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Evaluation of Platform Switching on Crestal Bone Stress in Tapered and Cylindrical Implants: A Finite Element Analysis

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Abstract

Purpose: To analyze and compare the stress distribution around tapered and cylindrical implants and investigate how different abutment diameters influence crestal bone stress levels.

Materials and Methods: Six finite element models of an abutment (5 mm, 4.3 mm, and 3.5 mm in diameter) and supporting implants (tapered and cylindrical) were designed. A vertical force of 100 N and a 15-degree oblique force of 100 N were applied separately on the occlusal surface, and von Misses stresses were evaluated in the cortical and cancellous bone.

Results: Higher stress was observed under oblique loading than under vertical loading of both tapered and cylindrical implants. Tapered implants demonstrated more stress under both vertical and oblique loading. Platform switching reduced peri-implant crestal bone stress in all models under vertical and oblique forces. The peri-implant crestal bone around tapered implants experienced 4.8% more stress under vertical loading and 35% more stress under oblique loading in comparison to bone around cylindrical implants (2.62 MPa with vertical loading, 8.11 MPa under oblique loading). Oblique loads resulted in much higher stress concentrations in the peri-implant crestal bone than vertical loads (238% in cylindrical and 308% in tapered implants). When the abutment diameter decreased, both models showed reductions of stress in the crestal bone under both types of loading.

Conclusion: In this finite element analysis, tapered implants increased crestal bone stress upon loading, and platform switching minimized the stress transmitted to the crestal bone in both tapered and parallel wall implants.

Key words: Finite element analysis, platform switching, dental implants, crestal bone stresses, implant loading, stress distribution

Introduction

The peri-implant bone level has been used as a criterion to determine the success of dental implants (Albrektsson *et al.*, 1986; Albrektsson and Isidor, 1994; Misch *et al.*, 2008; Smith and Zarb, 1989; Buser *et al.*, 1990; Roos *et al.*, 1997) because a stable bone level is an important prerequisite for preserving the integrity of the gingival margins and interdental papillae (Tarnow *et al.*, 1992; Choquet *et al.*, 2001). Implant success is typically assessed by serial radiographs at 1-year intervals from the date of implant placement. If the observed marginal bone loss is less than 1.5 mm in the first year and less than or equal to 0.2 mm in subsequent years, the implant can be considered successful (Albrektsson and Isidor, 1994). When the patient undergoes stage-two surgery (abutment placement) or if the abutment is

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placed immediately after the implant, which may expose the implant to the oral environment, peri-implant bone remodeling is initiated. The remodeling process involves marginal bone resorption, which is influenced by one or more of the following factors: (a) surgical trauma (Becker *et al.*, 2005); (b) excessive loading (Kim *et al.*, 2005); (c) the location, shape, and size of the implant-abutment microgap and its microbial contamination (Hermann *et al.*, 2001; Weng *et al.*, 2008; Ericsson *et al.*, 1995); (d) the biologic width and soft tissue considerations (Myshin and Wiens, 2005; Berglundh and Lindhe, 1996); (e) a peri-implant inflammatory infiltrate (Broggini *et al.*, 2006); (f) micromovements of the implant and prosthetic components (Hermann *et al.*, 2001; King *et al.*, 2002); (g) repeated screwing and unscrewing of prosthetic components (Abrahamsson *et al.*, 1997); (h) the geometry of the implant neck (Bratu *et al.*, 2009); and (i) the infectious process (Roos-Jansåker *et al.*, 2006). Hence, modifications to implant designs are now focused on reducing the bone stress around implants. In addition, changes in the design of the connection between an abutment and an implant, such as restoring an implant with a smaller-diameter abutment (the "platform-switching" concept), using microthreads at the coronal portion of the implant body (Schrotenboer *et al.*, 2008), and augmentation of implant diameter and/or length (Cynthia *et al.*, 2005), have also been suggested for this purpose.

Platform switching, which was introduced in 1991 (Lazzara and Porter, 2006), is an uncomplicated and effective means to control circumferential bone loss around dental implants. It also has the advantage of acceptable responses from hard and soft tissue. Thus, implants restored with a platform switch could be used for esthetic and biologic purposes (Canullo *et al.*, 2011). In addition, it is assumed that tapered implants have some advantages over implants with a cylindrical shape, for example, in challenging situations such as ridge concavities, extraction sites, and immediate loading or provisionalization, especially in the esthetic zone because of the improved primary stability (Alves and Neves, 2009; Rokn *et al.*, 2011). However, some studies have shown that the tapered implant design actually results in increased crestal bone stress (Cynthia *et al.*, 2005) or at least produces no positive effect on bone level changes (Vigolo and Givani, 2009). Currently, there is insufficient data, either experimental or clinical, regarding bone loss and mechanical stability when platform switching is used together with a tapered implant. Therefore, more studies are needed (Kitamura *et al.*, 2005) to explore this subject.

We aimed to evaluate crestal bone stress around dental implants with mismatched abutments and two different implant body forms (cylindrical and tapered) via finite element analysis (FEA). Finite element models were created to measure the amount of stress in the bone around tapered and cylindrical implants with matched abutments and to compare this with the same implants with mismatched abutments. The main hypothesis was that the crestal bone stresses are higher around tapered implants, and a mismatched abutment can compensate for this greater stress in a positive manner.

Materials and methods

Tapered and cylindrical implants, both with a coronal diameter of 5 mm, were selected to be matched with three different abutment diameters. The implant designs were unique but based on the design of the popular Replace Tapered Groovy (Nobel Biocare), which is threaded and features a rough surface. Six three-dimensional finite element models of an implant, an abutment, and the supporting structures were designed (two different implants x three different abutment diameters). Computed tomographic (CT) images from a volunteer patient, a candidate for implant therapy who provided written consent to use the images in this study (ethical approval 90-03-114-14704 Tehran University of Medical Sciences), were used to design the supporting structures for FEA. Each model consisted of a cancellous bone core surrounded by a cortical bone layer, 1 mm thick. The cortical bone thickness was derived from the patient's CT scans and from the literature (Kunavisarut *et al.*, 2002; Tabata *et al.*, 2010). The models were identical except for the type of implant (cylindrical or tapered) and abutment diameter (platform 1 [P1] = 5 mm [same diameter as the implant], P2 = 4.3 mm, and P3 = 3.5 mm; *Figure 1*).

SolidWorks 2010 (Concord, MA, USA) software was selected for the modeling phase. Then the models were transferred to ANSYS Workbench (version 11.0, ANSYS Inc, Canonsburg, PA, USA) for calculation of stresses. All the vital tissues were presumed elastic, homogeneous and isotropic, and the corresponding elastic properties (Young's modulus and Poisson ratio) were taken from the literature (*Table 1*) (Yang *et al.*, 1999; El Charkawi and El Waked, 1996; Fejerdy *et al.*, 2008; Ash, 2005; Craig and Farah, 1978; Geramy, 2000). Models were meshed and included 117,161 to 201,611 nodes to make up the 10-node quadratic tetrahedral body elements. All nodes at the mesial and distal extremes of the models were restrained so that all rigid body motions were prevented. A vertical force of 100 N was applied at the center of the occlusal surface of the abutment, and an oblique force of 100 N was applied at the occlusal surface with 15 degrees of angulation. Forces of 100 N were chosen because this figure is widely accepted in the literature as comparable to the average magnitude of occlusal force (O'Mahony *et al.*, 2001; Abu-Hammad *et al.*, 2007; Huang *et al.*, 2006; Siegele and Soltesz, 1989).

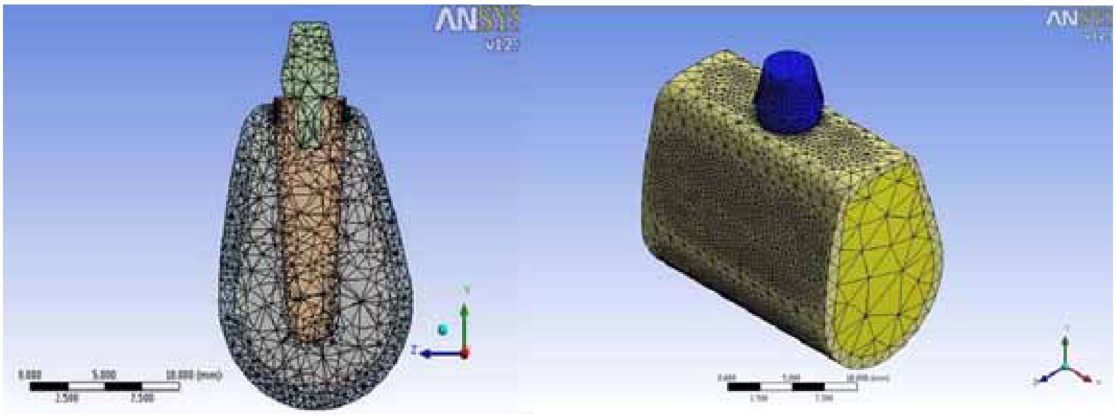

Figure 1. Overall models of the implant and abutment in bone.

Table 1. Mechanical properties of materials used in the models

	Young's Modulus in MPa	Poisson ratio
Implant	96,000	0.36
Cortical bone	34,000	0.26
Cancellous bone	13,400	0.38

The von Misses stresses were evaluated at nine nodes placed at equal distances from coronal to apical (six in the cortical bone and three in the cancellous bone). These nodes were in a plane parallel to the distal extremity of the model after "hiding" the implant. Most studies utilize three nodes in cortical bone and three in the cancellous bone for finite analysis; however, we chose six nodes in cortical bone and three in cancellous bone in this mandibular molar section from CT scans of a real patient to provide more exact data on crestal bone. Also, it is clear from the literature (Misch, 1990; Cochran, 2000) that the maximum stress on bone is mostly at the coronal part of the implant body, especially on the crestal cortical bone. And because the load exerts greater force at early stages of osseointegration we focused more on the cortical part. It was determined that the use of six layers would better distinguish the load distribution in this critical location, since the three layers in the most coronal cancellous part tolerate the greatest amount of load among the cancellous bone that surrounds the implant. Earlier studies found that the peak bone stresses resulting from vertical load components and those resulting from horizontal load components arise at the top of the marginal bone, and that they coincide spatially. These peak stresses are then added together to produce the risk of stress-induced bone resorption (Hansson, 2003).

No statistical analyses were performed because the study was designed to emulate a single imaginary clinical situation, and hence the results were not eligible for statistical analysis.

Results

The average von Misses stresses produced in the cortical bone and the adjacent cancellous bone in the different models are shown in *Tables 2* and *3*. *Figures 1 - 3* summarize the von Misses stresses that occurred in the peri-implant bone under vertical and oblique loading.

Vertical loading

The crestal bone stresses that occurred under vertical loading are shown in *Table 2*.

Cylindrical implant

With the 5 mm abutment (P1), the von Misses stresses ranged from a high of 6.74 MPa and followed a decreasing pattern, reaching 1.70 MPa at the deepest point of the cortical bone. This decrease continued in the cancellous bone to reach a low of 0.74 MPa (*Figure 4A*).

With the 4.3 mm abutment (P2), the stresses were 5.87 MPa at the most crestal portion and again decreased, similar to the pattern seen with the 5 mm abutment, reaching 1.12 MPa at the deepest portion of the cortical bone. The decreases continued and reached 0.52 MPa in the first layers of cancellous bone (*Figure 4A*).

With the 3.5 mm abutment (P3), the stresses in the cortical bone area ranged between 4.30 MPa in the crestal area and 1.00 MPa at the deepest point of the cortical bone. These values reached 0.48 MPa in the cancellous bone (*Figure 4A*).

Tapered implant

Under vertical loading of the tapered implant with the 5 mm (P1) abutment, the von Misses stresses began at 7.16 MPa in the most crestal portion and followed a decreasing pattern, reaching 1.62 MPa in the deepest point of the cortical bone. This decrease continued in the cancellous bone to reach 0.64 MPa (*Table 2, Figure 4B*).

With the 4.3 mm abutment (P2), stresses ranged from a high of 5.64 MPa in the most crestal bone and followed a similar decreasing pattern, reaching 1.49 MPa in the deepest cortical bone. They continued to decrease, reaching 0.59 MPa in the first layers of cancellous bone (*Figure 4B*).

Table 2. The average of von Misses stresses produced in cortical and the adjacent cancellous bone under *vertical load* in tapered and parallel implants with different abutment sizes (P1 = 5 mm abutment; P2 = 4.3 mm abutment; P3 = 3.5 mm abutment) at nine nodes (1-9) placed at equal distances from coronal to apical points (six in the cortical bone and three in the cancellous bone). The amount of stress is most pronounced in the most coronal part of the cortical bone (node 1) and is higher in tapered than cylindrical implants.

	Tapered implant			**Cylindrical implant**		
Bone layer nodes	**P1**	**P2**	**P3**	**P1**	**P2**	**P3**
Amount of stress in cortical bone						
1	7.16	5.63	4.43	6.73	5.86	4.30
2	5.07	3.45	2.50	4.88	4.17	1.69
3	3.71	2.15	1.34	2.70	2.28	1.20
4	2.24	1.82	1.16	2.22	1.99	1.02
5	1.78	1.53	1.11	2.06	1.21	1.07
6	1.62	1.49	1.23	1.69	1.11	0.99
Amount of stress in cancellous bone						
7	0.75	0.77	0.72	0.75	0.57	0.51
8	0.66	0.61	0.73	0.74	0.55	0.50
9	0.64	0.59	0.67	0.73	0.52	0.48

Table 3: The average of von Misses stresses produced in cortical and the adjacent cancellous bone under *oblique load* in tapered and parallel implants with different abutment sizes (P1 = 5 mm abutment; P2 = 4.3 mm abutment; P3 = 3.5 mm abutment) at nine nodes (1-9) placed at equal distances from coronal to apical points (six in the cortical bone and three in the cancellous bone). The amount of stress was most pronounced in the most coronal part of the cortical bone (node 1) and was approximately three times greater with oblique loading compared to vertical loading (21.16 vs 7.16).

	Tapered implant			**Cylindrical implant**		
Bone layer nodes	**P1**	**P2**	**P3**	**P1**	**P2**	**P3**
The von Misses stress in cortical bone						
1	21.16	18.64	14.56	19.52	16.52	11.98
2	15.61	13.37	11.21	6.83	5.51	4.81
3	11.30	9.62	8.02	6.43	5.48	5.07
4	6.08	5.13	4.89	5.67	5.22	4.84
5	5.14	5.10	4.38	5.14	4.95	4.87
6	5.03	4.50	4.11	5.03	4.72	4.05
The von Misses stress in cancellous bone						
7	4.40	4.18	4.00	1.91	1.78	1.61
8	2.31	2.19	2.09	1.68	1.55	1.49
9	1.93	1.81	1.78	1.59	1.42	1.39

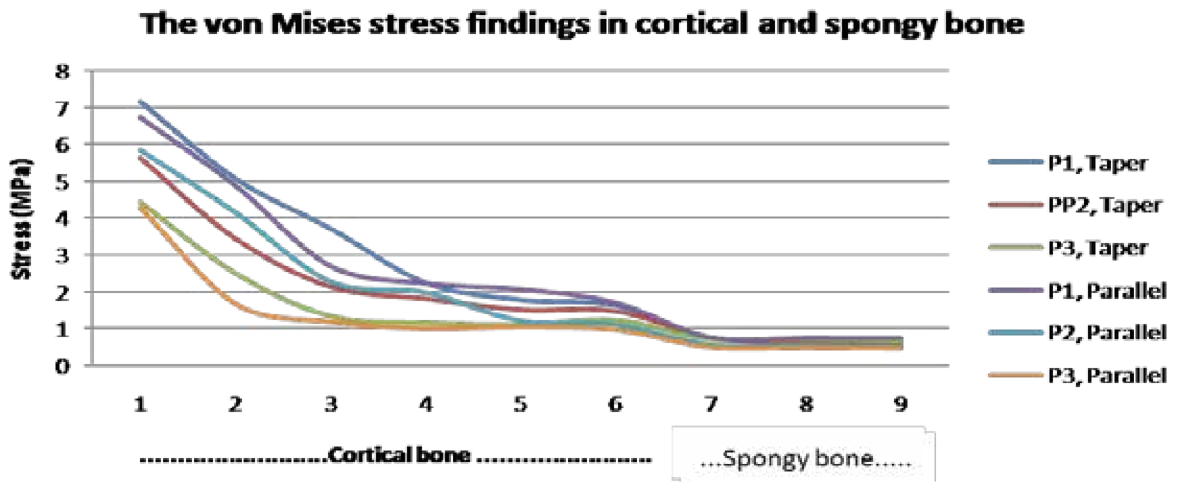

Figure 2. The von Misses Stress (MPa) findings in cortical and spongy bone under vertical loading. As the diagram demonstrates; the most pronounced stress was in crestal bone and around tapered implants.

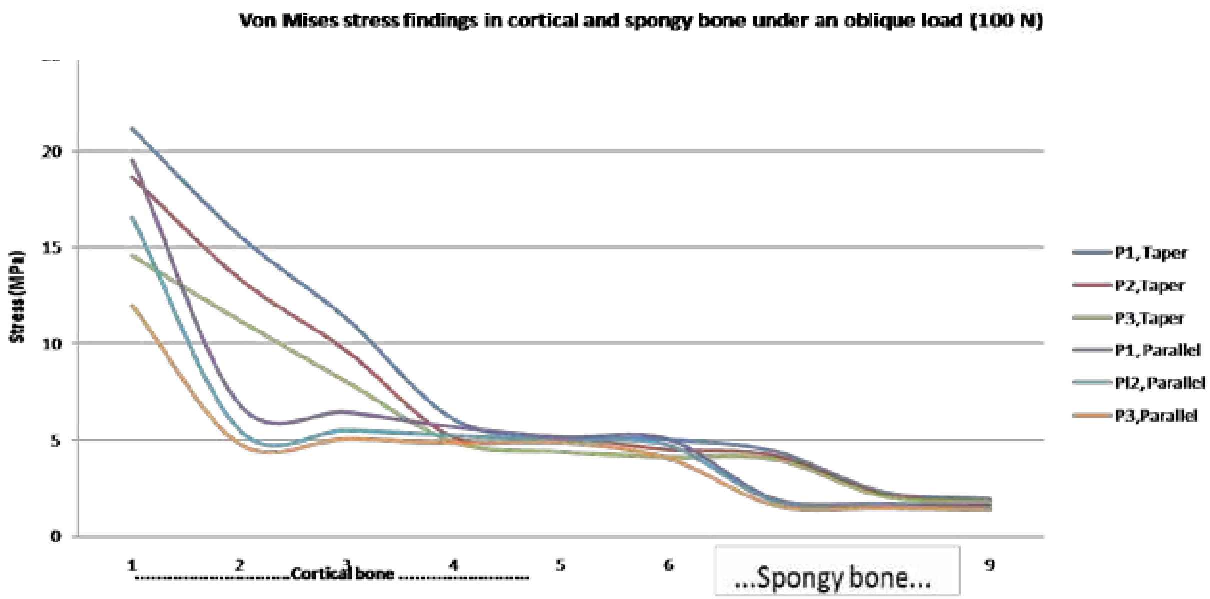

Figure 3. The von Misses Stress (MPa) findings in cortical and spongy bone under oblique loading. As the diagram demonstrates; the most pronounced stress was in crestal bone and around tapered implants.

With the smallest (3.5 mm) abutment (P3), stresses in the cortical bone ranged between 4.43 MPa in the crestal area and 1.24 MPa in the depth of the cortical bone. These values reached 0.67 MPa in the cancellous bone (*Figure 4B*).

Oblique loading

Crestal bone stresses that occurred under 15-degree oblique loading are shown in *Table 3*.

Cylindrical implant

Under oblique loading of the cylindrical implant with the 5 mm (P1) abutment, the von Misses stresses began at 19.52 MPa, followed a decreasing pattern, and reached 5.03 MPa in the deepest point of the cortical bone. This decrease continued in the cancellous bone to reach 1.59 MPa.

With the 4.3 mm (P2) abutment, stresses ranged from 16.52 MPa in the most crestal cortical bone to 4.72 MPa in the deepest cortical layer. The decreased stresses continued, reaching 1.43 MPa in the most apical layer of cancellous bone.

With the narrowest (3.5 mm) abutment (P3), the stresses observed in cortical bone ranged from 11.99 MPa in the crestal area to 4.06 MPa in the most apical area. These values reached 1.40 MPa in cancellous bone.

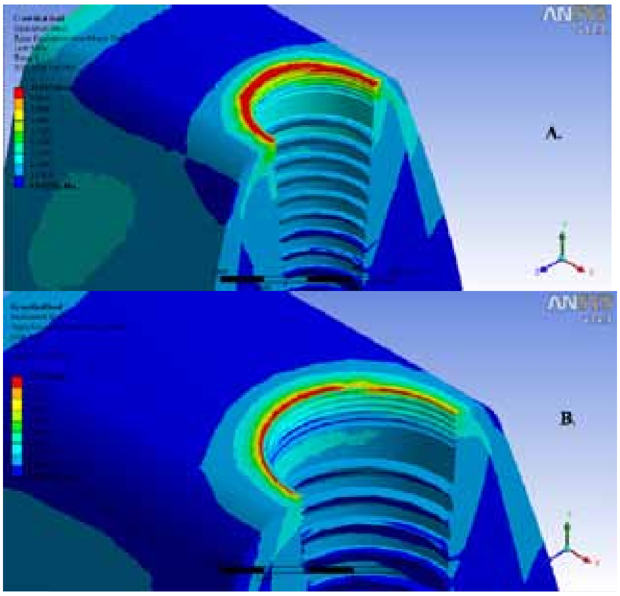

Figure 4. Distribution of von Misses stress around implants with different geometries. A, Model computed by FEA for tapered implant with length of 13 mm and diameter of 5 mm. B, Model computed for parallel implant with length of 13 mm and diameter of 5 mm. area of maximum stress in crestal bone is wider with implant with tapered wall. Red represents locality where maximum stress acts. Values of maximum stress in respective scales are higher for A as well.

Tapered implant

In the tapered implant with a 5 mm abutment (P1), the von Misses stresses began at 21.16 MPa, followed a decreasing pattern, and reached 5.04 MPa in the deepest point of cortical bone. This decrease continued in the cancellous bone to reach 1.93 MPa.

With the P2 (4.3 mm) abutment, stresses followed the same decreasing pattern: from 18.64 to 4.50 MPa in the cortical bone down to 1.8109 MPa in the deepest layer of cancellous bone.

With the smallest (3.5 mm) abutment (P3), stresses ranged from 14.56 MPa in the crestal area to 4.11 MPa in the apical portion of the cortical bone. These values reached 1.78 MPa in cancellous bone.

Discussion

The tapered implant has been a successful design, with many advantages, but the greater stresses in the crestal bone around it constitute a remarkable disadvantage (Cynthia *et al.*, 2005; Alves and Neves, 2009; Rokn *et al.*, 2011), making this implant design a challenge. On the other hand, some clinical reports have demonstrated minimal crestal bone resorption around implants that were restored by the platform-switching concept (Wegenberg and Froum, 2006; Lazzara and Porter, 2006; Maeda *et al.*, 2007; Rodríguez-Ciurana *et al.*, 2009). However, the concept of platform switching is not fully understood and is mainly influenced by manufacturers' recommendations.

The present study focused on the combined effect of a tapered implant body and platform switching on crestal bone stresses. The authors believe that this combination needs to be assessed to help clinicians use better implant-abutment designs in clinical practice and thus treat their patients appropriately.

The purpose of this study was to analyze the effect of platform switching and a tapered implant body on implant crestal bone stress via three-dimensional FEA; for this purpose, the authors designed the prosthetic structures (abutment and crown) and implant body as one solid object. The rationale behind the omission of design elements, such as the abutment connection system, was to consider the suprastructure and the implant as one rigid object so that nodes could be shared between the abutment and the implant body. Because this study did not intend to evaluate small displacements between the abutment and the implant, modeling the components as one piece decreased the number of calculations needed, thereby reducing the time needed to evaluate the stresses.

Although micromovement plays a vital role in implant stability, it was determined that micromovement would have been similar in the different models. Thus, the present model was kept simple to evaluate the effect of platform switching and body tapering without the complexity of additional parameters that were not being investigated in this study.

Regardless of implant design, the result showed that the mean amount of stress induced by oblique forces was much greater than that induced by vertical forces, as expected from the knowledge of off-axis loading. However, stresses decreased in all groups at the apical part of the implant body. For example, in the tapered implants with 5 mm abutments, the crestal bone stress was 2.62 MPa with vertical force application, whereas the stress increased to as much as 8.11 MPa upon oblique loading, demonstrating that oblique forces can increase the stresses on crestal bone by threefold or more compared to vertical forces.

When the crestal bone stresses in the different implant designs were compared under both vertical and oblique loads, greater crestal stress (around 36% for oblique force and 7% for vertical force) was seen in tapered designs versus the cylindrical implants (*Tables 2* and *3, Figures 2* and *3)*. These results are consistent with previous studies that showed that tapered implants produced more crestal bone stress than cylindrical implants (Cynthia *et al.*, 2005; Patra *et al.*, 1998). Rismanchian *et al.* (2010) observed increased peak tensile stresses in cortical bone around tapered implants in comparison to cylindrical implants. Mohammed Ibrahim *et al.* (2011)

found that a tapered implant exhibited higher stress levels in bone than a cylindrical implant, which seemed to distribute stresses more evenly.

However, some authors claim that tapered implants could decrease the stresses around both cortical and trabecular peri-implant bone (Rieger *et al.*, 1989; Huang *et al.*, 2005). They proposed that threads with increased depth in tapered implants could enlarge the bone-implant contact area, resulting in decreased crestal bone stresses around tapered endosseous implants. However, we believed that, because an increase in the bone-implant interface area could be detected with cylindrical implants (Rismanchian *et al.*, 2010) as well, this could not be an accurate reason for the decreased crestal bone stress around tapered implants.

Our results showed that, although tapered implants have many benefits in implant dentistry, as previously mentioned, they can result in greater crestal bone stress around implants. However, our data suggest that when the abutment diameter was reduced by 14% or 30%, less stress was transferred to the peri-implant bone around both tapered and cylindrical implants, regardless of the direction of force application (vertical or oblique). This finding supports the hypothesis that connecting an implant to a smaller-diameter abutment may decrease bone resorption by shifting the stress concentration away from the crestal bone interface and guiding the forces of occlusal loading along the axis of the implant (Maeda *et al.*, 2007). Furthermore, this result also confirms the serendipitous clinical finding of Lazzara and Porter (2006) in the late 1980s that led to the introduction of the platform-switching concept, in which a narrower prosthetic component is connected to a wider implant, resulting in reduced crestal bone resorption.

In this study, a greater internal shift of the prosthetic components resulted in less crestal bone stress. Becker *et al.* (2007) declared that, in the clinical situation, because inflammatory cells typically infiltrate the implant complex at the abutment-implant junction and form a 1.5 mm semispherical zone, when the outer edge of the implant-abutment junction is repositioned away from the external outer edge of the implant and adjacent bone, less crestal bone loss will be observed. As seen in *Table 2*, when a vertical force was applied, the crestal bone stresses in the tapered implant model with a 5 mm abutment were 2.62 MPa; this was reduced to 2.00 MPa with the 4.3 mm abutment, and a further reduction of the abutment size to 3.5 mm (P3) lowered the crestal bone stress to 1.54 MPa. When oblique forces were applied in the tapered and same size abutment group (P1), the mean crestal bone stress was the highest (8.11 MPa); this was reduced to 7.17 MPa in the P2 group and 6.12 in the P3 group. This is in agreement with the results of the recent randomized controlled study of Canullo *et al.* (2010), which also demonstrated an inversely proportional relationship between marginal bone loss and the extent of inward shifting of the prosthetic component. Some other studies have also shown an improved distribution of biomechanical stresses in the peri-implant bone tissue and preservation of interimplant bone height and soft tissue levels (Maeda *et al.*, 2007; Tabata *et al.*, 2011; Annibali *et al.*, 2012).

A study by Schrotenboer *et al.* (2008) showed that, although the use of microthreads may increase crestal stresses upon loading, a reduced abutment diameter could result in reduction of the stresses transmitted to the crestal bone and thus, perhaps, less bone loss. However, other studies demonstrated no significant difference in crestal bone loss between platform-switched and platform-matched implants. An animal study conducted by Becker *et al.* (2007) indicated that smaller abutments could reduce the marginal bone loss in the early days after abutment connection, but the matching and platform-switched implants showed the same amount of bone loss after 28 days. This finding is supported by a recent randomized clinical trial published by Enkling *et al.* (2011), who observed substantially similar crestal bone loss in platform-switched and platform-matched implants that were placed in the posterior mandible and followed for one year. The bone loss that occurred was thought to be related to the extent of microbial colonization rather than to platform switching.

The authors believe that clinically relevant conclusions can be drawn from this study if the stated assumptions are accepted. Although marginal bone remodeling can be attributed to biological width formation and other many factors, the most common causes of implant-related complications involve excessive stress (Misch, 2008), and according to the data obtained in this study, stress can be managed through the choice of appropriate implant designs and prosthetic components. This theoretical analysis suggests that an inward shifting of the prosthetic component could compensate for the more pronounced crestal bone stress seen around the tapered implants that are so popular today among clinicians because of their better stability and manipulation, especially in the esthetic zone and in compromised bone. The authors believe that this study may lead clinicians to choose an implant system more carefully with regard to geometry (parallel or tapered) and the subsequent implant-abutment connection configuration (matched versus mismatched). Because one of the main theories of the mechanism of bone resorption around dental implants is stress concentration caused by occlusal loading, it is wise for clinicians to manage this biomechanical aspect in planning implant therapy. This study demonstrated that, in the appropriate quantity of bone in the posterior region of the jaws, the amount of stress is more pronounced than in anterior areas; because it could decrease the bone-implant stress concentration, a parallel-wall implant with an abutment connection with the maximum possible platform switch may contribute to long-term implant success as a result of the minimized microdamage to the peri-implant bone. Cullinane and Einhorn (2002) stated that even loads below the ultimate stress tolerance can cause bone failure, in which the microdamage of the bone can no longer be repaired.

On the other hand, in the anterior zone, which features many bony concavities or depressions (Swasty *et al.*, 2009), the placement of parallel-wall implants is often not feasible because of insufficient bone or the proximity of vital structures. Therefore, clinicians should choose the tapered-design implant with a platform-switched connection in these areas. This selection will assist in better load distribution and minimal bone loss. Studies have demonstrated that crestal bone loss generally coincides with the level of the first thread of the implant and can jeopardize the treatment outcome, especially in esthetically sensitive cases in which facial soft tissue deficiencies make the crown appear longer than desired and gingival papillae support depends on the crestal bone underneath (Lai *et al.*, 2007; Reikie, 1995; Buser *et al.*, 2004; Belser *et al.*, 2004).

Limitations of the study

Finite element analysis is a descriptive and numeric method that assesses an individual situation in a specific condition that the researcher defines according to the scientific data to address issues that are questionable or are heavily debated among scientists (such as platform switching) that are not feasible to assess in a clinical setting or human population. Our study has assessed the stress distribution around dental implants with two different types of loads (vertical/axial and oblique). Thus, in contrast to other types of studies, the FEA analysis does not need statistical analysis. Rather, the significance of the study can be derived through comparison of the results with the large amount of data gathered from previous FEAs. Finite element analysis can help clinicians and implant manufacturers to find the most biologically favorable configurations in which dental implants can be constructed in an attempt to reduce the risks of clinical failure.

An in-depth understanding of stress profiles encountered by the implant — and more importantly, in the surrounding jawbone — can be gained through the use of FEA. This increase in the understanding of stress distributions and magnitudes within the implant and surrounding jawbone will aid the optimization of implant design and insertion technique. It is essential that the clinician have an understanding of the methodology, applications, and limitations of FEA in implant dentistry and become more confident in interpreting the results of FEA studies and extrapolating these results to the clinical situation.

The FEA is one of the most frequently used methods in stress analysis in both industry and science. It is used for analyzing hip joints, knee prostheses, and dental implants. The results of the FEA computation depend on many individual factors, including material properties, boundary conditions, interface definitions, and the overall approach to the model. It is apparent that the present model is only an approximation of the clinical situation. The application of a 3-dimensional (3D) model simulation with the non-symmetric loading by the masticatory force on a dental implant resulted in a closer approximation of "clinical reality."

Other limitations of FEA include the following:

1. Individuals vary in the amount and direction of forces exerted during masticatory function. Although a 15-degree angle and 100 N were chosen because they were shown to be comparable to the status *in vivo*, the actual forces and vectors can vary among individuals (Richter, 1998; Haraldson *et al.*, 1988; Geng *et al.*, 2001). The technique here was used to illustrate the possible differences between a tapered implant and its cylindrical counterpart, with or without platform switching (Richter, 1998). Any changes in force application (direction or amount) would, of course, change the outcome.
2. In some cases, elements are omitted from FEA to simplify the process and make it feasible. The rationale behind the omission of design elements, such as a Morse taper connection system, was because our model was designed to have node sharing between the abutment and implant body.
3. The use of anisotropic properties, rather than the isotropic properties used here for cortical and cancellous bone, may have had an effect on the results compared to actual bone structures (Patra *et al.*, 1998). Because the goal of this study was to investigate the effects on surrounding bone when only two design aspects (implant body shape and platform size) of an implant system were modified, it was more efficient and minimally complicated to use isotropic values instead of anisotropic values. Therefore, 3D FEA modeling satisfied the criteria of easily depicting stress differences without using unnecessarily complex geometries that were viewed to have an insignificant impact on this study.
4. A situation with 100% bone-to-implant contact was assumed in our model to create a modern model similar to photoelastic models. In contrast, most histometric studies done *in vivo* have found bone-to-implant contact of 30% to 70% (Pierrisnard *et al.*, 2003; Richter, 1998; Waskewicz *et al.*, 1994). Also, the biomechanical reaction of the jawbone differs for each patient and some micromovement occurs clinically, so the fixed interface between the implant and bone assumed in this and other FEAs does not accurately represent clinical reality.

5. The jawbone and implants are very complicated structures. It is difficult to establish an accurate and valid 3D finite element model using conventional modeling techniques. Two-dimensional (2D) representations of implants and jawbone structures were often assumed in previous studies, some of which also failed to recognize the difference between the cortical and trabecular bones. As such, the calculated results are often very different from the actual situation for 2D analysis; hence, they cannot be used to guide implant treatment (Canay *et al.*, 1996; Patra *et al.*, 1998; Lewinstein *et al.*, 1995). In addition, an assumption of homogeneous, linear, elastic material behavior for the jawbone is typical of FEAs, which is characterized by a single Young's modulus and Poisson's ratio (Lewinstein *et al.*, 1995; Mihalko *et al.*, 1992; Nishihara and Nakagiri, 1994). This, again, represents a simplification of the actual bone structure.
6. In most research reported to date, axially applied static loads were assumed, instead of the more realistic dynamic, cyclic loads directed at the occlusal angle encountered during mastication (Geng *et al.*, 2001).
7. With regard to implant surface roughness, the finite element method has not been employed widely to evaluate the effect of surface roughness of an implant on the stress profile produced within the surrounding jawbone. Ronold *et al.* (2003) experimentally analyzed the optimum value for titanium implant roughness in bone attachment using a tensile test. The results supported observations from earlier studies that suggested an optimal surface roughness for bone attachment to be in the range between 3.62 and 3.90 microns. The analysis also indicated that further attachment depended on mechanical interlocking between bone and implant.

There is still a lack of information regarding the mechanical properties of bone tissue with respect to the time-dependent process of structural rearrangement in response to permanent mechanobiologic stimuli. The clinical relevance of numerical methods in defining the biomechanics of dental implants and in emphasizing the necessity of an integrated clinical-mechanical approach must be confirmed through additional research. The development of finite element analysis studies with improved geometric models using dynamic loading, if possible, with different bone types, in animal experiments, and in longitudinal clinical trials are still necessary.

Conclusion and Recommendations

Although tapered implants might result in higher crestal bone stress, a reduced abutment diameter can result in lower stresses, with an inverse relationship to the extent of inward shifting of the abutment. There is still a lack of information regarding the remodeling that occurs in response to permanent mechanobiologic stimuli. The clinical relevance and reliability of numerical methods in defining the biomechanics of dental implants and in highlighting the necessity of an integrated clinical-mechanical approach must be confirmed by additional research.

A realistic jawbone model, with a wider range of characteristics to reflect differences between individual patients, must be constructed. Computed tomography images, which readily distinguish between cortical and trabecular bone, and computer image processing can be used together to construct a precise 3D geometric model using reverse engineering (George and Rasha, 2001; Marco *et al.*, 1998; Wei and Pallavi, 2002).

The unpredictable biomechanical response of the jawbone to a foreign object, i.e., stress shielding, has not been modeled previously using numeric methods. Ideally, computer software, together with CT images obtained during the healing process, would be developed to predict the degree of stress shielding in the peri-implant jawbone. This will improve our understanding of jawbone remodeling with different implant placement techniques, designs, and loading conditions. Alternatively, photoelastic stress analysis might be used to evaluate stress shielding experimentally (Waskiewicz *et al.*, 1994).

Numerous investigations have sought to determine the optimal geometry of the implant body, with mixed results (Pierrisnard *et al.*, 2003; Himmlova *et al.*, 2004). A new methodology, perhaps involving the use of the application programming interface function of commercial software, might be employed to determine the optimal combination of length, diameter, taper, and implant thread dimensions and configuration for each bone type in all three dimensions (Vena *et al.*, 2000).

When an implant is surgically placed into the jawbone, it is mechanically screwed into a drilled hole of a smaller diameter, resulting in high amounts of stress as a result of insertion torque and because the implant is cutting into the jawbone. As such, the stress condition in the jawbone will change accordingly. The long-term effects of such stresses remain unclear and should be investigated so that undesirable stresses can be minimized.

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References

Abrahamsson I, Berglundh T and Lindhe J. The mucosal barrier following abutment dis/reconnection. An experimental study in dogs. *Journal of Clinical Periodontology* 1997; **24**:568-572.

Abu-Hammad O, Khraisat A, Dar-Odeh N, Jagger DC and Hammerle CH. The staggered installation of dental implants and its effect on bone stresses. *Clinical Implant Dentistry and Related Research* 2007; **9**:121-127.

Albrektsson T and Isidor F. Consensus report of session IV. In: Lang NP and Karring T (Eds). *Proceedings of the 1st European Workshop on Periodontology*. London: Quintessence Publishing 1994; 365-369.

Albrektsson T, Zarb G, Worthington P and Eriksson AR. The long-term efficacy of currently used dental implants: A review and proposed criteria of success. *International Journal of Oral & Maxillofacial Implants* 1986; **1**:11-25.

Alves CC, Neves M. Tapered implants: from indications to advantages. *International Journal of Periodontics & Restorative Dentistry* 2009; **29**:161-167.

Annibali S, Bignozzi I, Cristalli MP, Graziani F, La Monaca G and Polimeni A. Peri-implant marginal bone level: a systematic review and meta-analysis of studies comparing platform switching versus conventionally restored implants. *Journal of Clinical Periodontology* 2012; **39**:1097-1113.

Ash M. *Wheeler's Dental Anatomy, Physiology and Occlusion*, 8th ed. Philadelphia: WB Saunders Company, 2005; 145-160.

Becker J, Ferrari D, Herten M, Kirsch A and Schwartz F. Influence of platform switching on crestal bone changes at non-submerged titanium implants: A histomorphometrical study in dogs. *Journal of Clinical Periodontology* 2007; **34**:1089-1096.

Becker W, Goldstein M, Becker BE and Sennerby L. Minimally invasive flapless implant surgery: A prospective multicenter study. *Clinical Implant Dentistry and Related Research* 2005; **7(Suppl. 1)**:S21-S27.

Belser UC, Schmid B, Higginbottom F and Buser D. Outcome analysis of implant restorations located in the anterior maxilla: A review of the recent literature. *International Journal of Oral & Maxillofacial Implants* 2004; **19 Suppl**:30-42.

Berglundh T and Lindhe J. Dimension of the periimplant mucosa. Biological width revisited. *Journal of Clinical Periodontology* 1996; **23**:971-973.

Bratu EA, Tandlich M and Shapira L. A rough surface implant neck with microthreads reduces the amount of marginal bone loss: A prospective clinical study. *Clinical Oral Implants Research* 2009; **20**:827-832.

Broggini N, McManus LM, Hermann JS, *et al.* Periimplant inflammation defined by the implant-abutment interface. *Journal of Dental Research* 2006; **85**:473-478.

Buser D, Martin W and Belser UC. Optimizing esthetics for implant restorations in the anterior maxilla: Anatomic and surgical considerations. *International Journal of Oral and Maxillofacial Implants 2004;* **19 Suppl**:43-61.

Buser D, Weber HP and Lang NP. Tissue integration of nonsubmerged implants. 1-year results of a prospective study with 100 ITI hollow-cylinder and hollow-screw implants. *Clinical Oral Implants Research* 1990; **1**:33-40.

Canay S, Hersek N, Akpinar I and Asik Z. Comparison of stress distribution around vertical and angled implants with finite element analysis. *Quintessence International* 1996; **27**:591-598.

Canullo L, Fedele GR, Iannello G and Jepsen S. Platform switching and marginal bone-level alterations: The results of a randomized-controlled trial. *Clinical Oral Implants Research* 2010; **21**:115-121.

Canullo L, Pace F, Coelho P, Sciubba E and Vozza I. The influence of platform switching on the biomechanical aspects of the implant-abutment system. A three dimensional finite element study. *Medicina Oral, Patología Oral y Cirugía Bucal* 2011; **16**:e852-856.

Choquet V, Hermans M, Adriaenssens P, Daelemans P, Tarnow DP and Malevez C. Clinical and radiographic evaluation of the papilla level adjacent to single-tooth dental implants. A retrospective study in the maxillary anterior region. *Journal of Periodontology* 2001; **72**:1364-1371.

Cochran DL. The scientific basis for and clinical experiences with Straumann implants including the ITI Dental Implant System: A consensus report. *Clinical Oral Implants Research* 2000; **11(Suppl 1)**:33-58.

Craig RG and Farah WJ. Stress from loading distal extension removable partial dentures. *Journal of Prosthetic Dentistry* 1978; **39**:274-277.

Cullinane DM and Einhorn TA. *Biomechanics of Bone*. San Diego, CA: Academic Press; 2002:17-32.

Cynthia S, Petrie CS and Williams JL. Comparative evaluation of implant designs: influence of diameter, length, and taper on strains in the alveolar crest. A three-dimensional finite-element analysis. *Clinical Oral Implants Research* 2005; **16:**486-494.

El Charkawi G and El Waked M. Effect of splinting on load distribution of extracoronal attachments with distal extension prosthesis, *in vitro*. *Journal of Prosthetic Dentistry* 1996; **76**:315-320.

Enkling N, Jöhren P, Klimberg V, Bayer S, Mericske-Stern R and Jepsen S. Effect of platform switching on peri-implant bone levels: a randomized clinical trial. *Clinical Oral Implants Research* 2011; **22**:1185-1192.

Ericsson I, Persson LG, Berglundh T, Marinello CP, Lindhe J and Klinge B. Different types of inflammatory reactions in peri-implant soft tissues. *Journal of Clinical Periodontology* 1995; **22**:255-261.

Fejerdy P, Borbely J, Schmidt J, Janh M and Hermann P. Removable partial denture design and its effect on remaining teeth, based on Hungarian national survey. *Fogorvosi Szemle* 2008; **101**:3-11.

Geng JP, Tan KB and Liu GR. Application of finite element analysis in implant dentistry: a review of the literature. *Journal of Prosthetic Dentistry* 2001; **85:**585-598.

George KK and Rasha AN. Adaptive reconstruction of bone geometry from serial cross-sections. *Artificial Intelligence in Engineering* 2001; **15:**227-239.

Geramy A. Alveolar bone resorption and the center of resistance modification: 3D analysis by means of the FEM. *American Journal of Orthodontics* 2000; **11**:399-405.

Hansson S. A conical implant-abutment interface at the level of the marginal bone improves the distribution of stresses in the supporting bone. An axisymmetric finite element analysis. *Clinical Oral Implants Research* 2003; **14**:286-293.

Haraldson T, Jemt T, Stalblad PA, and Lekholm U. Oral function in subjects with overdentures supported by osseointegrated implants. *Scandinavian Journal of Dental Research* 1988; **96:**235-242.

Hermann JS, Schoolfield JD, Schenk RK, Buser D and Cochran DL. Influence of the size of the microgap on crestal bone changes around titanium implants. A histometric evaluation of unloaded non-submerged implants in the canine mandible. *Journal of Periodontology* 2001; **72**:1372-1383.

Himmlova L, Dostalova T, Kacovsky A and Konvickova S. Influence of implant length and diameter on stress distribution: a finite element analysis. *Journal of Prosthetic Dentistry* 2004; **91:**20-25.

Huang HL, Lin CL, Ko CC, Chang CH, Hsu JT and Huang JS. Stress analysis of implant-supported partial prostheses in anisotropic mandibular bone: In-line versus offset placements of implants. *Journal of Oral Rehabilitation* 2006; **33**:501-508.

Huang L, Lynn Shotwell J and Wang H. Dental implants for orthodontic anchorage. *American Journal of Orthodontics and Dentofacial Orthopedics* 2005; **127**:713-722.

Kim Y, Oh TJ, Misch CE and Wang HL. Occlusal considerations in implant therapy: Clinical guidelines with biomechanical rationale. *Clinical Oral Implants Research* 2005; **16**:26-35.

King GN, Hermann JS, Schoolfield JD, Buser D and Cochran DL. Influence of the size of the microgap on crestal bone levels in non-submerged dental implants: A radiographic study in the canine mandible. *Journal of Periodontology* 2002; **73**:1111-1117.

Kitamura E, Stegaroiu R, Nomura S and Miyakawa O. Influence of marginal bone resorption on stress around an implant—a three-dimensional finite element analysis. *Journal of Oral Rehabilitation* 2005; **32**:279-286.

Kunavisarut C, Lang LA, Stoner BR, *et al.* Finite element analysis on dental implant-supported prosthesis without passive fit. *Journal of Prosthetic Dentistry* 2002; **11**:30-40.

Lai YL, Chou IC, Liaw YC, Chen HL, Lin YC and Lee SY. Triple immediate therapy (ridge expansion, soft tissue augmentation, and provisional restoration) of maxillary anterior single implant. *Journal of Periodontology* 2007; **78**:1348-1353.

Lazzara RJ and Porter SS. Platform switching: A new concept in implant dentistry for controlling postrestorative crestal bone levels. *International Journal of Periodontics & Restorative Dentistry* 2006; **26**:9-17.

Lewinstein I, Banks-Sills L and Eliasi R. A finite element analysis of a new system (IL) for supporting an implant-retained cantilever prosthesis. *International Journal of Oral and Maxillofacial Implants 1995*; **10**:355-366.

Maeda Y, Miura J, Taki I and Sogo M. Biomechanical analysis on platform switching: Is there any biomechanical rationale? *Clinical Oral Implants Research* 2007; **18**:581-584.

Marco V, Cinzia Z and Luisa P. TRI2SOLID: an application of reverse engineering methods to the creation of CAD models of bone segments. *Computer Methods and Programs in Biomedicine* 1998; **56**:211-220.

Mihalko WM, May TC, Kay JF and Krause WR. Finite element analysis of interface geometry effects on the crestal bone surrounding a dental implant. *Journal of Implant Dentistry* 1992; **1**:212-217.

Misch CE. *Contemporary Implant Dentistry*, 3rd ed. St Louis: Mosby Elsevier, 2008; 69, 202.

Misch CE. Density of bone: Effect on treatment plans, surgical approach, healing, and progressive bone loading. *International Journal of Oral Implantology* 1990; **6**:23-31.

Misch CE, Perel ML, Wang HL, *et al.* Implant success, survival, and failure: the International Congress of Oral Implantologists (ICOI) Pisa Consensus Conference. *Implant Dentistry* 2008; **17**:5-15.

Mohammed Ibrahim M, Thulasingam C, Nasser KS, Balaji V, Rajakumar M and Rupkumar P. Evaluation of design parameters of dental implant shape, diameter and length on stress distribution: a finite element analysis. *Journal of the Indian Prosthodontic Society* 2011; **11**:165-171.

Myshin HL and Wiens JP. Factors affecting soft tissue around dental implants: A review of the literature. *Journal of Prosthetic Dentistry* 2005; **94**:440-444.

Nishihara K and Nakagiri S. Biomechanical studies on newly tailored artificial dental root. *Biomedical Material Engineering* 1994; **4**:141-149.

O'Mahony AM, Williams JL and Spencer P. Anisotropic elasticity of cortical and cancellous bone in the posterior mandible increases peri-implant stress and strain under oblique loading. *Clinical Oral Implants Research* 2001; **12**:648-657.

Patra AK, DePaolo JM, D'Souza KS, DeTolla D and Meenaghan MA. Guidelines for analysis and redesign of dental implants. *Implant Dentistry* 1998; **7**:355-368.

Pierrisnard L, Renouard F, Renault P and Barquins M. Influence of implant length and bicortical anchorage on implant stress distribution. *Clinical Implant Dentistry and Related Research* 2003; **5**:254-262.

Reikie DF. Restoring gingival harmony around single tooth implants. *Journal of Prosthetic Dentistry* 1995; **74**:47-50.

Richter EJ. *In vivo* horizontal bending moments on implants. *International Journal of Oral and Maxillofacial Implants* 1998; **13**:232-244.

Rieger MR, Fareed K, Adams WK and Tanquist RA. Bone stress distribution for three endosseous implants. *Journal of Prosthetic Dentistry* 1989; **61**:223-228.

Rismanchian M, Birang R, Shahmoradi M, Talebi H and Jabar Zare R. Developing a new dental implant design and comparing its biomechanical features with four designs. *Dental Research Journal* 2010; **7**:70-75.

Rodríguez-Ciurana X, Vela-Nebot X, Segalà-Torres M, *et al.* The effect of interimplant distance on the height of the interimplant bone crest when using platform-switched implants. *International Journal of Periodontics & Restorative Dentistry* 2009; **29**:141-151.

Rokn A, Ghahroudi AR, Mesgarzadeh A, Miremadi A and Yaghoobi S. Evaluation of stability changes in tapered and parallel wall implants: a human clinical trial. *Journal of Dentistry (Tehran)* 2011; **8**:186-200.

Ronold HJ, Lyngstadaas SP and Ellingsen JE. Analysing the optimal value for titanium implant roughness in bone attachment using a tensile test. *Biomaterials* 2003; **24**:4559-4564.

Roos J, Sennerby L, Lekholm U, Jemt T, Gröndahl K and Albrektsson T. A qualitative and quantitative method for evaluating implant success: A 5-year retrospective analysis of the Brånemark implant. *International Journal of Oral & Maxillofacial Implants* 1997; **12**:504-514.

Roos-Jansåker AM, Lindahl C, Renvert H and Renvert S. Nine- to fourteen-year follow-up of implant treatment. Part II: presence of peri-implant lesions. *Journal of Clinical Periodontology* 2006; **33**:290-295.

Schrotenboer J, Tsao YP, Kinariwala V and Wang HL. Effect of microthreads and platform switching on crestal bone stress levels: a finite element analysis. *Journal of Periodontology* 2008; **79**:2166-2172.

Siegele D and Soltesz U. Numerical investigations of the influence of implant shape on stress distribution in the jaw bone. *International Journal of Oral & Maxillofacial Implants* 1989; **4**:333-340.

Smith DE and Zarb GA. Criteria for success of osseointegrated endosseous implants. *Journal of Prosthetic Dentistry* 1989; **62**:567-572.

Swasty D, Lee JS, Huang JC, *et al.* Anthropometric analysis of the human mandibular cortical bone as assessed by cone-beam computed tomography. *Journal of Oral and Maxillofacial Surgery* 2009; **67**:491-500.

Tabata LF, Assunção WG, Adelino Ricardo Barão V, de Sousa EA, Gomes EA and Delben JA. Implant platform switching: Biomechanical approach using two-dimensional finite element analysis. *Journal of Craniofacial Surgery* 2010; **21**:182-187.

Tabata LF, Rocha EP, Barão VA and Assunção WG. Platform switching: biomechanical evaluation using three-dimensional finite element analysis. *International Journal of Oral & Maxillofacial Implants* 2011; **26**:482-491.

Tarnow DP, Magner AW and Fletcher P. The effect of the distance from the contact point to the crest of bone on the presence or absence of the interproximal dental papilla. *Journal of Periodontology* 1992; **63**:995-996.

Vena P, Verdonschot N, Contro R, Huiskes R. Sensitivity analysis and optimal shape design for bone-prosthesis interfaces in a femoral head surface replacement. *Computer Methods Biomechanics and Biomedical Engineering* 2000; **3**:245-256.

Vigolo P and Givani A. Platform-switched restorations on wide-diameter implants: A 5-year clinical prospective study. *International Journal of Oral & Maxillofacial Implants* 2009; **24**:103-109.

Waskewicz VJ, Ostrowski GA and Parks JS. Photoelastic analysis of stress distribution transmitted from a fixed prosthesis attached to osseointegrated implants. *Journal of Oral and Maxillofacial Implants* 1994; **9**:405-411.

Wegenberg B and Froum SJ. A retrospective study of 1925 consecutively placed immediate implants from 1988 - 2004. *International Journal of Oral & Maxillofacial Implants* 2006; **21**:71-80.

Wei S and Pallavi L. Recent development on computer aided tissue engineering –a review. *Computer Methods and Programs in Biomedicine* 2002; **67**:85-103.

Weng D, Nagata MJ, Bell M, Bosco AF, de Melo LG and Richter EJ. Influence of microgap location and configuration on the periimplant bone morphology in submerged implants. An experimental study in dogs. *Clinical Oral Implants Research* 2008; **19**:1141-1147.

Yang H-S, Lang LA and Felton DA. FEA on the effect of the splinting in FPDs. *Journal of Prosthetic Dentistry* 1999; **81**:721-728.