	37.6	11.3	83.3	43.3	27.3

Table IV. Highest principal maximum and principal minimum (tensile maximum and compression maximum) values (MPa) recorded in the cancellous bone.

	Principal stress maximum (Maximum tensile stress) (P_{max})			Principal stress minimum (Maximum compression stress) (P_{min})		
Model	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)
MCS (Microthread)	16.4	50.2	14.4	21.5	48.1	13.1
NMCS (Non-microthread)	30.9	18.9	10.8	31.2	20.2	12.1

Table V. Highest von Mises stress values (MPa) recorded in the implant–abutment, framework and veneering material.

	Implant–Abutment			Framework			Veneering material		
	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)	Vertical (MPa)	Horizontal (MPa)	Oblique (MPa)
MCS (Microthread)	89.3	36.4	26.6	69.7	22.0	34.4	116.1	51.0	54.7
NMCS (Non-microthread)	76.0	38.7	29.3	98.8	20.6	29.6	137.6	111.0	99.6

in the veneering part in both models seemed very similar.

Under the vertical load stress was concentrated at the distal connector, first premolar's collar area and occlusal surfaces of the premolars (Figure 7A). The stress was observed at the connectors in the case of the oblique and horizontal loads (Figure 7B).

Discussion

In recent years in order to improve osseointegration, studies have focused on implant material, shape and surface characteristics [27]. There are an increasing number of studies targeting the cause of failure in clinical practice, using various stress analysis methods [30]. Bioengineering studies have been useful in understanding the response of biological tissues and bone to mechanical stress [28]. In the present study FEA was used to evaluate the stress transfer properties of two currently marketed implants which support 3-unit cantilever FPDs.

FEA is a technique for obtaining a solution to a complex mechanical problem by dividing the problem domain into a collection of much smaller and simpler domains (elements). The geometry of the components in an implant supported FPDs are extremely complex, therefore FEA has been viewed as the most suitable tool for analyzing them. In the present study FEA has been used in the prediction of biomechanical performance of two currently marketed implants that differ significantly in macroscopic collar geometry which support 3-unit cantilever FPDs [17,28].

In the current study, several assumptions and simplifications have been made with regard to the material properties and model generation. In FEA models bone is frequently modeled as isotropic, in fact it is anisotropic [26]. The properties of the materials modeled in the study, particularly the living tissues, however, are different. For example, it is well known reality that the actual cortical bone of the mandible is transversely isotropic and inhomogeneous [9,29]. The structures in the model were all assumed to be homogenous, isotropic and linear elastic and the implants were assumed to be 100% osseointegrated. Furthermore, histomorphometric studies indicated that there is never a 100% bone–implant interface [31]. In addition, in this study, the modeled section of the mandible was composed as a cancellous core surrounded by 2 cm homogenous cortical layer. An actual mandible has more compact bone at the inferior border and less compact bone on the superior border. Therefore, inherent limitations of FEA must be acknowledged.

When applying FEA to dental implants, not only axial and horizontal loads (moment-causing loads) but also a combined load (oblique occlusal load) must be considered because the latter represents a more realilistic occlusal directions [18]. Thus, vertical,

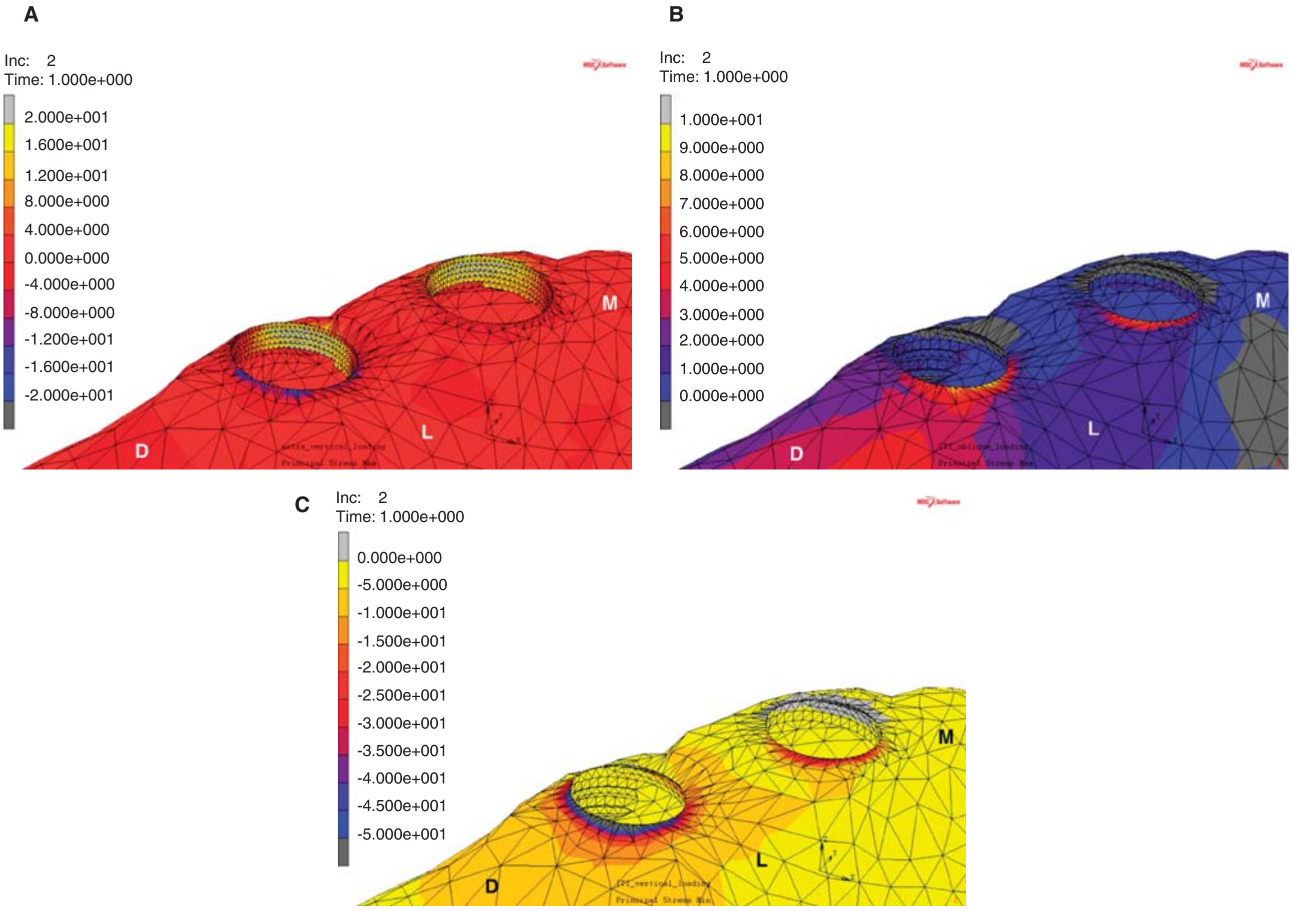

Figure 3. Stress distribution patterns recorded in the cortical bone; (A) P_{max} for the MCS model in the case of vertical load, (B) P_{max} for the NMCS model in the case of oblique load, (C) P_{min} for the NMCS model in the case of vertical load.

horizontal and oblique static loads were applied homogeneously to each crown of the whole system in the present study.

Two different implant systems with two distinct collar geometries have been evaluated in the current study. However, as the two implant system shared mostly an identical fixture–abutment junction, difference in the design of implant collar mainly would lead to distinctive features at the stress distribution in the bone, at the implant-abutment complex, framework and veneering part. In the current study, a 14 mm long implant is used to demonstrate the NMCS and 13 mm long for the MCS, accordingly, and the diameters of the modeled implants were 4.1 and 4.0 mm, respectively. Nevertheless, some clinical studies demonstrated that the success rate of an implant was proportional to the implant length, but some authors have observed that the length of the implant had little influence on the amount of stress level [19]. The current study has been designed to maintain the original product lines of the manufactures as well as to mimic the real clinical situation. Furthermore, one can choose any of these implants to use in the same indication because of the dimensional correspondence between the bench marks.

It was evaluated that 3-unit cantilever FPDs replacing a tooth better distribute occlusal forces than 2-unit FPDs of similar design [5]. It was also shown that short-span FPDs with cantilever extensions represented a predictable treatment modality. Therefore, 3-unit cantilever FPDs with a short-span length were designed in this study.

It was reported that in cantilever prostheses, when the vertical load was applied on the cantilever FPDs, the most distal implants represent the fulcrum and, therefore, this creates a class I lever system, which dramatically alters the direction and magnitude of forces on the abutment [6]. Leverage is one of the primary causative factors for excessive force concentrations which lead to bone loss around implants [20,26]. Therefore, biomechanics of the cantilever FPD designs is an important issue, which is still under investigation in implant dentistry.

In this study, it was evaluated that in the case of vertical load, slightly higher stress concentrations were located at the cortical bone in the MCS model when compared with the NMCS model. With the microthreaded collar design, the stresses resulting from vertical load tend to be concentrated within the implant collar immediately below the crestal

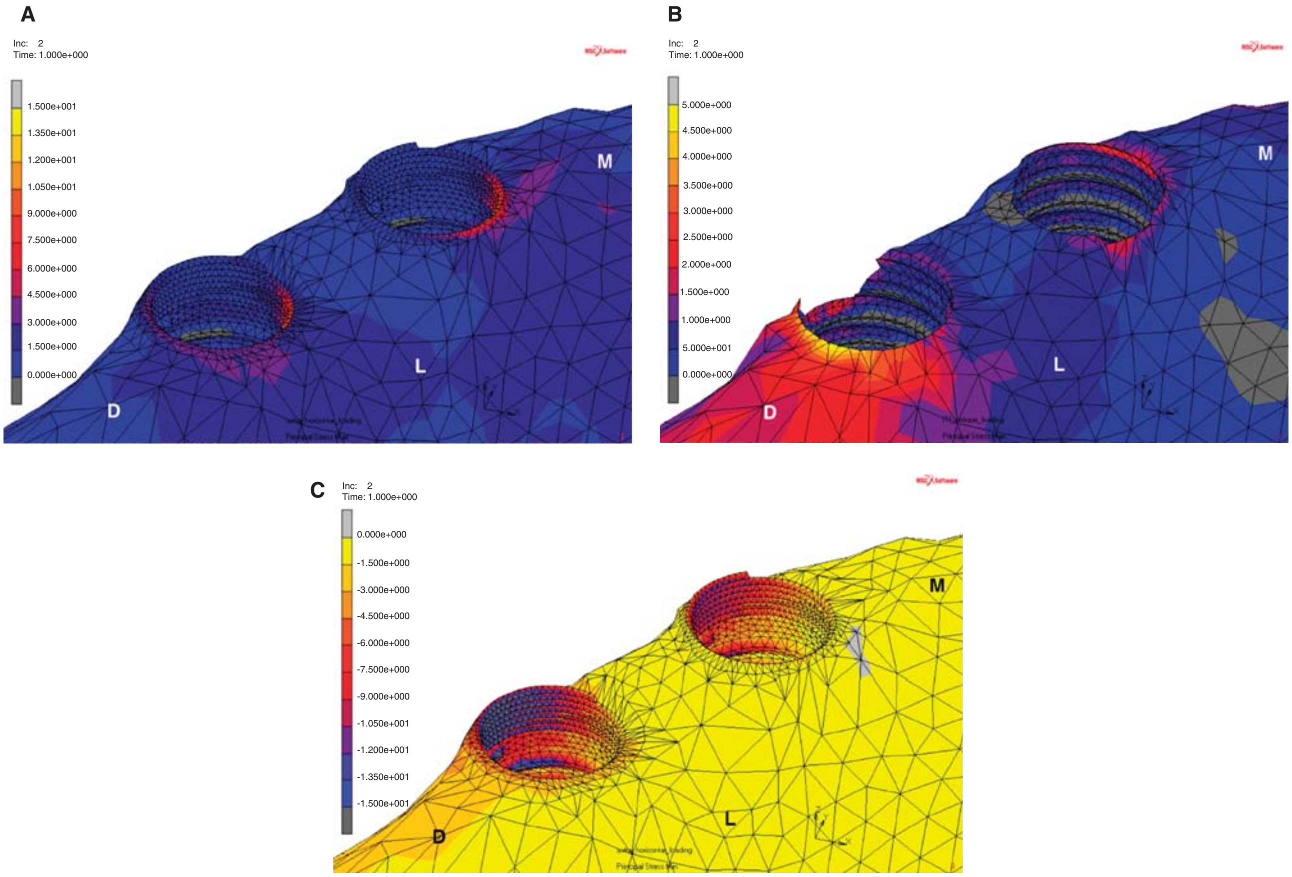

Figure 4. Stress distribution patterns recorded in the cancellous bone; (A) $P_{\max}$ for the MCS model in the case of horizontal load, (B) $P_{\max}$ for the NMCS model in the case of oblique load, (C) $P_{\min}$ for the MCS model in the case of horizontal load.

bone. The same tendency was also reported in other FEA studies [14,17,19]. This location is consistent with findings from experiments and clinical studies that demonstrated that bone loss begins around the implant neck [1,2,7,11]. The reduction of the available bone volume at the cervical region of cortical bone may have a strong influence on the stress around the implants [32].

In the case of horizontal and oblique load the stresses concentrated at the cortical bone in the NMCS model were higher than the stresses in the MCS model. This result may be explained by

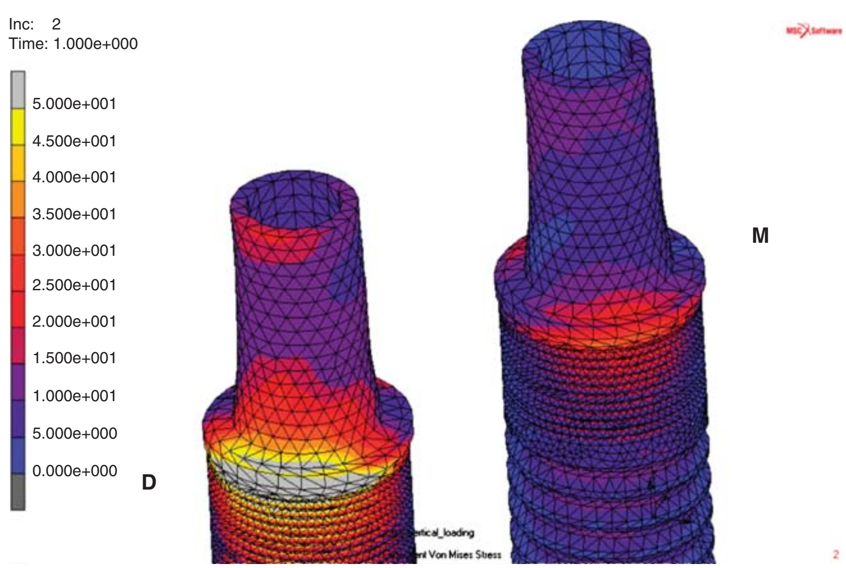

Figure 5. von Mises stress distribution patterns recorded in the implant–abutment complex for the MCS model in the case of vertical load.

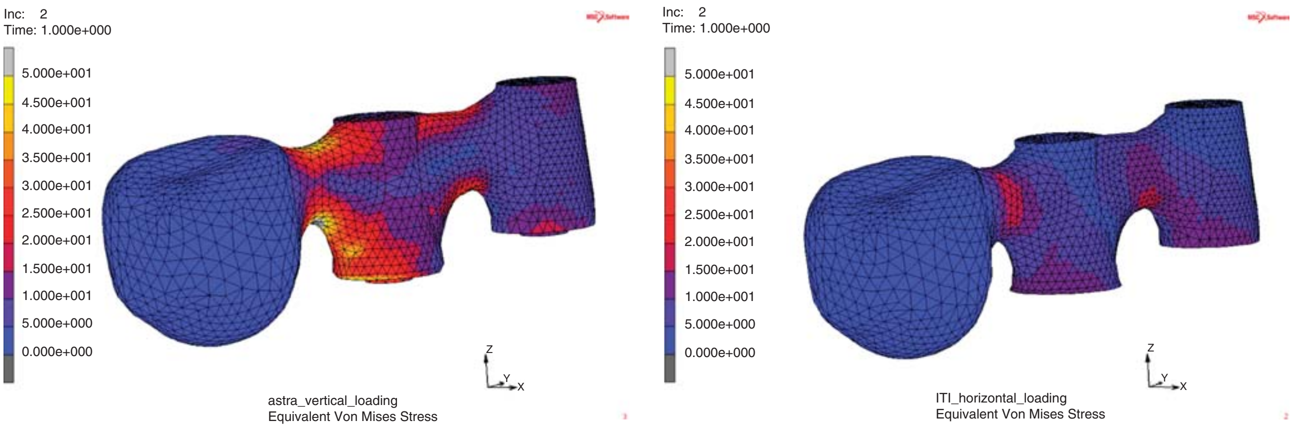

Figure 6. von Mises stress distribution patterns recorded in the framework; (A) MCS in the case of vertical load, (B) NMCS in the case of horizontal load.

the effects of microthreads as retention elements in the cortical bone. Microthreaded collar structure does not allow a sliding motion between implant and bone, therefore the implant and bone complex act as a body under the horizontal load case.

The yield stress of cortical bone has been reported as 130 MPa [15]. Maximum equivalent stress in this study in the cortical bone (64.9 MPa) is lower than this value, which revealed no area to be critically loaded with respect to mechanical stresses. This fact yields to the result that the analysis type is rightfully selected to be an elastic one, since the equivalent stresses of the other bodies are also lower than the corresponding yield stresses.

It was shown that, in the NMCS model, the stresses in the cancellous bone were generally lower than the stresses in the cortical bone. The stresses induced by the occlusal loads might immediately be transferred from implant to the cervical cortical bone, whereas much lower remaining stress may spread over the cancellous bone at the apical region. However, in the MCS model, the stresses at the cancellous bone caused by horizontal and oblique loads were higher than the stresses at the cortical bone. This might be due to the effect of the stress transfer properties of implants with microthreaded collar geometry which support 3-unit cantilever FPDs. It can be concluded that the microthread collar structure under 3-unit cantilever FPDs decrease the stress values in cortical bone under the horizontal and oblique load when compared with non-microthread collar geometry, whereas the stress values are higher in cancellous bone.

It is well known that cortical bone, having a higher modulus of elasticity than cancellous bone, is stronger and more resistant to deformation [15]. The stresses generated by the mechanical loads during the function of implants with microthreaded collar geometry, which support cantilever FPDs, may have influences on the resorption of cancellous bone which support cantilever fixed partial dentures. The effects of the above-mentioned factor on the stress distribution in the bone should be critically analyzed further in future studies.

In a clinical study, comparing implant-supported FPDs with and without cantilever extensions,

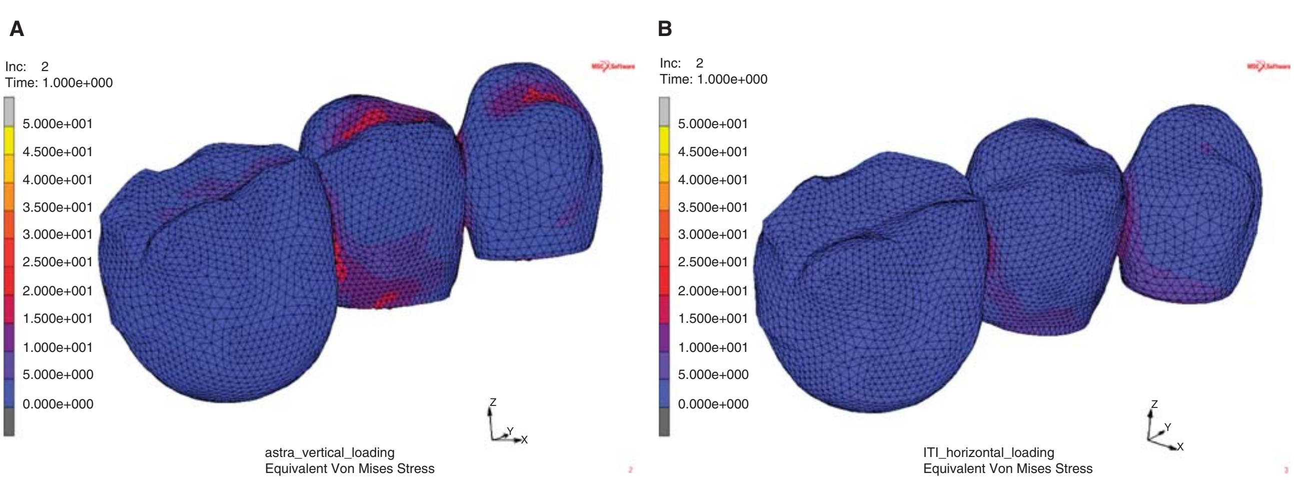

Figure 7. von Mises stress distribution patterns recorded in the veneering part; (A) MCS in the case of vertical load, (B) NMCS in the case of horizontal load.

Romeo et al. [7] reported an overall implant survival rate of 97% and a prosthesis success rate of 98% during a follow-up period of 1–7 years. The survival rates were similar for both treatments and, hence, it was concluded that implant-supported FPDs with cantilever extensions represented a predictable therapy.Moreover, findings from *in vivo* studies revealed that the enhanced stress occurred mainly at the bone crest adjacent to the distal surface of the implant that was facing the cantilever extension [2,7,15]. The current study results confirmed that the cortical bone around the implant facing the cantilever extension was the most stress-bearing portion of the system independently of the collar geometry. However, another point that must be stressed is the direction of the affecting chewing forces and the type of the resultant stress. The results of our study show that there were no differences between the NMCS and MCS models in terms of stress localization and pattern in both models, while the magnitude of the stress values varies in the models. Nevertheless, under vertical, oblique and horizotal load conditions, the compressive stress concentration ($P_{\min}$) has been noted on the distal, distolingual and lingual cortical surroundings, respectively, at the neighboring implant next to the cantilever extension.

The stresses in implants, abutments, framework and veneering structure are also important to evaluate because these structures, being the stiffest components of an implant prosthodontic system, bear a great amount of stress and are responsible for transmitting the load to the bone. A well-planned and well-executed prosthesis is essential to avoid excessive and unnecessary forces in the bone and implant components. Mechanical integrity of the prosthesis and implant combined to the framework rigidity is essential to the longevity of the prosthesis [6,7].

In the current study, higher von Mises stresses have been noted in NMCS models at the veneering material. The reported most frequent technical complications with cantilevered FPDs included veneer fractures due to increased bending movement at the junction of the pontic and retainer [13]. However, stresses at the veneering material decreased with microthreaded collar implant design.

With fixed partial dentures, acrylic resins and composites fractured more often than porcelain [14]. Recent studies where porcelain was used instead of resin are also reporting fractures, but they occur less frequently than with resin. Davis et al. [10] proposed that a porcelain superstructure had a stiffening effect on the prostheses and a more evenly distributed force to framework.

In the present study, higher von Mises stresses have been noted for MCS models in the implant–abutment complex under the vertical load, whereas NMCS models had higher von Mises stresses in the framework in the case of vertical load. When compared the stresses in veneering materials, lesser von Mises stresses have been noted in the MCS model. These results were due to the difference in load transfer mechanisms depending on the collar design of the implants. Since these stresses at the framework and veneering material are lower than the yield strengths of the materials (cobalt–chromium alloy = 720 MPa, Feldspathic Porcelain = 500 MPa) they do not seem to have any clinical relevance on the mechanical as well as the biomechanical outcome [15,23].

The study has demonstrated that the stress values are concentrated on connectors at the framework and veneering part, which is in agreement with the previous studies [33,34]. This behavior may be caused by the reduced cross-section of the area. Abutment and pontics are united by a connector that satisfies certain structural requirements. It must provide enough strength to resist the forces of occlusion that cause flexure of the connector, the interface and the abutment casting [35].

Conclusions

This study is part of a series of ongoing FEA studies focused on stress distribution around the microthread and non-microthread collar geometries with various implant retained prosthesis designs. Implant supported cantilever FPDs have been studied in the current study. It is concluded that collar geometry affects the stress transfer properties of implants which support 3-unit cantilever FPDs. Both of the implant designs permit the distribution of the functional forces to the bone and prosthesis without overloading the bone/implant interface and the prosthetic structures of an implant supported 3-unit cantilever FPDs. As a simple and economic procedure, edentulous ridges next to implants may be reconstructed by means of cantilevers instead of pre-implant reconstructive or regenerative procedures.

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