

Effect of Microthreads and Platform Switching on Crestal Bone Stress Levels: A Finite Element Analysis

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Background: The aims of this study were to investigate the effects of implant microthreads on crestal bone stress compared to a standard smooth implant collar and to analyze how different abutment diameters influenced the crestal bone stress level.

Methods: Two-dimensional finite element imaging was used to create a cross-sectional model of an implant (5-mm platform and 13 mm in length) placed in the premolar region of the mandible. The two tapered implant models consisted of one with microthreads at the crestal portion and the other with a smooth neck. The implant model was reverse-engineered to resemble a commercially available microthread implant. Abutments of different diameters (4.0 mm: 20% platform switching; 4.5 mm: 10% platform switching; and 5.0 mm: standard) were loaded with a force of 100 N at 90° vertical and 15° oblique angles. Finite element analysis was used to analyze the stress patterns in bone, especially in the crestal region.

Results: Upon loading, the microthread implant model had 29% greater stress (31.61 MPa in oblique and 9.31 MPa in vertical) at the crestal bone adjacent to the implant than the smooth-neck implant (24.51 and 7.20 MPa, respectively). When the abutment diameter decreased from 5.0 to 4.5 mm and then to 4.0 mm, the microthread model showed a reduction of stress at the crestal bone level from 6.3% to 5.4% after vertical loading and from 4.2% to 3.3% after oblique loading. The smooth-neck model showed a reduction of stress from 5.6% to 4.9% after vertical loading and from 3.7% to 2.9% after oblique loading.

Conclusions: Microthreads increased crestal stress upon loading. Reduced abutment diameter (i.e., platform switching) resulted in less stress translated to the crestal bone in the microthread and smooth-neck groups. *J Periodontol* 2008;79:2166-2172.

KEY WORDS

Alveolar bone loss; computer assisted; dental abutment; dental implants; numerical analysis.

The restoration of edentulous areas by using dental implants has been well documented and shown to have predictable outcomes.¹⁻³ The longevity of the implants relies primarily on their stability at placement.¹ However, early peri-implant bone loss has been observed in many implant systems and after different surgical approaches.^{1,3-5} In particular, crestal bone loss is the leading symptom of implant failure after osseointegration and the achievement of primary stability. The majority of crestal bone loss occurs during the first year of implant function, and it can be as much as 1.2 mm corono-apically.⁶ The presence of crestal bone is one of the key factors that influences the appearance or maintenance of peri-implant soft tissue architecture.

There are many suggested causes for early implant bone loss, two of them being occlusal overload and implant crest module.⁵ 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 An example of a suspected bone morphology alteration due to stress is the apical migration of crestal bone down to the first thread of many implant systems.^{1,5,8} It has also been hypothesized that the bone loss may slow down at the first thread when the force changes from crest shear force to the compressive force induced by the thread itself.⁵ In general, a functional implant may encounter many different

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forces, such as rotation, shear, and compression. It was found that the cortical bone layer withstands compressive force the best.⁹

Therefore, an implant system should be designed so that it can best 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 rely on the mechanical integrity of the prosthesis and implant¹⁰ as well as the ability of peri-implant structures to withstand and positively adapt to the applied forces.¹¹ Application of too much stress can cause bone resorption or even failure of the implant–bone interface, whereas lack of stress may lead to atrophy or even bone loss.^{12–14} Thus, the possibility of mechanical rearrangement leading to an increase or decrease in peri-implant bone structure quality is highly related to the magnitude and frequency of force applied to it.

One design concept is a rough external surface on the transosteal portion of an implant fixture. The mechanical benefit is an increase in implant–bone contact available for stress translation.¹⁵ One type of roughened surface results from the addition of microthreads^{||} to the neck. A clinical trial¹⁶ demonstrated possible preservation of crestal bone contact with implant systems using microthreads.

A second design strategy in implant systems is a switched platform abutment on an implant fixture. In essence, platform switching is the placement of a smaller-diameter abutment relative to the platform of the fixture. Primarily associated with the internalization of the microgap,⁵ this design has also been suggested as beneficial for crestal bone preservation.¹⁷ Because platform switching involves a change in the implant system design structure, this feature may also have a role in stress translation from implant to bone. In this study, the term “platform switching” refers to a reduced-diameter abutment in relation to the diameter of the fixture platform.

To investigate the influence of the crestal module, specifically in the platform-switching systems, a computer model was developed to apply finite element analysis (FEA). The use of FEA in implant analysis has been widely demonstrated and published.¹⁸ Because the geometries involved with modeling implants and the alveolar process are very complex, FEA is considered the most suitable tool for analyzing them. This type of analysis allows researchers to predict stress distribution in the contact area between implants and cortical bone as well as around the apex of the implants in trabecular bone.¹⁸ Therefore, the aims of this study were to investigate the effects of implant microthreads on crestal bone stress levels compared to a standard smooth implant collar and to analyze how different abutment diameters influence crestal bone stress levels.

MATERIALS AND METHODS

A two-dimensional finite element model (Fig. 1) of a completely osseointegrated endosseous titanium implant in the posterior mandible was created for the purposes of stress analysis in conjunction with design alterations. FEA software[¶] was used to generate the model, create the mesh of the individual elements, and perform the analysis of the resulting models.

The model of the posterior mandible in cross-section was constructed using measurements and geometries similar to another study¹⁹ with isotropic material properties (Table 1).^{20–22} An isotropic material is defined as having identical physical properties in all directions; therefore, only two independent material constants exist. This model consisted of thick cortical bone surrounding dense trabecular bone, which is classified as type II bone.²³ The resulting mandibular cross-section was 28 mm in vertical dimension and 14.1 mm at the greatest horizontal dimension. The thickness of the cortical bone section ranged from 0.595 to 1.515 mm, with the crestal region measuring 1.5 mm.

The smooth-neck (control fixture) implant model consisted of a restorative platform width of 5 mm, a crestal width of 5.5 mm, and a length of 13 mm. The dimensions of the implant model selected in this study represent one of the most commonly used sizes in clinical practice. The microthread (test fixture) implant model was replicated with the exception of microthreads replacing the smooth-neck portion. The original diameter was maintained throughout this portion. Microthreads and main body threads were modeled as V-shaped projections in which the microthreads and body threads had a distance of 0.2 and 0.8 mm, respectively. The smooth and microthread neck portion was 4.8 mm in height and modeled to be in 100% contact with surrounding structures. The 3-mm tall abutment models consisted of 5-, 4.5-, and 4-mm diameters.

Complete osseointegration at the implant–bone interface was simulated by combining the nodes of the implant and bone models. Similar integration of the abutment and implant body was adopted to be a single unit. This eliminated any potential influence from the micromovement between components.

The mesh consisting of the implant body, abutment, and bone consisted of 42,159 elements and 41,902 nodes, which was for control and test fixtures with 5-mm abutments. A two-dimensional (2D) plane stress element and h-method discretization were used for computation and analysis of the model. Boundary conditions were modeled to fix the inferior one-third of

|| Astra Implant ST, Astra Tech, Mölndal, Sweden.

¶ NISA V15, Cranes Software, Troy MI.

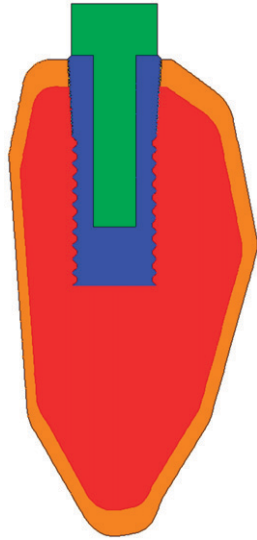


Figure 1.

Implant and surrounding mandibular bone cross-section used for finite element modeling and analysis. Microthread model shown, with orange and red sections representing cortical and trabecular bone, respectively. The implant and abutment are shown as blue and green, respectively.

Table 1.

Comparison of Material Properties

Component	Modulus of Rigidity (MPa)	Poisson Ratio
Titanium (Sakaguchi and Borgersen, 1995 ²⁰)	117×10^3	0.30
Trabecular bone (Borchers and Reichart, 1983 ²¹)	1,370	0.31
Cortical bone (Rice et al., 1988 ²²)	2,727	0.30

the mandibular model in all degrees of freedom to minimize stress interferences.

Model analysis consisted of two groups: smooth neck (control) and microthread (test), each with three abutment diameters (5, 4.5, and 4 mm). The 5-mm abutment represented the control diameter; the 4.5- and 4-mm abutments represented a diameter reduction of 10% and 20%, respectively. This reduction in abutment diameter represented the concept of platform switching.

Force application was performed in oblique and vertical conditions using 100 N as a representative masticatory force.^{11,19,24,25} For oblique loading, a force of 100 N was applied at 15° from the vertical axis. This translated into ~26.8 N in the horizontal direction and 96.3 N in the vertical direction. In both cases, the transfer of load was simulated to be from

the center apical surface of the abutment, through the implant body, to the peri-implant tissues.

Stress analysis of all models consisted of mapping von Mises stress patterns upon the application of vertical and oblique loading scenarios. von Mises stress, a type of applied stress analysis used in FEA, is measured in MPa. The values used in the comparisons were located at the most crestal cortical bone, adjacent to the implant fixture, in the mandibular model.

RESULTS

In all models, the greatest concentration of stress was located at the crestal level adjacent to the implant and at the points of the microthreads and body threads. The localized crestal stress adjacent to the implant model decreased drastically within a horizontal distance of 3 mm. Figures 2 and 3 show the crestal portion of the model adjacent to the smooth neck or microthreads, respectively.

Upon oblique loading, the microthread implant model with a control abutment had 29% greater stress at the crestal bone adjacent to the implant than the smooth-neck implant. Under vertical load, the microthread model displayed a similar 29% increase in crestal bone stress. Specifically, von Mises stresses in the microthread model were 31.61 MPa for oblique loading and 9.31 MPa for vertical loading; in the smooth-neck model they were 24.51 and 7.20 MPa, respectively.

When the abutment diameter decreased from 5.0 to 4.5 mm, and then to 4.0 mm, the microthread model showed a reduction in crestal bone stress levels from 6.3% to 5.4% after vertical loading and from 4.2% to 3.3% after oblique loading, respectively. The stress level with the smooth-neck model decreased from 5.6% to 4.9% after vertical loading and from 3.7% to 2.9% after oblique loading (Table 2).

DISCUSSION

The purpose of this study was to use a two-dimensional computer model (e.g., FEA) to analyze the effect of microthreads and platform switching on implant crestal bone stress where the greatest stress was noted. Although 2D and three-dimensional (3D) models have their advantages and disadvantages, this study used a 2D model because of ease of manipulation, simplicity, efficiency, and cost-effectiveness. The aim of this study was not to replicate exact in vivo stresses, but rather to illustrate a possible difference between a microthread and a smooth-neck implant counterpart coupled with abutments simulating platform switching. Because this study was modeled in 2D, antirotational elements of an implant system were not considered crucial and were not implemented into the design model, because the force vector was also in a 2D plane. The rationale behind the omission of

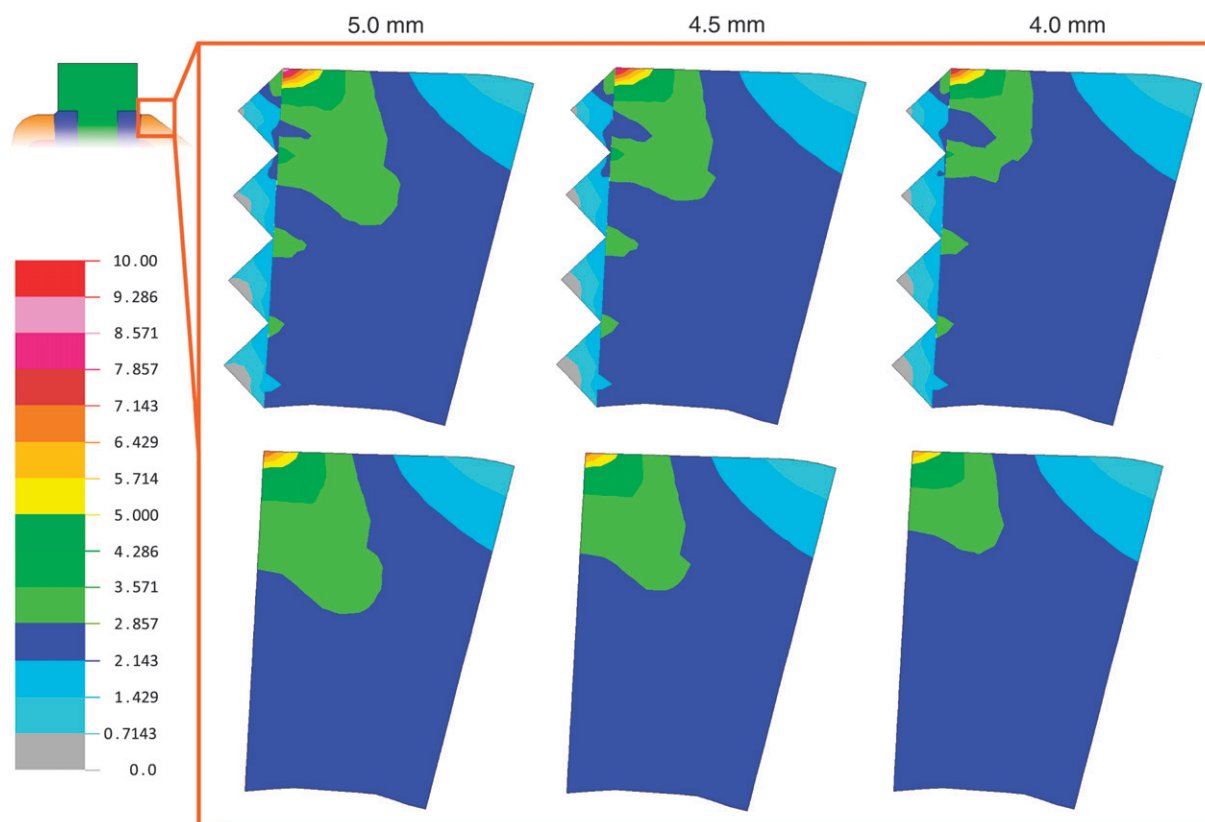


Figure 2.

Vertical loading stress plots. Top row shows microthread model; bottom row shows smooth-neck model. Key shown in von Mises stress (MPa).

design elements, such as a Morse taper connection system, was due to the fact that our model was designed to have node sharing between the abutment and implant body. Because of this mating condition, there is little significance to the shape of the abutment that inserts into the implant body. Micromovement among components of this model was also omitted. Although micromovement plays a vital role in implant stability, it was determined that micromovement would have been consistent between different models and it would have been redundant to create in each model. Thus, the simplicity of the modeled abutment, implant, and mandibular cross-section was sufficient to demonstrate the effects of microthreads and platform switching without adding complicated design parameters not examined in this study.

This 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 porosity, which is found in *in vivo* trabecular bone; this is in agreement with a study done by Akagawa et al.,²⁶ in which trabecular bone was modeled as a solid substance and a more natural porous substance. Using the isotropic property instead of anisotropic bone properties for trabecular and cortical

bone may have had an effect on the results.¹⁹ This study also modeled the bone–implant contact as a consistent 100% to create a modern model similar to photoelastic models, whereas most common bone–implant contact percentages ranged from 30% to 70%. Because the goal of this study was to investigate the effects on surrounding bone when only two design aspects (thread pattern 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, 2D 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. A 3D model may demonstrate varying amounts of stress in all planes. However, in this 2D model, we compared stress differences between the abutment and implant models because the 2D model only showed one plane of the stress pattern. This was much easier to analyze based on our study objectives.

For the analysis portion of this study, it was determined that a vertical and oblique loading model should be tested. A 15° angle and a loading force of 100 N were chosen because this force was shown in other studies^{11,19,24} to be more comparable to *in vivo*

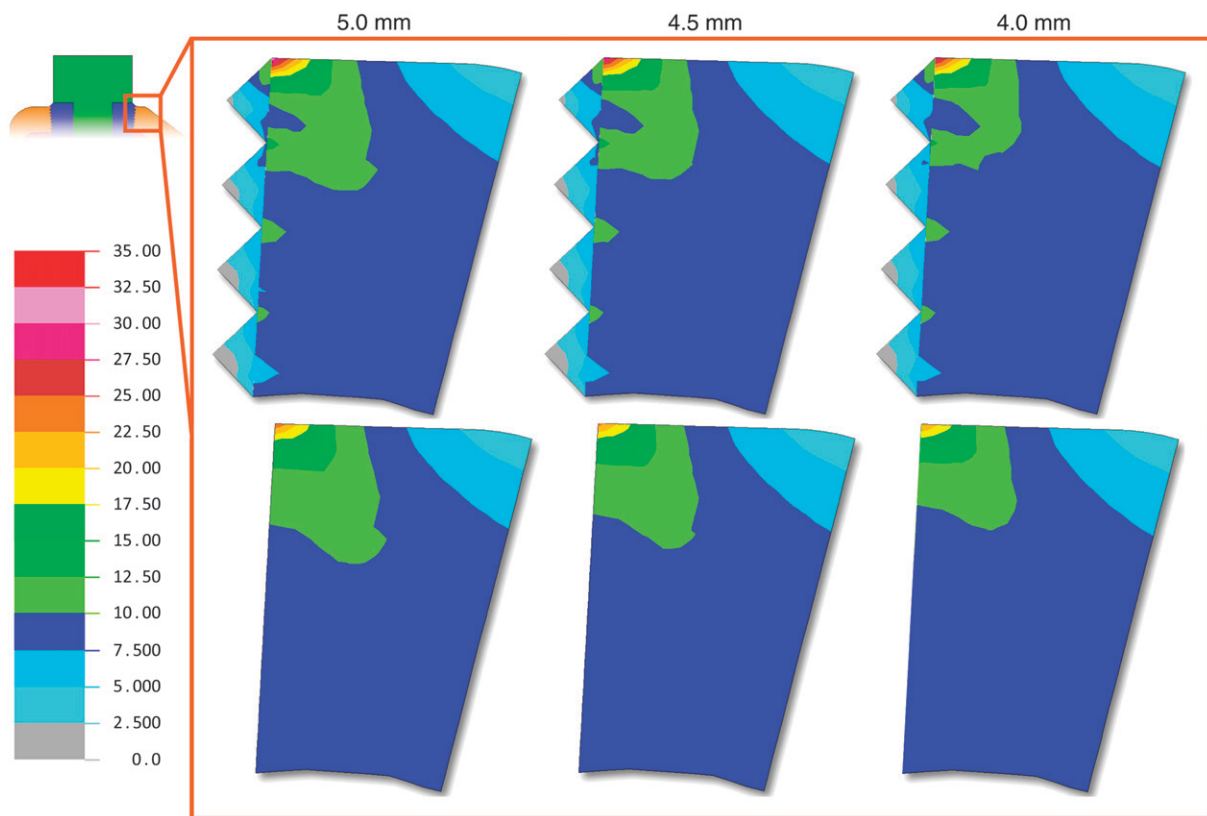


Figure 3.

Oblique loading stress plots. Top row shows microthread model; bottom row shows smooth-neck model. Key shown in von Mises stress (MPa).

Table 2.

Results from FEA Analysis for Oblique and Vertical Loading Conditions

Crestal Stress Adjacent to the Implant Neck (MPa)			
Abutment diameter	4.0 mm	4.5 mm	5.0 mm
Vertical loading			
MT	8.26	8.72	9.31
SN	6.48	6.80	7.20
Oblique loading			
MT	29.29	30.30	31.61
SN	22.93	23.62	24.51

MT = microthread model; SN = smooth-neck model.

mastication and is a biologically feasible action that can be performed on an implant *in vivo*.^{24,25} Also, the application of a horizontal vector creates the most shear stress in cortical bone and was shown to be the component of force best avoided for implant success.¹⁸ To augment the oblique condition, an additional model, with vertical loading of 100 N, was created. Although these forces and angle represent possible applications of force to a dental implant,

the actual vector of force can vary among individuals.^{10,27} 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.

Data from this study showed that the microthread implant had 29% greater stress at the crestal bone adjacent to the implant than the smooth-neck implant in oblique and vertical load. This result was proportionally similar to another study¹⁹ that used similar angles and load applications. The range (6.48 to 24.51 MPa) of cortical von Mises stress seen in our study may be slightly higher than in the *in vivo* condition and can be attributed to the linear FEA modeling used in the study. Thus, an increase or decrease in force application would have respective effects on the data produced. Nonetheless, based on histologic examination and FEA results, previous studies^{14,27} 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, 60 MPa was regarded as a threshold strain value above which bone failed to heal after fatigue.²⁸ Although our data were not modeled to be in complete compliance with an *in vivo* model, the design model used in this study is clinically supported by these findings. Lee

et al.²⁹ demonstrated in human patients that there may be maintenance of marginal bone loss in microthread implants over smooth-neck systems. The rationale behind the claim of microthread implants maintaining crestal bone while translating higher stress may be explained by the compressive nature of stress translated by threads, as suggested by Oh et al.⁵ This remains to be verified in future in vitro studies.

In addition, our data suggested that when the abutment diameter was reduced by 10% or 20%, it resulted in less stress transferred to the crestal bone, regardless of the type of thread pattern (microthread or smooth) or direction of force (vertical or oblique). This is in support of the hypothesis that suggests that platform switching may increase the distance between the abutment inflammatory cell infiltrate and the alveolar crest, thus reducing the abutment design's bone-resorptive effect.³⁰ Furthermore, our results also confirmed the speculation made by Lazzara and Porter,³¹ who found that repositioning the implant-abutment interface away from the crestal bone into a more confined area reduced bone resorption at the crestal bone level. However, a recent animal study³² indicated that a smaller-diameter abutment (platform switching) reduced crestal bone loss only for 28 days. After 28 days the matching and platform-switching abutments showed the same amount of bone loss. This finding was further supported by a recent study published by Jung et al.,³³ who showed implants with non-matching implant-abutment diameters (platform switching) demonstrated some bone loss; however, it was a small amount. The bone loss that occurred in the study was more related to implant placement depth rather than to platform switching. For example, the greatest bone loss occurred when the implant-abutment junction was placed 1 mm below the bone crest.³³ These conflicting reports suggest that there is a need for more studies to validate the influence of platform switching on crestal bone level.

CONCLUSIONS

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

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