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Finite element stress analysis of Ti-6Al-4V and partially stabilized zirconia dental implant during clenching

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Abstract

Objective. The purpose of this paper is to compare the differences in stress between Ti-6Al-4V and PS-ZrO_2 dental implant during clenching and whether these changes are clinically significant to limit the use of zirconia in oral implantology. ***Materials and methods.*** The model geometry was derived from position measurements taken from 28 diamond blade cut cross-sections of an average size human adult edentulous mandible and generated using a special sequencing method. Data on anatomical, structural, functional aspects and material properties were obtained from measurements and published data. Ti-6Al-4V and PS-ZrO_2 dental implants were modelled as cylindrical structure with a diameter of 3.26 mm and length of 12.00 mm was placed in the first molar region on the right hemimandible. ***Results.*** The analysis revealed an increase of 2–3% in the averaged tensile and compressive stress and an increase of 8% in the averaged Von Mises stress were recorded in the bone–implant interface when PS-ZrO_2 dental implant was used instead of Ti-6Al-4V dental implant. The results also revealed only relatively low levels of stresses were transferred from the implant to the surrounding cortical and cancellous bone, with the majority of the stresses transferred to the cortical bone. ***Conclusion.*** Even though high magnitudes of tensile, compressive and Von Mises stresses were recorded on the Ti-6Al-4V and PS-ZrO_2 dental implants and in the surrounding osseous structures, the stresses may not be clinically critical since the mechanical properties of the implant material and the cortical and cancellous bone could withstand stress magnitudes far greater than those recorded in this analysis.

Key Words: *finite element analysis, FEA, mandible, dental implant materials, clenching, metallic dental implant, ceramic dental implant, partially stabilized zirconia, titanium, Ti-6Al-4V, PS-ZrO_2*

Introduction

Dental sub-gingival implants have been used to support fixed partial dental prostheses for many decades, but they have not always enjoyed a favourable reputation. This situation has changed dramatically with the development of endosseous osseointegrated dental implants [1]. They are the nearest equivalent replacement to the natural tooth and are therefore a feasible addition in the rehabilitation of edentulous patients because of the periodontal diseases, trauma or developmental anomalies. Various materials, mainly Ti and its alloys, and related techniques have been introduced to allow bone growth onto dental implants such as barrier membranes [2,3], hydroxyapatites [4] and grit-blasting techniques [5]. Most of these studies are able to report complete bone fill of the eventual gaps clinically.

There are a number of dental implant systems which offer predictable long-term results backed by good scientific research and clinical trials. Interest in the use of dental implants has increased steadily over the last three decades. This interest has stimulated the development of many new dental systems and different implant designs. Modern engineering techniques have been used to design implant types through the use of computer-based models using computer-aided design systems [6–13].

Titanium and its ternary alloys, such as Ti-6Al-4V, based on their physical and chemical properties, appear to be especially suitable for dental implants and prostheses for more than 25 years [14]. The passivating oxide on the implant surface allows close apposition of hard and soft tissues, physiological fluids and proteins to the titanium surface. This process, whereby living tissue and an implant become

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structurally and functionally connected, is referred to as osseointegration [15]. Titanium also has been successfully used as a biocompatible implant material and continual improvements in both device design and clinical implantation techniques have led to well-accepted and predictable procedures.

Ceramic materials can successfully replicate the esthetic qualities of natural teeth. However, despite their strength under compression, ceramic materials are brittle and have limited tensile strength. Ceramic materials also do not exhibit any significant deformation before failure, unlike metals [16].

Zirconia has growingly become a material of choice in implant dentistry, not only as the dental implant abutments, but as whole dental implants [17]. Zirconia, a ceramic material, is the fully oxidized form of zirconium and, depending on the temperature, can exist in several phases. In recent years, developments have led to the introduction of new dental implants manufactured using advanced ceramic materials such as zirconia. Zirconia-based ceramic materials have properties that lend themselves to the use in the mouth as it is indistinguishable from natural teeth as zirconia is white in colour [17,18]. Zirconia also possesses several advantages which make it favourable as dental implants, such as high corrosion resistance, and they can inhibit crack growth and prevent catastrophic failure [16].

The clinical success of an implant is largely determined by the manner in which the mechanical stresses are transferred from the implant to the surrounding bone without generating forces of a magnitude that would jeopardize the longevity of implant and prosthesis [19].

The function of a dental implant system is to restore dentition by providing a means of transmitting masticatory forces to the mandibular or maxillary bone. The importance of understanding the way in which the stresses and distortion acting in the dental implants and its surrounding bone structure are distributed is of paramount importance in the field of prosthetic replacement where the principal aim is to replace a damaged tooth so that the patient can function effectively.

Bone tissue is known to remodel its structure in response to applied stress. Variations in the internal state of stress in bone determine whether constructive or destructive bone remodelling takes place. Low stress levels around a dental implant may result in disuse atrophy similar to the loss of alveolar crest after the removal of natural teeth. On the other hand, abnormally high stress concentrations in the supporting tissues can result in patient discomfort, pressure necrosis and the eventual failure of the implant system [20].

Photoelasticity, finite element analysis and experimental strain measurements have been used to determine the state of stress in tissues around dental implants [6,7,12,21–26].

Three-dimensional finite element analysis was used by Degerliyurt et al. [27] to compare stress distribution around the endosseous titanium implants using three different implant fixture geometries. Loads were applied to each of these fixtures: vertically 70 N, with an inclination of 60° obliquely (buccolingually) 35 N and horizontally (mesiodistally) 14 N. Tensile and compressive stresses on each simulated mandible were calculated using finite element analysis software.

Caglar et al. [28] evaluated the von Mises, compressive and tensile stresses occurring on three different zirconia dental implants and surrounding bone with three-dimensional finite element analysis. Oblique loading of 178 N and horizontal loading of 25.5 N were applied. They concluded that the von Mises, compressive and tensile stresses occurring in the cortical bone were higher than those observed in trabecular bone.

Stress distribution around two prosthesis-implant systems was evaluated by Cruz et al. [29] using three-dimensional finite element analysis. Implants were arranged in either a straight-line or an intra-bone offset configuration. The systems were modelled with three titanium implants placed in the posterior mandible following a straight line along the bone. An axial load of 100 N and a horizontal load of 20 N were applied on the centre of the crown of the middle implant.

The design optimization of a tooth implant-supported fixed prosthesis was investigated theoretically by Dargahi et al. [30]. A three-dimensional finite element analysis was utilized to simulate the stress distribution and deformation, with an emphasis on the material selection for various parts of the prosthesis. This mandibular prosthesis was supported by six implants. The properties of three different materials for implants and four different materials for framework were incorporated into 12 different models.

A detailed survey of the literature suggested that no comprehensive study has been carried out to compare the stresses in and around a partially stabilized zirconia (PS-ZrO_2) dental implant and a Ti-6Al-4V dental implant during functional loading and whether these stresses are clinically significant to limit the use PS-ZrO_2 as an implant material.

In the past the main areas of study into the stresses and distortions acting in and around the implants have been centred on a simple type of investigation, in which the investigators have recorded the stress and distortion with a single force applied to the implant. No reaction forces generated by the muscles of mastication were utilized in the calculations of stresses and distortions.

There have been various reports on two-dimensional and three-dimensional finite element analysis of the stresses and distortions acting in and around dental implants. These models appear to be very remote from the actual anatomical and biomechanical conditions.

The aim of this paper is to gain a fundamental insight into the biomechanical behaviours of PS-ZrO_2 and Ti-6Al-4V dental implants inserted into the human mandible using three-dimensional finite element analysis (FEA). A fundamental insight into the biomechanical behaviour of this system and the functional aspects is a necessity if appropriate surgical techniques and prosthetic devices are to be further developed.

Materials and method

Finite element mandible model

A dry human mandible has been used to define the geometry of the model. The mandible was cross-sectioned vertically into 28 sections. Each cross-section of the bone was divided into five sections, the outer four representing the cortical bone and the inner one the cancellous bone. For this analysis three nodes on the symmetry plane were fixed in space by the use of spring elements.

Cylindrical dental implant with a diameter of 3.26 mm and length of 12.00 mm was modelled as solid structures with abutment that was 2 mm high. The depth of the implant-cortical interface varies from location to location on the mandible model. For this study, the implant was placed in the first molar region on the right hemi-mandible and has a depth of 2 mm.

For finite element analysis, the mesh on the reconstructed section is very difficult to interconnect with the mesh on the mandible model. In order to overcome the connectivity problem, coupling links were used.

For this analysis, eight coupling links were used to reconnect the cortical and cancellous bone on the reconstructed section to the mandible model. After connecting the implant to the surrounding bone structure, the three-dimensional finite element model was completed. The model consisted of 700 solid elements and 3738 nodes (Figure 1).

Muscle forces

The muscle forces considered in this analysis were selected from various investigators [31–34]. All the forces were assumed to be symmetrical with respect to the mid-line and to have an equal magnitude on the right and left side of the mandible. The forces exerted by contracting muscles were represented by vectors. Similar assumptions were made by a number of workers [35–37]. This assumption is reasonably true when the muscle is homogeneous and acts as a whole.

The direction of these vectors can be defined by the connecting lines between the centroids of the origins and the insertions of the muscles. Descriptions of these areas of attachments were derived from the anatomical literature [31,33,38–40] and three-dimensional observations and measurements on different skulls.

For this present work, the bite force (clench force) was directed at an angle of ~85° to the averaged occlusal plane. Each bite force was equally divided between two teeth symmetrically placed with respect

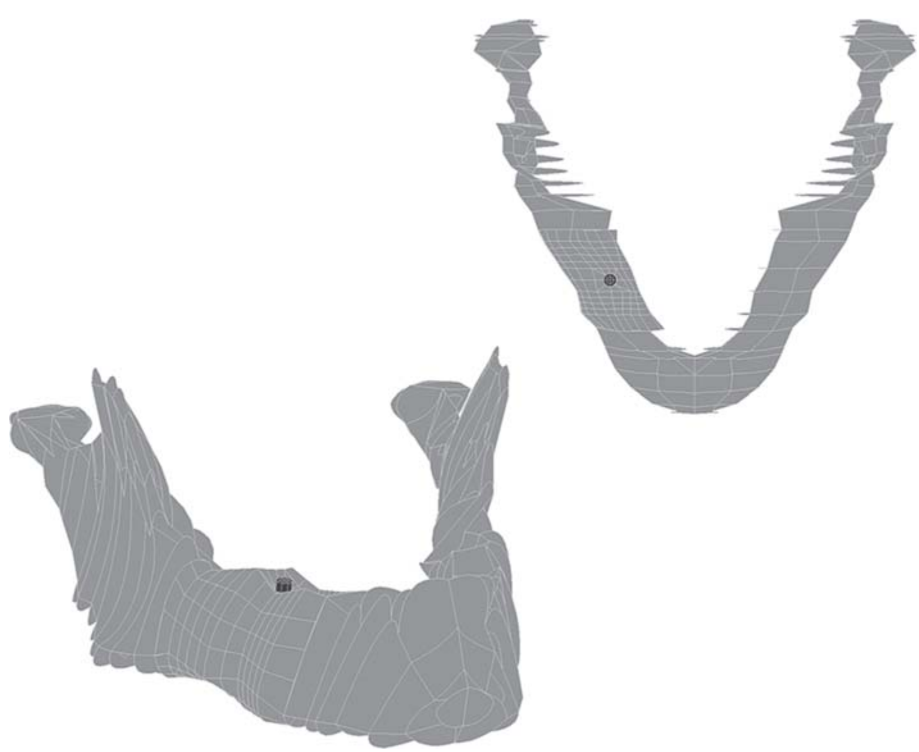

Figure 1. Front view of the finite element model of the human mandible with the insertion of a dental implant (produced in STRAND7).

Table I. Calculated muscle, joint reaction and bite force magnitudes (N) acting on the mandible during clenching.

Muscles	Force (N)
Masseter	340.0
Temporalis^{Anterior}	264.3
Temporalis^{Posterior}	264.3
Medial Pterygoid	191.4
Lateral Pterygoid	378.0
Joint Reaction Force	471.9
Bite Force (2nd premolar)	246.3
Bite Force (1st molar)	157.4
Openers	155.0

to the midsagittal plane, thus simulating Pruim et al.'s [34] experiment.

During clenching all muscles were assumed to be active. Calculated muscle forces are given in Table I. The reaction forces were assumed to be acting at the centre of the condyles (Figure 2).

Material properties

Data on the material properties of mandibular cortical and cancellous bones were taken from published data which were determined on small specimens obtained from the mandibles of cadavers by means of ultrasonic wave methods [41–43] and other material testing techniques [44,45]. The cortical and cancellous bone of the mandible can be considered to be transversely isotropic, with a higher elastic modulus in the longitudinal direction and a lower elastic modulus in all transverse directions. Therefore, all the individual elements for both cortical and cancellous bone in our model were represented as transversely isotropic.

In this analysis, the Young's Modulus and Poisson's ratio for the cortical bone were obtained from Arendts and Sigolotto [44,45] and cancellous bone were obtained from Turner et al. [46]. The material properties of the cortical and cancellous bone, Ti-6Al-4V and PS-ZrO_2 used for the finite element computation are shown in Tables II and III.

Results

Dental implant

The maximum tensile, compressive and von Mises stress recorded on the Ti-6Al-4V and PS-ZrO_2 dental implant is given in Table IV. The analysis revealed an increase of 23–27% in the maximum tensile, compressive and von Mises was recorded on the dental

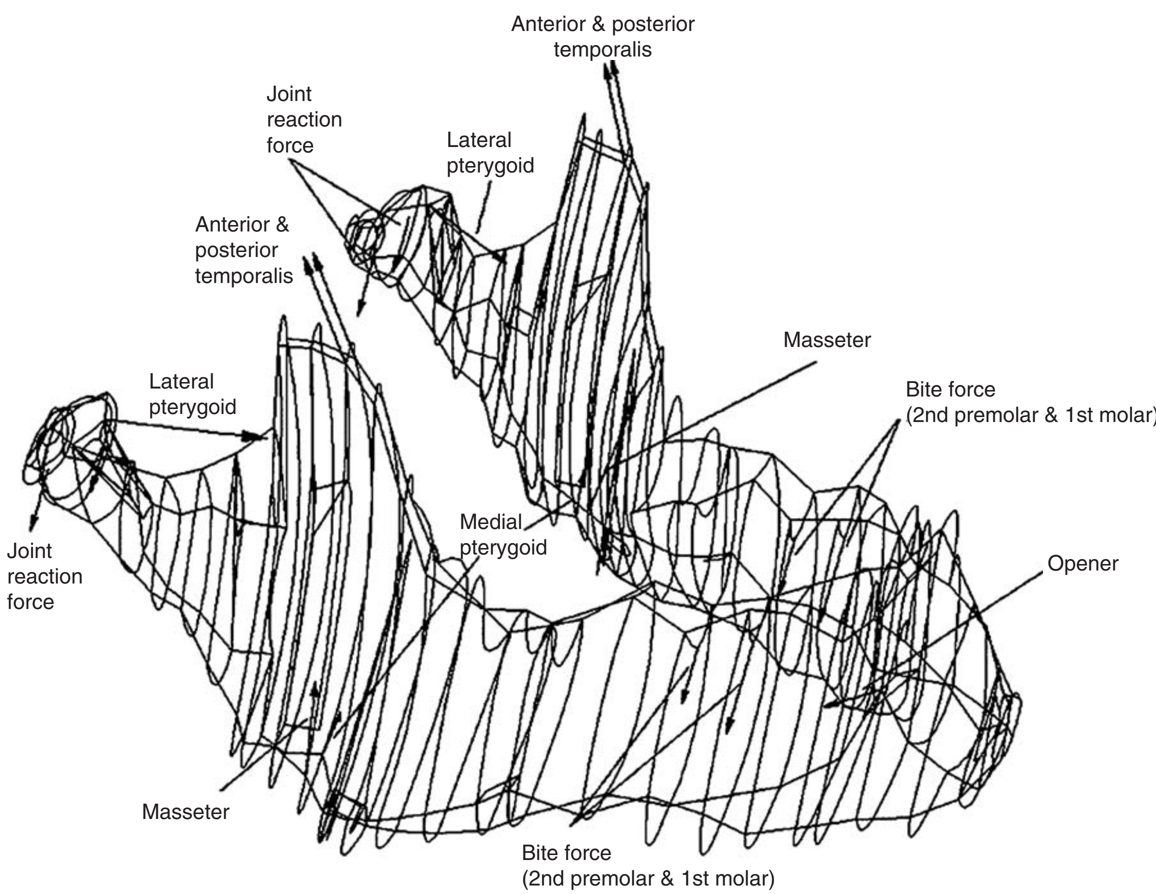

Figure 2. Applied muscle, joint reaction and bite forces and co-ordinate system during clenching.

Table II. Mechanical properties used in finite element computations.

	Young's Modulus (GPa)			Poisson's Ratio			Tensile strength (MPa)	Compressive strength (MPa)
	E_1	E_2	E_3	ν_{12}	ν_{23}	ν_{13}		
Cortical bone	6.9	8.2	17.3	0.315	0.325	0.310	77–169	130–160
Cancellous bone	0.32	0.39	0.96	0.3	0.3	0.3		

implant when PS-ZrO$_2$ is used as the implant material.

The distribution of tensile, compressive and von Mises on the Ti-6Al-4V dental implant is shown in Figure 3. Tensile stress is most evident at the back of the implant, while compressive and von Mises stresses are most evident towards the front of the implant. The result also shows that similar locations of tensile, compressive and von Mises stress were produced on the PSZrO$_2$ dental implants.

Osseous structures

The averaged tensile, compressive and von Mises stresses obtained in the bone–implant interface around the Ti-6Al-4V and PS-ZrO$_2$ dental implants are shown in Table V. The values were obtained by averaging the sum of the tensile, compressive and von Mises stress recorded at the implant–bone interface. The tensile, compressive and von Mises stress recorded in the same region without the insertion of dental implant was also included as a reference.

The result shows when a PS-ZrO$_2$ dental implant is used, an increase of 2–3% in the averaged tensile and compressive stress and an increase of 8% in von Mises stress was recorded in the bone–implant interface. The result also indicated that a decrease of 27–45% in the averaged tensile, compressive and von Mises stress was recorded when a dental implant is incorporated into the finite element model. The decrease in the average tensile, compressive and von Mises stress can be attributed to the Ti-6Al-4V and PS-ZrO$_2$ dental implants being much stiffer than the cortical and cancellous bone.

The distribution of tensile, compressive and von Mises stress in the osseous structure around the Ti-6Al-4V dental implant is shown in Figure 4 with the use of contour plots. The contour plots revealed the tensile and von Mises stress mainly acted on the buccal side of the bone. Tensile stress is also observed at the molar bite point around the dental implant. The compressive stress acted mainly on the lingual side of the osseous structure around the Ti-6Al-4V dental implant. The result also indicated that the majority of stresses are transferred to the cortical bone instead of cancellous bone.

It is also noteworthy that the stress that was transferred to the cortical and cancellous bone from the implant was relatively low, i.e. for example, 14 MPa was recorded on the PS-ZrO$_2$ dental implants; however, only 3.6 MPa was recorded at the bone–implant interface. This can be explained by the fact that a large portion of the bite force has been spent on deforming the implant; as a result, only a small portion of the bite force will be transferred from the implant to the surrounding cortical and cancellous bone structures.

Mandible structure

The maximum tensile, compressive and Von Mises stresses recorded at the condylar region of the mandible with the insertion of Ti-6Al-4V and PS-ZrO$_2$ dental implants is shown in Table VI. The result indicates that there was no change in the maximum tensile, compressive and Von Mises stress when different implant materials were used during clenching.

Discussion

The aim of this study was to evaluate the biomechanical behaviours of Ti-6Al-4V and PS-ZrO$_2$ dental implants inserted into the human mandible during clenching, using a three-dimensional anatomically realistic finite element model.

The finite element method is used to precisely calculate local stress–strain distributions in geometrically complex structures. The predictive accuracy of the finite element model is influenced by the geometric detail of the object to be modelled, the material properties and the applied boundary conditions. Finite element analysis has become widely used in all biomechanical fields, especially for assessing

Table III. Mechanical properties of the dental implants used in this study.

	Young's Modulus (GPa)	Compressive strength (MPa)	Tensile strength (MPa)	Poisson's ratio
Titanium-Aluminium-Vanadium (Ti-6Al-4V)	114	450–1850	900–1172	0.34
Partially-stabilized Zirconia (PS-ZrO$_2$)	210	1990	800–1500	0.31

Table IV. Maximum tensile, compressive and Von Mises Stress recorded on the Ti-6Al-4V and PS-ZrO_2 dental implants.

Dental implant	Tensile stress (MPa)	Compressive stress (MPa)	Von Mises stress (MPa)
Ti-6Al-4V	11.02	−12.39	11.37
PS-ZrO_2	14.00	−15.30	14.20

stresses and strains in dental implants and the surrounding bone structures as well as in normal bone.

Both principal stresses and Von Mises stress has been used by various investigators in the past to assess the stresses observed on the implant during loading tasks. Principal stress (tensile and compressive) values are important for composite materials such as bone, because failure occurs when both the tensile and compressive stress is greater than or equal to the ultimate tensile or compressive strength of bone.

In the last 20 years, many different types of implants have been used in prosthodontics for the replacement of missing teeth (tooth replica implants, subperiosteal, endosseous and endodontic-endosseous implants).

Finite element stress analysis reveals, although a high magnitude of tensile, compressive and Von Mises stress was observed on the Ti-6Al-4V and PS-ZrO_2 dental implants, these stresses may not be clinically critical since the mechanical properties of the implant material itself could withstand stress magnitudes far greater than those recorded in this analysis (Table III).

The results obtained from the bone–implant interface also revealed that even though an increase in the tensile, compressive and Von Mises stress were recorded when PS-ZrO_2 dental implant was used instead of Ti-6Al-4V dental implant, again clinically, the stresses may not be significance since the ultimate tensile and compressive strength of human cortical bone is far greater that the stress levels experienced during clenching (Table II).

The result also indicated that the stresses recorded on the mandible with the inclusion of implants remained unchanged when Ti-6Al-4V and PS-ZrO_2 dental implant was used.

Forces on the prosthesis, for example, during chewing, will be transferred to the implants and this will lead to stress in the bone surrounding the implants. Bone tissue is known to remodel its structure in response to mechanical stress. Variations in the

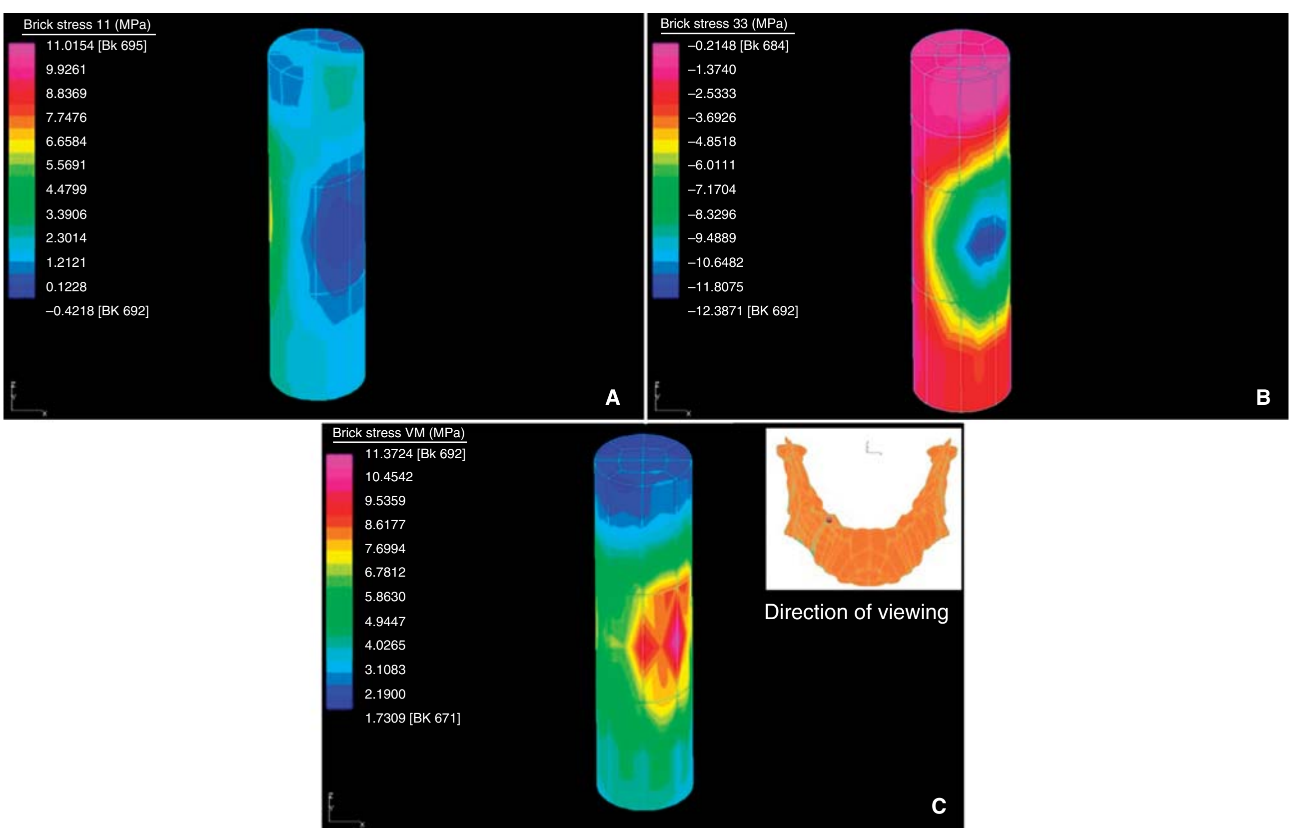

Figure 3. (A) Tensile, (B) Compressive and (C) Von Mises stress observed on the Ti-6Al-4V dental implant (front view).

Table V. Averaged tensile, compressive and Von Mises stress (MPa) recorded at the bone–implant interface around the Ti-6Al-4V and PS-ZrO_2 dental implants.

Dental implant	Tensile stress (MPa)	Compressive stress (MPa)	Von Mises stress (MPa)
Ti-6Al-4V	3.55	−3.68	5.22
PS-ZrO_2	3.64	−3.76	5.63
No implant	6.46	−5.59	7.75

internal state of stress in bone determine whether constructive or destructive remodelling will take place.

Bone resorption and stress shielding will occur if no load transfer is achieved during functional movements. On the other hand, abnormally high stress concentration in the supporting tissues can result in failure of the implant. The long-term function of a dental implant system will depend on the biomechanical interaction between bone and implant.

The clinical success of an implant is largely determined by the manner in which the mechanical stresses are transferred from implant to the surrounding bone without generating forces of a magnitude that would jeopardize the longevity of implant and prosthesis. This is achieved through osseointegration.

Several factors are involved in achieving osseointegration. They include metal composition, suitable implant geometry, absence of overheating during site preparation and adequate bone quality.

During various functional movements the dental implant could be subjected to an alternating load oscillating through zero. The result is micromovements between the implant and the bone, which do not lead to osseointegration, but, rather, fibrous encapsulation. This micromovement causes an increase in loosening of the implant. This condition, and subsequent fibrosis, greatly facilitates infection.

The aim of this study was to produce and obtain more insight into the influence of the design of a subgingival dental implant system during functional movement on the stress and distortion distribution on the implant, as well as in the surrounding mandibular bone. Knowledge of the changes in stresses and distortion acting on the implant and in the surrounding bone structure and the design parameters that could influence these changes might lead to better dental implant design and case selections.

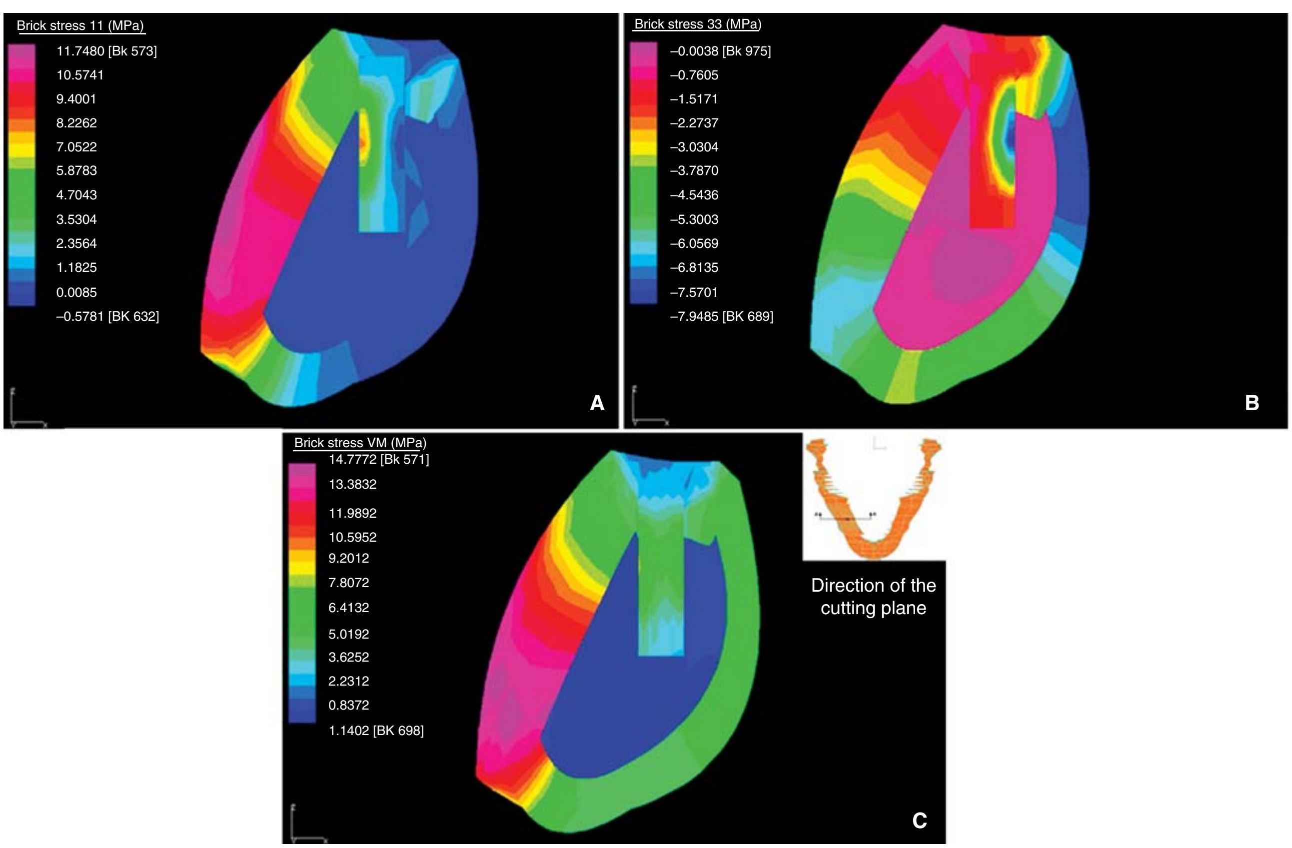

Figure 4. (A) Tensile, (B) Compressive and (C) Von Mises stress in the osseous structure around the Ti-6Al-4V dental implant (cross-sectional frontal view).

Table VI. Maximum tensile, compressive and Von Mises stress (MPa) recorded at the condylar region of the mandible with the insertion of Ti-6Al-4V and PS-ZrO_2 dental implants.

Dental implant	Tensile stress (MPa)	Compressive stress (MPa)	Von Mises stress (MPa)
Ti-6Al-4V	27.24	−36.23	38.91
PS-ZrO_2	27.24	−36.23	38.91
No implant	27.24	−36.23	38.91

Conclusion

More understanding and insight into the influence of implant material during functional movement on stress and distortion in and around the implant has been obtained. These results can be used as a comparison to determine the clinical feasibility and survivability of ceramic and metallic dental implants.

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