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Original Article

Influence of Implant Thread's Cross-Sectional Design On The Micromotion Of The Implants In Different Bone Qualities: Finite Element Analysis

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Abstract

Aim: To evaluate the effect of the implant's thread cross sectional design on the micromotion of the implants in different bone qualities.
Materials and methods: In this finite element analysis study, four implant models exhibiting four different thread designs were created using Solidworks software. Each model was assembled in four different bone quality models resulting in 16 groups. Each group was examined for micromotion under 100N force using ANSYS software.
Results: D1 bone showed the lowest micromotion while D4 bone showed the highest, regardless the thread design. In D1 and D2 bone, v-shaped thread showed the lowest micromotion, which was lower than the square thread and the buttress was lower than the reverse buttress. While in D3 and D4 bone, the lowest micromotion was observed with the square thread, which was lower than the v-shaped and the reverse buttress thread was lower than the buttress thread.
Conclusion: Micromotion decreases as the bone quality increases irrelevant of the implant thread design. In D1 and D2 bone, the best thread to be used is the v-shaped thread, while in D3 and D4 bone, the best is the square shaped thread.

Keywords: Thread design, micromotion, bone qualities, finite element analysis, bone-implant contact.

Introduction

Various factors contribute to the success or failure of implants and should be considered for proper and suitable treatment plan. These factors include implant related factors as implant design and patient related factors whether local or systemic factors such as; age, medical condition, soft tissue biotype and quantity & quality of bone. **Carl E. Misch** has classified the bone quality according to density into D1, D2, D3 and D4. This classification differentiated between them according to the ratio between cortical and cancellous bone and the size of the trabecular spaces of the cancellous bone, which influences the bone-to-implant contact (BIC) and consequently affects the initial stability and micromotion of the implants. (Misch, 2008; Hsu *et al.*, 2013)

Defective bone quality seriously affect the prognosis of dental implants since it affects the implant stability, which is considered one of the most important criteria for implant success. Initial implant stability is part of the prerequisites for an implant osseointegration (secondary stability). One of the problems encountered during implant surgeries was insufficient initial stability, which could be enhanced by various factors including special surgical techniques, implant designs and surface treatments. (Alghamdi, 2018) (Alghamdi, 2018)

Implant designs include the implant diameter, implant taper and implant thread geometry, which have an important role in management of the biomechanical loads as they affect the distribution of peri-implant stresses that are transferred from the implant-supported prosthesis to the surrounding bone and biological tissues (English, 2005; Mosavar, Ziaei and Kadkhodaei, 2015)**.** Peri-implant stress distribution and osseointegration are important factors responsible for the dental implant long-term success.

Many variables affect the thread geometry as: thread pitch, thread depth and thread cross-sectional design (Mosavar, Ziaei and Kadkhodaei, 2015)**.** There are four cross-sectional thread designs described in the literature, which are; buttress, reverse buttress, v-shaped and square. The best thread design is the one that should provide the following requirements: optimal surface area, optimal initial bone-implant contact (BIC), optimal initial stability, and facilitates stresses dissipation (English, 2005; Mosavar, Ziaei and Kadkhodaei, 2015)**.** Evaluation of the implant thread design and application of biomechanical principles are crucial for further clinical success (Eraslan and İnan, 2010).

Therefore, implant thread design is considered to be an extremely important factor as it is increasing implant surface area thus raising bone-to-implant contact (BIC), dissipating stresses, increasing initial stability, which consequently enhances the implant osseointegration and success (Manikyamba and Mc, 2017)**.**

Finite element analysis (FEA) has been used for a long time to predict the biomechanical performance of various dental implant macro and micro-designs as well as the effect of various clinical factors on implant success (Geng, Tan and Liu, 2001). FEA is a technology that provides a numerical analysis at any location within the mathematical models, which are of irregular geometry and different material properties.(Desai H, 2014; Parkhe *et al.*, 2015). FEA has become an important analytical tool in the evaluation of implant systems in dentistry (Parkhe *et al.*, 2015) since it resembles real clinical situations and when both (FEA and clinical) results were compared, FEA outcomes were found to be corresponding to the clinical outcomes.(Reddy, Rajasekar and Abdemagyd, 2018)

So, does the implant thread cross-sectional design affect the micromotion and initial stability of the implants in different bone qualities?

Materials and Methods

In this finite element analysis study, 4 implant models, exhibiting 4 different thread designs, were created using Solidworks, and each model was assembled with a superstructure and examined for micromotion under 100N force in 4 different qualities of bone using ANSYS software.

It was done through two main stages:

A. Designing the models and assembling them.
B. Analysis of the models

A. Designing the models and assembling them

3D models of the components (Implant models, bone models, superstructure model and gingival model) were designed and assembled together. This was done simultaneously as designing the superstructure depends on the assembly of the implant model in the bone models as well as the gingival model, which depends on the superstructure assembly.

Four implant models with different thread cross-sectional designs (buttress, reverse buttress, v-shaped and square thread designs) were assembled in each bone type model (D1, D2, D3 and D4) resulting in 16 models (Table 1)

Implant and bone models

Referring to the design of ISI-II Neobiotic implant specifications with some modifications, a 3D implant model of length 10 mm, 4 mm platform diameter and 3º apical tapering was created. One implant model was drawn and then the cross sectional design of the implant thread was changed while maintaining the thread depth (0.36mm) and thread pitch (0.50mm) constant, resulting in four implant models, each with a different thread cross-sectional design (buttress, reverse buttress, v-shaped and square thread designs) as shown in figure 1.

Cortical and cancellous bone models were created individually using **Mimics Medical 21.0** & **3-matic Medical 13.0** by segmentation from the dicom file of a patient's CBCT scan. Both bone models were imported to **3-matic Medical 13.0**, followed by importing the implant model and aligning it within the bone models.

Superstructure and gingival models

A zirconia crown was created using **Exocad Dental DB 2.2 Valletta**, then a gingival model was created using **3-matic Medical 13.0** and **Exocad** and aligned with the implant and bone models.

Bone qualities

Different bone qualities were obtained by calculating the volume bone mineral density (vBMD) of cortical and cancellous bone for each bone quality corresponding to its density (HU units) (Misch, 2008; Gulsahi, 2011; Giambini *et al.*, 2016)**.** Then the cortical and cancellous volume percentage were calculated (Table 2). Cortical and cancellous bone were superimposed with the obtained variable percentages to mimic different bone qualities (D1, D2, D3, D4) as shown in figure 2.

B. Analysis of the model

The process of finite element analysis was carried out by mechanical static structural **ANSYS 18.2** software. All the models were presented as a function of area and a cement layer of 0.2 µm thickness was created. The maximum and minimum elements size used in the current FEA models were 0.5mm and 0.0221mm respectively.

For stress linear static analysis, elastic (Young's) modulus and Poisson's Ratio are two essential parameters that need to be defined. They are enough for defining the linear part of the stress strain curve of any isotropic material. All model components were materials considered to be isotropic, homogenous, and linearly elastic materials. All the contacting structures were assumed to possess 100% contact at the interface. The type of contact was defined as bonded contact between all components.

Conventional loading was done using a 100N load, which was applied vertically on the central fossae of the teeth and the implant divided so that 40N on second molar, 40N on implant in first molar site and 20N on the second premolar.

All movements at the base of the mandible were restrained during load application in all directions. Therefore, a boundary condition (zero displacement) was applied on the bottom nodes of the mandible in the directions (X, Y and Z) and the ANSYS software was used for processing.

Results

The results were obtained for maximum micromotion and maximum principal stresses were recorded (Table 3) and were blotted in flowcharts shown in figure 3.

Micromotion

In D1 bone, the results of this study revealed that model 1(c) reflects the lowest micromotion and model 1(b) reflects the

Table 1: 16 groups of four implant models with different thread design in four qualities of bone.

Thread design/ Bone quality	D1	D2	D3	D4
Buttress	Model 1(a)	Model 2(a)	Model 3(a)	Model 4(a)
Reverse Buttress	Model 1(b)	Model 2(b)	Model 3(b)	Model 4(b)
V-shaped	Model 1(c)	Model 2(c)	Model 3(c)	Model 4(c)
Square-shaped	Model 1(d)	Model 2(d)	Model 3(d)	Model 4(d)

Table 2: Cortical and cancellous bone percentages in different bone types.

	Cortical bone volume %	Cancellous bone volume %
D1	80	20
D2	70	30
D3	45	55
D4	20	40

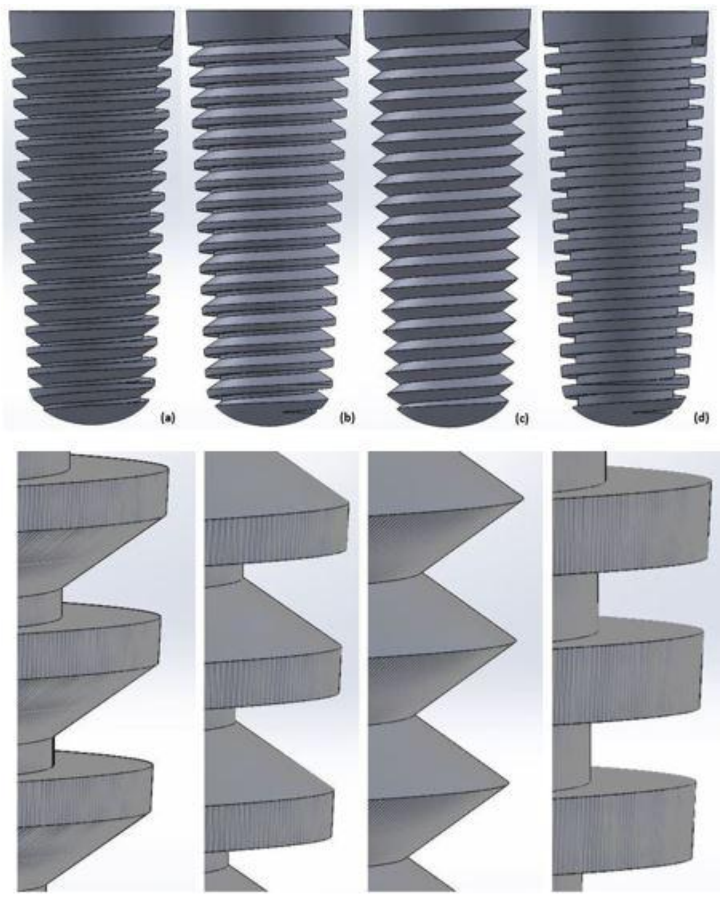

Figure 1: The 4 implant models with their corresponding cross-sectional thread design (a) buttress, (b) reverse buttress, (c) v-shaped and (d)square.

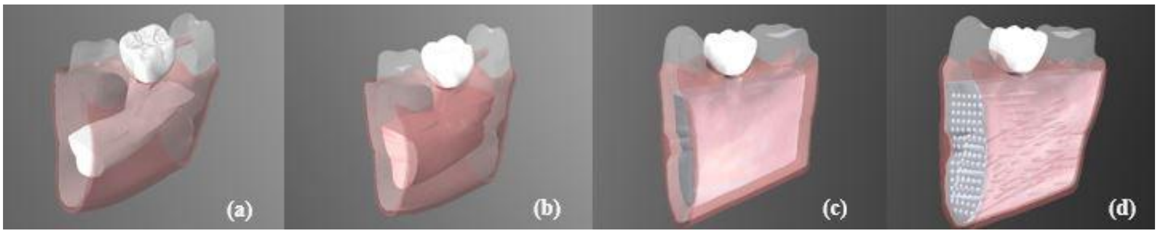

Figure 2: Different bone qualities with different % of cortical and cancellous bones (a) D1 bone (b) D2 bone (c) D3 bone (d) D4 bone.

Table 3: Maximum micromotion and maximum principal stress outcomes for all the models.

	Maximum micromotion (μm)	Maximum Principal Stress (MPa)
Model 1(a)	3.24	0.62
Model 1(b)	3.52	0.51
Model 1(c)	2.67	0.56
Model 1(d)	2.88	0.68
Model 2(a)	3.20	0.84
Model 2(b)	4.27	0.80
Model 2(c)	2.54	0.81
Model 2(d)	3.10	0.88
Model 3(a)	4.89	2.05
Model 3(b)	4.52	1.95
Model 3(c)	4.74	1.98
Model 3(d)	3.52	1.65
Model 4(a)	15.89	6.95
Model 4(b)	13.89	4.95
Model 4(c)	16.53	7.66
Model 4(d)	11.14	4.05

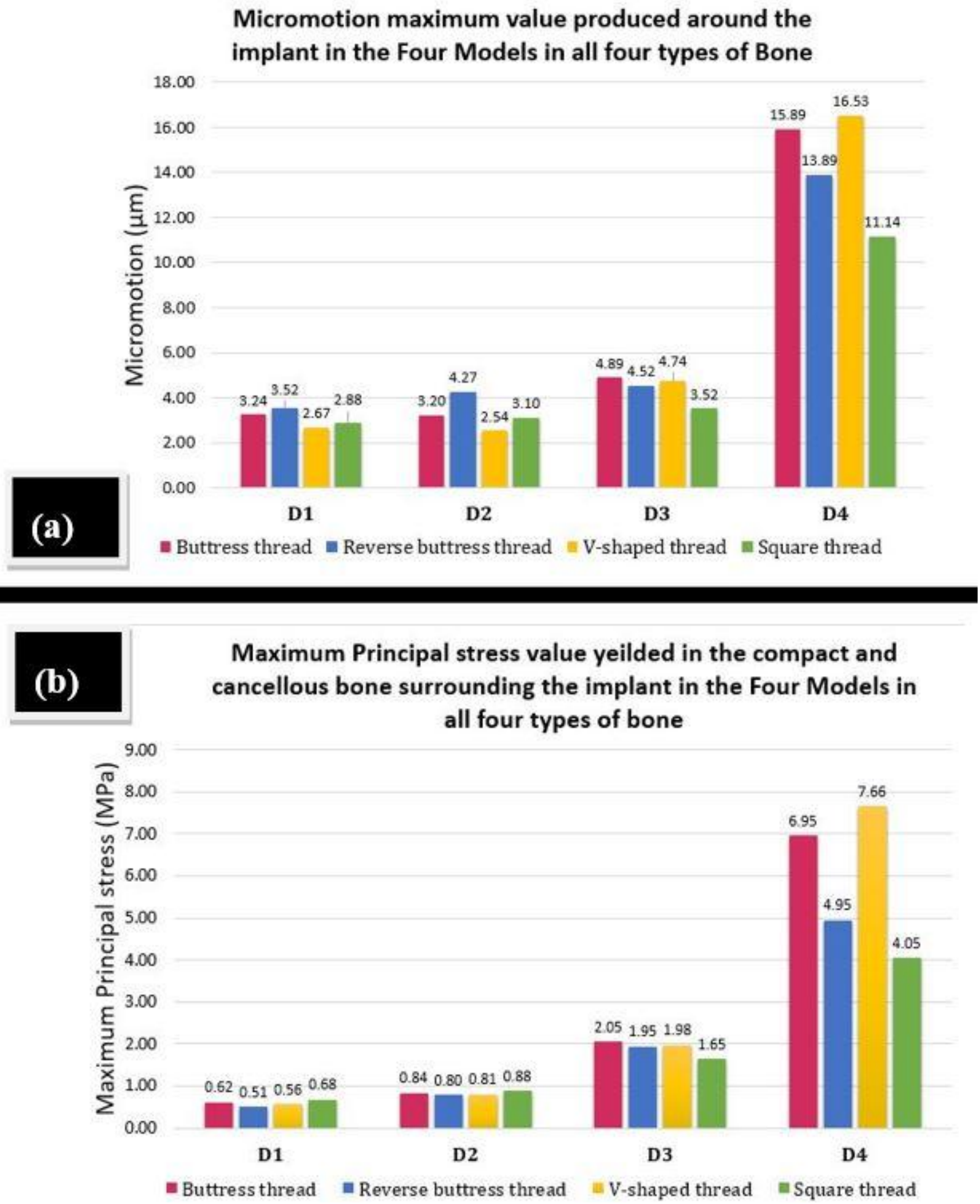

Figure 3: (a)A flow chart comparing between the maximum micromotion of the 16 groups, (b) A flow chart comparing between the maximum principal stresses of the 16 groups.

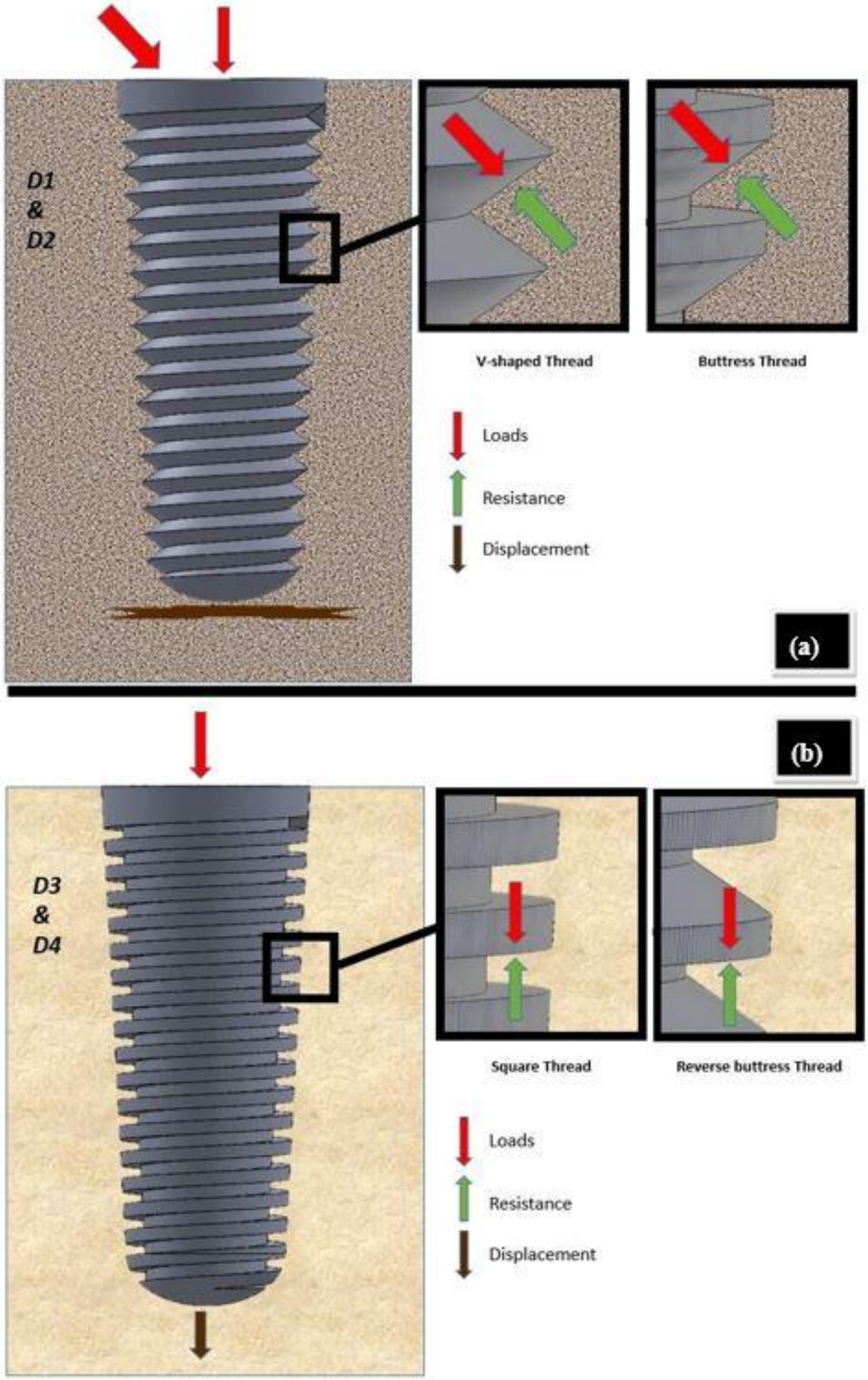

Figure 4: (a) Implant displacement pattern in D1 and D2 bone, (b) Implant displacement pattern in D3 and D4 bone

The other two models 1(a) and 1(d) revealed higher micromotion values than model 1(c) by 21.35% and 7.87% respectively.

In D2 bone, model 2(c) reflects the lowest micromotion and model 2(b) reflects the highest micromotion value among the models, which was 68.11% higher than model 2(c). The other two models 2(a) and 2(d) revealed higher micromotion values than model 1(c) by 25.98% and 22.05% respectively.

In D3 bone, model 3(d) reflects the lowest micromotion and model 3(a) reflects the highest micromotion among the models, which was 38.92% higher than model 3(d). The other two models 3(b) and 3(c) revealed higher micromotion values than model 3(d) by 28.41% and 34.66% respectively.

In D4 bone, model 4(d) reflects the lowest micromotion and model 4(c) reflects the highest micromotion among the models, which was 48.38% higher than model 4(d). The other two models 4(a) and 4(b) revealed higher micromotion values than model 4(d) by 42.64% and 24.69% respectively.

Stresses

In D1 bone, model 1(b) yielded the lowest stress and model 1(d) has the highest stress generated in the peri-implant bone which was higher than model 1(b) by 33.33%. The other two models 1(a) and 1(c) yielded higher stress values than model 1(b) by 21.57% and 9.80% respectively.

In D2 bone, model 2(b) yielded the lowest stress and model 2(d) has the highest stress generated in the peri-implant bone, which was higher than model 2(b) by 10%. The other two models 2(a) and 2(c) yielded higher stress values than model 2(b) by 5% and 1.25% respectively.

In D3 bone, model 3(d) yielded the lowest stress and model 3(a) has the highest stress generated in the peri-implant bone, which was higher than model 3(d) by 24.24%. The other two models 3(b) and 3(c) yielded higher stress values than model 3(d) by 18.18% and 20% respectively.

In D4 bone, model 4(d) yielded the lowest stress and model 4(c) has the highest stress generated in the peri-implant bone, which was higher than model 4(d) by 89.14%. The other two models 4(a) and 4(b) yielded higher stress values than model 4(d) by 71.60% and 22.22% respectively.

Discussion

Load application on implants induces three types of stresses in bone, which are compressive, tensile, and shear stresses. Maximum principal stresses was the target of this study as it measures the tensile stresses, which is the most harmful type, and it is used with brittle objects. Maximum equivalent stresses measures the sum of all the stresses induced and it is usually used with ductile objects (Trivedi, 2014)**.**

In the current study, the least micromotion values were observed in D1 bone and the highest micromotion values were observed in D4 bone. It was noticed that micromotion increases as the bone quality decreases for all thread designs. Micromotion values in D4 bone is higher by average value of 276.20 % than D1 bone (figure 3).

This is in agreement with (Wong *et al.*, 2005) who examined the effect of varying cortical and cancellous bone modulus on initial stem micromotion and interface bone strain using FEA. It was found that both the reduction of the modulus of cortical and cancellous bone caused an increase in the initial micromotion and interface bone strain.

It also coincides with (Ma *et al.*, 2014) who investigated the influence of thread pitch, helix angle, and compactness on micromotion in immediately loaded implants in bone of varying density (D2, D3, and D4) using five models of the three-dimensional finite element in three types of bone. Implant micromotion was assessed as the comprehensive relative displacement. They found that vertical relative displacement was affected by thread pitch, helix angle, and compactness. Under vertical loading, displacement was correlated positively with thread pitch and helix angle but negatively with compactness. Under horizontal loading in D2, the influence of pitch, helix angle, and compactness on implant stability was limited; however, in D3 and D4, the influence of pitch, helix angle, and compactness on implant stability is increased. The additional evidence was provided that trabecular bone density has less effect on implant micromotion than cortical bone thickness. Bone type amplifies the influence of thread pattern on displacement.

(Sugiura *et al.*, 2016) investigated the effects of bone density and crestal cortical bone thickness at the implant-placement site on micromotion (relative displacement between the implant and bone) and the peri-implant bone strain distribution under immediate-loading conditions. Delayed and immediate loading protocols were simulated as well as various bone parameters, including low or high cancellous bone density, low or high crestal cortical bone density, and crestal cortical bone thicknesses ranging from 0.5 to 2.5 mm. A buccolingual oblique load of 200N was applied to the top of the abutment. It was concluded that cancellous bone density might be a critical factor for avoiding excessive micromotion in immediately loaded implants. Crestal cortical bone thickness significantly affected the maximum extent of micromotion and peri-implant bone strain in simulations of low-density cancellous bone under immediate loading.

The reaction of both implants and peri-implant bone to loads is greatly dependent on the bone quality. Different bone qualities can yield different deformation patterns, which can alter the pattern of implant micromotion and the pattern of stress transfer.

Upon applying occlusal loads on anatomical teeth forms two types of stresses are generated namely vertical and lateral displacing forms. Vertical stresses have the tendency to displace the implant vertically while lateral stress tend to tip or rotate the implant around a central point located usually at the center of the implant body.

In high bone quality as D1 and D2 types, the bone possess relatively higher elastic modulus of elasticity and stiffness when compared to D3 and D4. This result in two different behavior patterns. In good quality bone, stress tend to concentrate coronally at the crestal portion with low tendency of apical stress propagation. On the contrary, in poor bone qualities the implant shows greater micromotion with tendency for stress propagation up to the implant apex. Such difference is well demonstrated in the literature.

This is consistent with the work of (Holmes and Loftus, 1997) who examined the influence of bone quality on the transmission of occlusal forces for endosseous dental implants using finite element analysis. The study modeled a 3.75 x 10-mm threaded implant placed in a 12 x 11 x 8-mm section of bone and by varying the elastic parameters assigned to the bone elements; four bone quality categories were established. A 100N load was applied at the occlusal surface of the restoration at a 30º angle to the vertical axis of the implant. Maximum von Mises stress concentrations were observed at the coronal aspect of the implant fixture in all four cases. Therefore, it was concluded that the placement of implants in bone with greater thickness of the cortical shell and greater density of the core will result in less micromovement and reduced stress concentration.

This also coincides with (Sevimay *et al.*, 2005) who aimed to investigate the effect of 4 different bone qualities on stress distribution in an implant-supported mandibular crown, using finite element analysis. A 3-D FE model of a mandibular section of bone with a missing second premolar tooth was developed, and solid 4.1 × 10-mm screw-type dental implant system and a metal-ceramic crown were modelled. A 300N vertical load was applied to the buccal cusp and the distal fossa of the crown followed by analysis. It was concluded that for the bone qualities investigated, stress concentrations in compact bone followed the same distributions as in the D3 bone model, but because the trabecular bone was weaker and less resistant to deformation than the other bone qualities modeled, the stress magnitudes were greatest for D3 and D4 bone.

(Kitagawa *et al.*, 2005) investigated how the thickness and Young's modulus of cortical bone influenced stress distribution in bone surrounding a dental implant using FEA. The results showed that von Mises equivalent stress was at its maximum in the cortical bone surrounding dental implant. It was found that maximum von Mises equivalent stress in bone decreased as cortical bone thickness increased. On the other hand, maximum von Mises equivalent stress in bone increased as Young's modulus of cortical bone increased. Therefore, it was concluded that von Mises equivalent stress was sensitive to the thickness and Young's modulus of cortical bone.

For D1 and D2 the bone generally exhibits lower deformation when external load is applied. In such cases, the apical displacement of the implant is relatively hindered by the underlying bone. The implant will show a greater tipping tendency at the cervical region rather than being apically displaced. The cervical implant tipping will tend to concentrate the stress at the crestal bone part with little propagation to the apical regions. In other words, in good quality bone occlusal forces will generate rotational bending in the implant body rather than apical displacement. The direction of such rotational bending is best counteracted by an inclined thread surface perpendicular to the rotation direction. This inclined surface is available at the bottom surface of v-shaped threads and the inclined surface of the buttress thread. The inclined surface of the reverse buttress is not perpendicular to such bending and square threads do not have such an incline. For such reason, it seems that the lowest implant displacement in D1 and D2 bone was observed with the v-shaped thread, which was lower than the square thread, and the buttress thread was lower than the reverse buttress thread (figure 4).

On the contrary, D3 and D4 bone exhibit lower stiffness values allowing for easier apical displacement under vertical loads. Vertical displacement is best counteracted by flat surface perpendicular to the resultant vertical displacement. Such a flat surface is available for both square shaped threads and the bottom surface of the reverse buttress thread. For this reason, the results were reversed, where the lowest displacement values in D3 and D4 bone qualities was observed with the square thread which was lower than the v-shaped thread and the reverse buttress thread was lower than the buttress thread (figure 4).

Conclusion

Micromotion decreases as the bone quality increases irrelevant of the implant thread design. In D1 and D2 bone, the best thread to be used is the v-shaped thread, while in D3 and D4 bone, the best is the square shaped thread.

Conflict of Interest and Source of Funding

No Conflict of interest

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