RESEARCH

Influence of Fine Threads and Platform-Switching on Crestal Bone Stress Around Implant—A Three-Dimensional Finite Element Analysis

Pardeep Khurana, MDS[1]*
Arun Sharma, MS[2]
Kiranmeet Kaur Sodhi, MDS[3]

The aims of this study were to investigate the effect of implant fine threads on crestal bone stress compared to a standard smooth implant collar and to analyze how different abutment diameters influenced the crestal bone stress level. Three-dimensional finite element imaging was used to create a cross-sectional model in SolidWorks 2007 software of an implant (5-mm platform and 10 mm in length) placed in the premolar region of the mandible. The implant model was created to resemble a commercially available fine thread implant. Abutments of different diameters (5.0 mm: standard, 4.5 mm, 4.0 mm, and 3.5 mm) were loaded with a force of 100 N at 90° vertical and 40° oblique angles. Finite element analysis was done in COSMOSWorks software, which was used to analyze the stress patterns in bone, especially in the crestal region. Upon loading, the fine thread implant model had greater stress at the crestal bone adjacent to the implant than the smooth neck implant in both vertical and oblique loading. When the abutment diameter decreased progressively from 5.0 mm to 4.5 mm to 4 mm and to 3.5 mm the thread model showed a reduction of stress at the crestal bone level from 23.2 MPa to 15.02 MPa for fine thread and from 22.7 to 13.5 MPa for smooth collar implant group after vertical loading and from 43.7 MPa to 33.1 MPa in fine thread model and from 36.9 to 20.5 MPa in smooth collar implant model after oblique loading. Fine threads increase crestal stress upon loading.)Reduced abutment diameter that is platform switching resulted in less stress translated to the crestal bone in the fine thread and smooth neck.

Key Words: implant fine thread, crestal bone stress, platform switching, finite element analysis

INTRODUCTION

The success of dental implants is highly dependent upon the integration between the implant and the intra-oral hard/soft tissue. The initial breakdown of the implant-tissue interface generally begins at the crestal region in successfully osseointegrated endosteal implants regardless of surgical approaches, with the potential to cause implant failure.

The implant body has macroscopic design-like threads whereas the crest module is often smoother to impair plaque retention if crestal bone loss occurs. Clinical success and longevity of endosteal dental implants are controlled in large by the health of the surrounding crestal region of bone and soft tissue. Early implant failure is also the consequence of too much stress applied to the implant system.[1] The first report quantifying early crestal bone loss was a 15-year retrospective study by Adell[2] who reported 1.2-mm marginal bone loss from the first thread during healing and in the first year after loading with average of 0.005 mm to 0.13 mm bone

[1] Department of Conservative Dentistry & Endodontics, Swami Devi Dyal Dental College & Hospital, Panchkula, India.
[2] Department of Mechanical Engineering, Michigan Technological University, Houghton, Mich.
[3] Department of Prosthodontics, Swami Devi Dyal Dental College & Hospital, Panchkula, India.
* *Corresponding author,* e-mail: drpkhurana@yahoo.com
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loss annually thereafter. The initial transosteal bone loss around the implant forms a V- or U-shaped pattern, which has been described as ditching or saucerization around the implant.[3] The stress forms a V-shaped pattern similar to the crestal bone loss. Earlier studies of three-dimensional finite element analysis of an implant within bone with axial load show that the values are greatest at the crest and gradually decrease in intensity as the stress is dissipated throughout the implant length. The apical end of implant receives no appreciable stress.[4]

The criteria for implant success as given by Smith and Zarb[5] states that vertical bone loss should be $<$ 0.2 mm annually following the first year of implant function. A postrestorative remodeled crestal bone generally coincides with the level of the first thread on most standard diameter implants. The first thread changes the shear force of the crest module to a component of compressive force to which the bone is most resistant. There are many suggested causes for early implant bone loss, two of them being occlusal overload and implant crest module.[6] The crest module of the implant body refers to the transosteal region of the implant that receives the stress from the implant after loading.[7] It has been hypothesized that the bone loss may slow down at the first thread when the force changes from crest as shear force to the compressive force induced by the thread itself.[3] In general, a functional implant may encounter many different forces such as rotation, shear, and compression; it was found that the cortical bone layer withstands compressive force the best.[8]

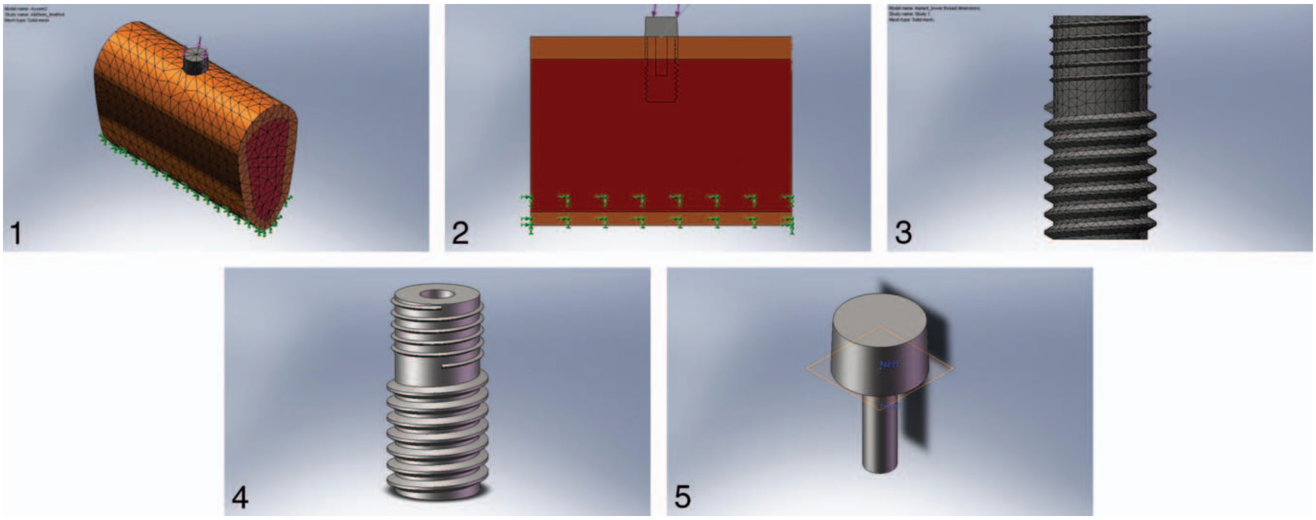
Therefore an implant system should be designed so that it can distribute stress to bone in a manner that supports a restoration in function and encourage osseous attachment. The functionality and longevity of these implant systems depend on the mechanical integrity of the prosthesis and the implant[9] as well as the ability of peri-implant structures to withstand and positively adapt to the applied forces.[10]

One design concept is rough external surface on the transosteal surface of an implant fixture. The mechanical benefit is an increase in an implant's bone contact available for stress translation. One type of roughened surface results from the addition of threads to the neck. A second design in the implant system is a switched platform abutment on an implant fixture, which has added clinical benefits of optimal management of prosthetic space and improved bone support for short implants. The aim of the study was to investigate the effect of implant threads on crestal bone stress compared to a standard smooth implant collar and to analyze how different abutment diameters influence crestal bone stress. To investigate the inference of the crestal module a computer model was developed to apply finite element analysis.[11] As the geometries involved with modeling implants and alveolar process are very complex, finite element analysis is considered the most suitable tool for analyzing them.

Materials and Methods

The three-dimensional finite element model created in SolidWorks 2007 (Dessault Systems, Waltham, Mass) of a completely osseo-integrated endosseous titanium implant in the posterior mandible also created in Solidworks 2007 was made for the purpose of stress analysis. The FEA software used was COSMOSWorks software (Dessault Systems), which helped generate the model, create the mesh of individual elements, and perform the analysis of the resulting model. An isotropic model is used, having identical physical properties in all directions. This model consisted of thick cortical bone surrounding dense trabecular bone, which is classified as type 2 bone.[12] (Figures 1 and 2)

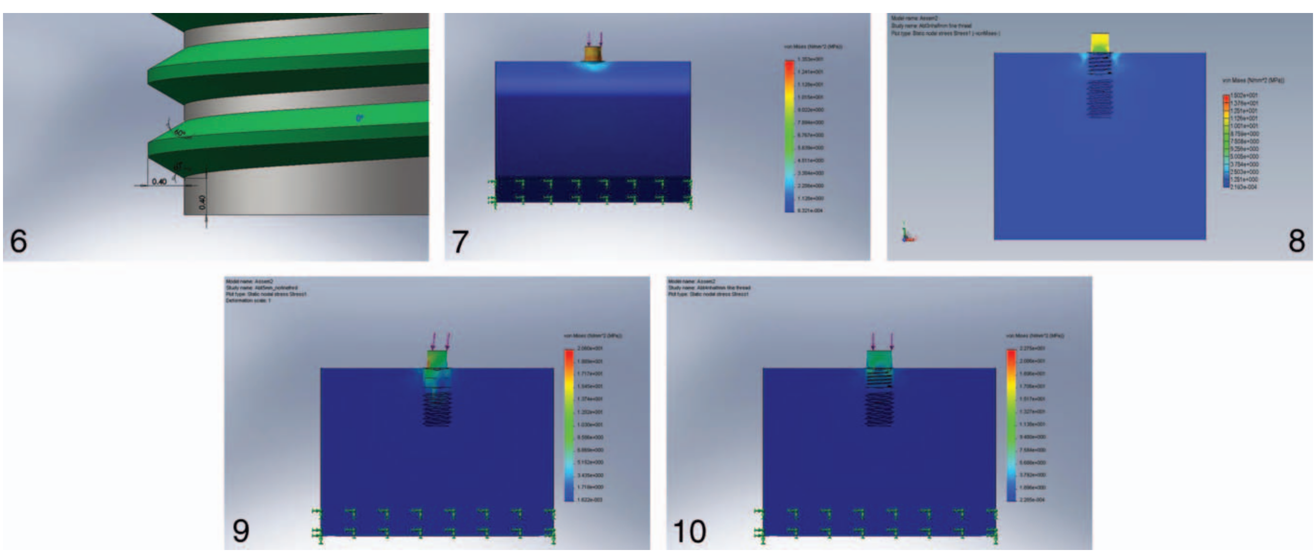
The resulting mandibular cross section was 28 mm in vertical dimension and 14.1 mm at the greatest horizontal dimension. Thickness of the cortical bone ranged from 0.595 mm to 1.515 mm with the crestal region measuring 1.5 mm, which is similar to another study.[13,14] The smooth neck implant model consisted of a restorative platform width of 5 mm, and a length of 10 mm. (Figures 3 and 4) The implant fine thread model was replicated with the exception of threads replacing the smooth neck portion. The fine thread in the cervical area was 0.23 mm in length and had a "V" shape, and the body implant thread had a 0.40 mm depth and 60° angle. The V-shaped thread tips are blunted thus rendering a square profile (Figure 6). The abutment models were 5 mm, 4.5 mm, 4 mm, and 3.5 mm in diameter (Figure 5). Complete osseo-integration at the implant bone interface was simulated by combining the nodes of the implant and bone



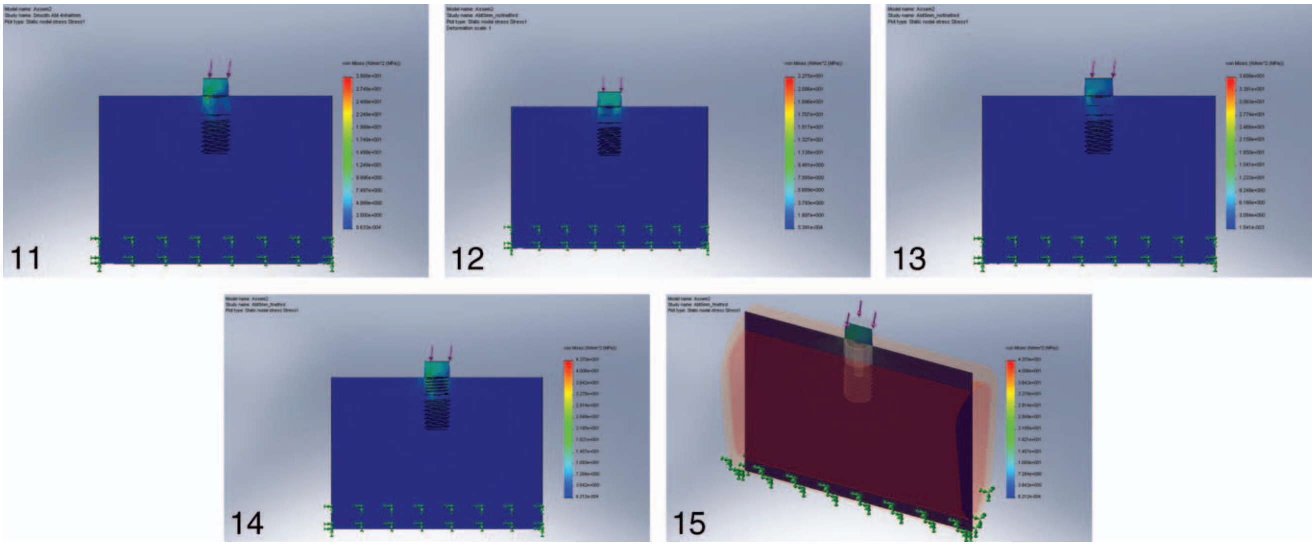
Figures 1–5. Figure 1. Finite element mesh assembly showing cortical and cancellous bone dimension with implant. **Figure 2.** Cross-section assembly view showing cortical (color orange) and cancellous bone (color red) along with implant (color gray). **Figure 3.** Implant mesh model. **Figure 4.** Implant model. **Figure 5.** Abutment, 3 mm in length (color gray).

model. The number of nodes used was 57 727 and number of elements used was 61 298 (Figure 1).

Similar integration of the abutment and implant body was adopted to be a single unit. This eliminated any potential influence from micro-movements between components. Model analysis consisted of two groups: smooth neck (control) and fine thread (test) each with four abutment diameters (5 mm, 4.5 mm, 4.0 mm, and 3.5 mm) (Figure 5). The 5-mm abutment represented the control diameter; 4.5 mm, 4 mm, and 3.5 mm represented a diameter reduction of 10%, 20%, and 30% respectively. This reduction in abutment diameter represented the concept of platform switching. Force applications were performed in oblique and vertical conditions using 100N as a representative masticatory force,[15] for oblique loading a 100 N force were applied at 40° from the vertical axis. This translates into 28.4N in horizontal direction and 94.2N in the vertical direction. The transfer of load was simulated to be from the central apical surface of the abutment, through the implant body to the



Figures 6–10. Figure 6. Implant lower body thread dimensions. **Figure 7.** Von Mises Stress Plot 3.5-mm smooth implant surface. **Figure 8.** Von Mises Stress Plot 3.5-mm fine thread. **Figure 9.** Stress plot 3.5-mm smooth implant surface with 40° oblique load. **Figure 10.** Stress plot 4.5-mm fine thread implant surface.



FIGURES 11–15. FIGURE 11. Stress plot 4.5-mm with 40° oblique load. **FIGURE 12.** Stress plot 5-mm smooth implant surface. **FIGURE 13.** Stress plot 5-mm smooth implant collar oblique load. **FIGURE 14**. Stress plot 5-mm fine thread oblique load. **FIGURE 15.** Stress plot 5-mm fine thread implant surface with 40° oblique load (Translucent).

perimucosal tissue. Stress analysis of all models consisted of mapping von Mises stress patterns upon the application of vertical and oblique loading. Von Mises stress used in FEA is measured in MPa. The values used in comparison were located at the most crestal cortical bone adjacent to implant fixture.

Results

In all models, stress concentration is greatest at the crestal level adjacent to the implant and at the top part of fine threads and body threads. The localized crestal adjacent to the implant model decreases towards the body threads. Upon oblique loading, the fine thread model with control abutment diameter has 46.96% greater stress at the crestal bone adjacent to the implant than the smooth neck implant (Figures 13 through 15). Under vertical load, the fine thread model displayed a 38.49% increase in crestal bone stress. Specifically, von Mises stresses in the fine thread model for control model were 43.7 MPa for oblique loading and 23.2 MPa for axial loading; in the smooth neck model they were 36.99 MPa and 22.75 MPa (Figure 12), respectively. In the fine thread model, when abutment diameter decreased from 5 mm to 4.5 mm and then to 4.0 mm and 3.5 mm, the reduction in crestal bone stress levels shown was 1.9%, 32.3%. and 2.34% on axial loading (Figures 8 and 10) and from 17.5%, 13.96%, and 7.02% after oblique loading, respectively. The stress level in the smooth neck model decreased from 18.24%, 12.74%, and 19.95% on vertical loading (Figure 7) and from 18.92%, 14.43%, and 20.01% after oblique loading (Figures 9 and 11).

Discussion

The purpose of the study was to use a three-dimensional computer model to evaluate the effect of fine threads and platform switching on implant crestal bone stress where the greatest stress was noted. FEA is broken down into many small elements. These small elements are joined to build a whole structure. Finite element analysis makes it possible to evaluate a detailed and complex structure in a computer based model. This study was primarily aimed at investigating linear analysis (elastic properties) compared to the nonlinear property difference between implant and bone present in vivo, hence the exact replica of in vivo stress was not given material importance. The simplicity of modeled abutment, implant, and mandibular cross-section was sufficient to demonstrate the effect of fine-thread and platform switching without adding complicated design parameters, because of mating condition between the abutment and implant body, there is little significance to the shape of abutment that inserts into implant body.

Table 1

Comparison of material properties (Schrotenboer et al[4])

Components	Modulus of rigidity (MPa)	Poisson Ratio
Titanium	117×10^3	0.30
Trabecular bone	1370	0.31
Cortical bone	2727	0.30

The study assumed isotropic properties for cortical and trabecular bone. The trabecular model used in this study was modeled as a solid, isotropic material with no porosities that are generally found in vivo trabecular bone. Using the isotropic properties instead of anisotropic bone properties for trabecular and cortical bone may have an effect on the result.[16] This study also modeled bone-implant contact as a consistent 100% to create a model similar to photo elastic models, whereas most common bone-implant contact percentage ranged from 30% to 70%.[16] The study model also demonstrated varying amounts of stress in all planes. The magnitude of bite force varies as a function of anatomical region and state of the dentition. For measurement of the analysis portion of this study, it was determined that the oblique loading model should be tested at 40° angle and a loading force of 100 N was chosen due to its acceptance in previous studies.[10] It is more comparable to in vivo mastication and a biologically feasible action that can be performed on an implant in vivo.[15,17] The application of horizontal vector creates the most shear stress in cortical bone and was shown to be the component of force best avoided for implant success.[11] Another group in which 100 N force was applied in vertical direction on computerized models was created. Although these forces and angles represent possible applications of force to the dental implant, the actual vector of force can vary among individuals.[18] Because of the constraint on the mandibular portion of the model, only the inferior one-third of the mandible was surrounded by a fixed constraint to minimize any interference during analysis.

Analysis of this study showed that the fine thread implant had 46.9% and 38.4% greater stress at the crestal bone adjacent to the implant than the smooth-neck implant in oblique and vertical load. These were similar to another study.[13] The range of von Mises stress in this study (13.53–44.2 MPa) is slightly higher than an in vivo condition and can be attributed to linear FEA modeling used in this study. Bone is an organ that responds to a number of factors including local mechanical forces. Cortical and trabecular bone are modified by modeling and remodeling.[19] Histological examination by Frost [20] showed 4 micro stain zones for compact bone and related them to the mechanical adaptation of stain. Other FEA studies showed a stress equivalent to 1.6 MPa was sufficient to avoid crestal bone loss from disuse atrophy in the canine mandibular premolar region on the other end. Sixty MPa was regarded as a threshold stain value above which bone failed to heal after fatigue.[21,22] Lee et al reports that the bone around the implant may be remodeled at a rate of 500% each year compared with normal trabecular physiologic remodeling of 20% per year.

Table 2

Result of finite element analysis of crestal bones stress level and platform switching showing von Mises Stresses at implant and cortical interface on vertical loading

	Axial Load = 100 N							
	Implant Without Microthreads				Implant With Microthreads			
	von Mises Stress (MPa)				von Mises Stress (MPa)			
Abutment Diameter (mm)	Min	Max	Average Stress (MPa)	Overall Max Stress (MPa)	Min	Max	Average Stress (MPa)	Overall Max Stress (MPa)
5	0.46	6.07	3.64	22.75	0.32	7.6	3.6	23.2
4.5	0.22	6.9	4	18.6	0.45	8	3.8	22.75
4	0.44	7.74	3.67	16.2	0.17	7.8	3.5	15.4
3.5	0.8	7.6	3.75	13.4	0.41	7.7	3.6	15

TABLE 3

Results from finite element analysis for oblique and vertical loading conditions

Crestal Stress Adjacent to Implant Neck (MPa)				
Abutment Diameter (mm)	5	4.5	4	3.5
Vertical Loading Crestal Stress (MPa)				
Fine Threads	23.2	21.75	15.38	15.02
Smooth	22.75	18.60	16.23	13.53
Oblique Loading Crestal Stress (MPa)				
Fine Threads	43.7	40.2	35.6	33.1
Smooth	36.99	29.99	25.66	20.5

Various studies on human patients have demonstrated that fine threads maintain the level of bone as around thread stress is of compressive nature and bone is strongest to compressive force compared to tensile load and shear force, which is present around smooth collar.[23,24] Implant design mainly affects the amount of bone loss. Though other causes of periosteal reflection, osteotomy preparation, autoimmune response to bacteria and biological width are not the primary causative agents after first two weeks of implant exposure and more than 1 mm of bone loss. Khon stated that local stain field within the bone around implant is inhomogeneous. He speculated that stain is highest at the tip of each thread and it decreased from the exterior to the interior region of the thread as shown in this study, where stress is concentrated at the top part of the body thread of the implant.

Bone loss around the smooth collar is due to the nature of shear force around the implant; FEA studies show that crestal bone loss with smooth collar design is often a consequence of disuse atrophy.[21] Our results suggest that when dental fixture is restored with an abutment of smaller diameter, it results in less stress transferred to the crestal bone, regardless of thread type (fin or smooth) or direction of force (vertical or oblique). This attributes to a greater distance between implant abutment junction and alveolar crestal bone, which can lead to protection from the microflora contained in the microgap.[25] When the size of abutment was reduced by 10%, 20%, and 30% the reduction in stress was 18.24%, 12.7%, and 19.95%, respectively, for smooth collar axial load. Three-dimensional finite element analysis, which examined biomechanical advantages of platform switching, concluded that it shifted the stress concentration area away from the cervical bone-implant interface, but it also has the disadvantage of increasing the stress in the abutment or abutment screw.[26] Stress in the abutment area was slightly higher: 12 MPa for 3.5-mm abutment compared to 9.20 MPa for the control 5.0-mm abutment. In accordance with other studies our study suggests that the fine threads on titanium surface extending to the implant shoulder provide osseous integration along the entire length of the implant.[27] Other studies also suggest that bone loss that occurred around implant was more related to implant placement depth rather than to platform switching.[28]

CONCLUSION

Fine threads increase crestal stress upon loading. Fine threads are helpful as they change shear stress to compressive or tensile stress, which increase bone resistance to load.[29]

When the concept of platform switching was applied by decreasing the diameter of the abutment, less stress was translated to the crestal bone in fine thread and smooth neck groups. Platform switching reduced stress to a greater degree in the thread model compared to the smooth neck model.

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