Mechanical Basis for Bone Retention Around Dental Implants

Harold Alexander,[1] John L. Ricci,[2] George J. Hrico[3]

[1] Orthogen Corporation, 505 Morris Avenue, Suite 104, Springfield, New Jersey 07081

[2] Department of Biomimetics and Biomaterials, New York University College of Dentistry, New York, New York

[3] Design Engineering Analysis Corporation, Canonsburg, Pennsylvania

Received 4 October 2006; revised 7 March 2007; accepted 7 March 2007
Published online 23 April 2007 in Wiley InterScience (www.interscience.wiley.com). DOI: 10.1002/jbm.b.30845

Abstract: This study, analytically, through finite element analysis, predicts the minimization of crestal bone stress resulting from implant collar surface treatment. A tapered dental implant design with (LL) and without (control, C) laser microgrooving surface treatment are evaluated. The LL implant has the same tapered body design and thread surface treatment as the C implant, but has a 2-mm wide collar that has been laser micromachined with 8 and 12 μm grooves in the lower 1.5 mm to enhance tissue attachment. *In vivo* animal and human studies previously demonstrated decreased crestal bone loss with the LL implant. Axial and side loading with two different collar/bone interfaces (nonbonded and bonded, to simulate the C and LL surfaces, respectively) are considered. For 80 N side load, the maximum crestal bone distortional stress around C is 91.9 MPa, while the maximum crestal bone stress around LL, 22.6 MPa, is significantly lower. Finite element analysis suggests that stress overload may be responsible for the loss of crestal bone. Attaching bone to the collar with LL is predicted to diminish this effect, benefiting crestal bone retention. © 2007 Wiley Periodicals, Inc. J Biomed Mater Res Part B: Appl Biomater 88B: 306–311, 2009

Keywords: biomechanics; bone remodeling; dental/endosteal implant; finite element analysis; implant interface

INTRODUCTION

The final outer finish of a dental implant plays an important role in the ability of bone to grow on the implant surface. There are a great variety of finishes presently in use throughout the dental implant industry. For example, machined, acid etched, laser machined, blasted, and alloys coated with materials that induce bone activity are currently in use. Combinations of these techniques are often used to optimize implant fixation. For example, Szmukler-Moncler et al.[1] have shown that sandblasting with large grit followed by acid etching (SLA) increases osseointegration (bone growth on the surface) by 50% after 10 weeks of healing. A machined finish creates grooves on the order of 0.5–1 μm. Once a machined finish is polished, the surface of the implant is smooth to the nanometer level. On the cellular level, neither a machined finish nor a polished finish provides a surface with a texture promoting osseointegration. A blasted and etched implant creates small flaws of 2–20 μm in size,[1] which bone cells are able to grow into and then produce a mineralized matrix. This provides a better means for implant attachment and leads to a higher percentage of osseointegration over the surface of the implant.

To produce an ordered, less random microtexture, the authors of this article have utilized laser texture grooving with an Eximer laser. This study explores the possibility of a mechanical explanation for the effectiveness of this novel approach to reduce crestal bone resorption around dental implants. In a previous series of *in vitro* experiments, the effects of various laser-machined substrate microstructures on the attachment, spreading, orientation, and growth of fibroblast and osteoblast cell types was examined.[2] The most important result to arise from these studies was the development of a series of microgrooved surfaces with groove widths and depths in the range of 6–12 μm. These surfaces, it is hypothesized, facilitate stress transfer from the implant to the supporting bone. The resultant topography triggers changes in cytoskeletal structure, stimulation of tyrosine phosphorylations, and expression of m-RNA for fibronectin.[3] The experimental surfaces were found to optimally control orientation of attached cells, prevent cell migration perpendicular to the microgrooves, and substantially inhibit fibroblast growth by inhibiting cell spreading. Specifically, 12-μm grooves showed the best potential for inhibition of fibrous tissue growth relative to bone cell

Correspondence to: Harold Alexander (e-mail: halexander@orthogencorp.com)

© 2007 Wiley Periodicals, Inc.

growth, and 8-μm grooves showed the most effective inhibition of cell migration across the grooves, in effect acting as a migration barrier. These surfaces were also found to directionally inhibit migration of epithelial cells.

Another study examined bone and soft tissue response to these experimental surfaces as well as to blast-roughened surfaces of the same metal composition in a canine implantable chamber model.[4] The laser-microgrooved surfaces exhibited less fibrous encapsulation and more extensive bone integration than their blast-roughened counterparts, as well as orientation of adjacent bone microstructure. The surface microgrooves exhibited interdigitation with bone resulting in a mechanical interlock whose integrity was demonstrated in tensile testing to stress levels as high as 10 MPa.

These *in vitro* and *in vivo* studies provide strong evidence that surface microtexturing can be utilized to control bone and soft tissue response to the implant surface. In the case of smooth implant surfaces, fibroblasts attach, spread, and proliferate readily, resulting in formation of a fibrous capsule that restricts bone formation. Ridge-groove widths of the order of magnitude of the cells themselves guide cell migration and orientation.[5] Microgrooved surfaces optimally restrict apical migration of fibroblast proliferation spreading, allowing the slower-growing osteoblasts to proliferate and migrate coronally along the implant collar. It has been hypothesized that stress transfer from the implant to the crestal bone interlocked through the microgrooved surface of the collar produces a stable crestal bone topography less likely to show resorption during the initial year of function.[6,7]

Dental implants are designed to bear the loads caused by teeth during mastication. The goal is to have the maximum amount of bone engaged with the body of the implant, and thus provide the most stability. It is well documented throughout the literature that crestal bone loss averages more than 1 mm in the first year, and at least 0.10 mm each following year.[8] Accumulation of crestal bone loss over the lifetime of an implant affects the load-bearing capability of the implant and leads to cosmetic problems or implant failure. Crestal bone loss results from bone response to biological factors present at the bone–implant interface and bone response to mechanical factors of loading. As a result, the crestal bone around the implant takes a saucer-like shape, which continues to become more pronounced as time progresses.

Dental implants are loaded in multiple ways. Teeth are subject to axial loads, bending and twisting moments, shear forces, and a combination of any or all of these loading mechanisms. The transfer of loads from the implant to bone, along with the stress ranges created by the loads, is assumed to affect the osseointegration and bone remodeling around the implant. Mechanosensing theory, evidence supporting the concept, and how it applies to the levels of loading along the bone–implant interface have been the subjects of many research efforts.

Stress adaptation of bone was first hypothesized by Wolff's Law in the 1880s.[9] There certainly is evidence in animal evolution that bone has a mechanism of performing mechanosensory functions. It is not established whether the mechanosensitivity mechanism is governed by the local stress, the local strain or the frequency of the loading phenomenon; or a combination of the above. The work of Mosley and Lanyon[10] seems to argue for a strain rate response for bone remodeling. There is evidence that bone responds to local strain[11] or to stress,[12,13] but, is it the total stress that governs or a component; deviatoric or dilatational?

Burger and Klein-Nulend[14,15] proposed there is a bone cell network that links the bone cell signal due to strain to a cellular signal, which causes bone resorption or bone formation. Mechanotransduction in bone is described by them through the following mechanism: Bone loading → Matrix strain → Mechanosensing by bone cells → Bone formation by osteoblasts or bone resorption by osteoclasts. Within bone lies a complex three-dimensional network of lacunae and canaliculi. These micropores are filled with interstitial fluid that supply the bone cells with nutrients and provide a means for the bone cells to sense mechanical changes. As bone is stressed, the interstitial fluid flow causes mechanical shear stress and strain-generated potentials. By measuring the production of anabolic factors, nitric oxide, and prostaglandins, it has been experimentally shown that osteocytes have the ability to sense fluid flow and communicate intracellularly. Furthermore, the presence of nitric oxide and prostaglandins during bone loading has been correlated with other cellular reactions, i.e. endothelial cells in blood vessels, which require intracellular communication. The recruited bone cells are transferred through the lacunocanalicular porosity to the necessary area of the bone. A combination of the anabolic messages and cell transfer brings either the osteoclasts to remove bone or osteoblasts to form bone.

The mechanotransduction mechanism as described by Burger[14] can be used to explain the cellular activity involved with remodeling bone under various load conditions. Under normal load conditions, bone has just enough mechanical stimulation to provide osteocytes with nutrients and waste removal. Disuse of bone leads to extremely low mechanical stimulation and nearly zero fluid shear flow. A low level of fluid shear does not create the necessary flow needed for nutrient supply and waste removal. As a result, disuse causes osteocyte death, recruitment of osteoclasts, and elimination of bone until the cell supporting fluid shear returns to normal and re-establishes osteocyte activity.[14] Damage to the network from overstress (beyond the yield stress) also leads to osteocyte death and consequent recruitment of osteoclasts and bone destruction. The von Mises criterion, also known as the octahedral shearing stress theory, predicts failure by yielding when the octahedral shearing stress at a point achieves one half the maximum principle stress at yielding.[16] The von Mises stress, $s_{0,}$ is

defined as:

$$s_0 = \sqrt{1/2[(s_1 - s_2)^2 + (s_2 - s_3)^2 + (s_3 - s_1)^2]}$$

where s_1, s_2, and s_3 are the principal stresses.

The von Mises stress, a commonly used measure of distortional stress, would appear to be a useful parameter to assess the bone–implant interface. The finite element method is used here to model the von Mises stresses present along the bone–implant interface to assess the effect of more aggressive attachment strategies.

MATERIALS AND METHODS

Crestal bone response to dental implants with and without a bone attaching collar has been evaluated through finite element stress analysis (FEA). The implant is a tapered design (C) with and without a laser micromachined collar (LL) tradenamed LaserLok™ (BioLok International, Deerfield Beach, FL). This tapered implant has a reverse buttress thread design. The LL implant has a collar that has been laser micromachined with 12-μm wide grooves that have been previously shown to optimize the surface for bone attachment.[2,4]

Finite element analysis (FEA), also known as finite element modeling (FEM), was developed to perform structural analysis on complex shapes. In structural engineering, FEA provides the ability to predict failure ranges by stress analysis during early stages of design.

For the purposes of this analysis, the implants were assumed to be 13-mm long with 4-mm diameter collars. To simulate high mastication forces in the esthetic zone (front teeth), 80 N of axial loading and side loading applied 6 mm above the top of the collar were analyzed. All implants were assumed to be manufactured of titanium alloy and were placed in a "bony bed" of cancellous bone with a 1-mm thick cortical shell, similar to what is found in the human mandible.

Material properties were assumed to be isotropic and linear elastic. A cortical bone modulus of 15 GPa, a cancellous bone modulus of 1.5 GPa, and a titanium alloy modulus of 110 GPa were assumed. Poisson's ratio of 0.3 was used for both bone materials and a value of 0.35 was assumed for the titanium implant. A three-dimensional, half-symmetry model with 130,000 elements was developed using the ANSYS finite element program. This model is shown in Figure 1. The titanium implant is excluded in this figure for clarity. All of the elements in the model are 10-node tetrahedrons, which have three translational degrees-of-freedom at each node. These elements have quadratic displacement behavior and are well suited for modeling irregular meshes. In establishing the appropriate mesh geometry, a standard test for convergence was performed. This resulted in the finer mesh elements shown in the

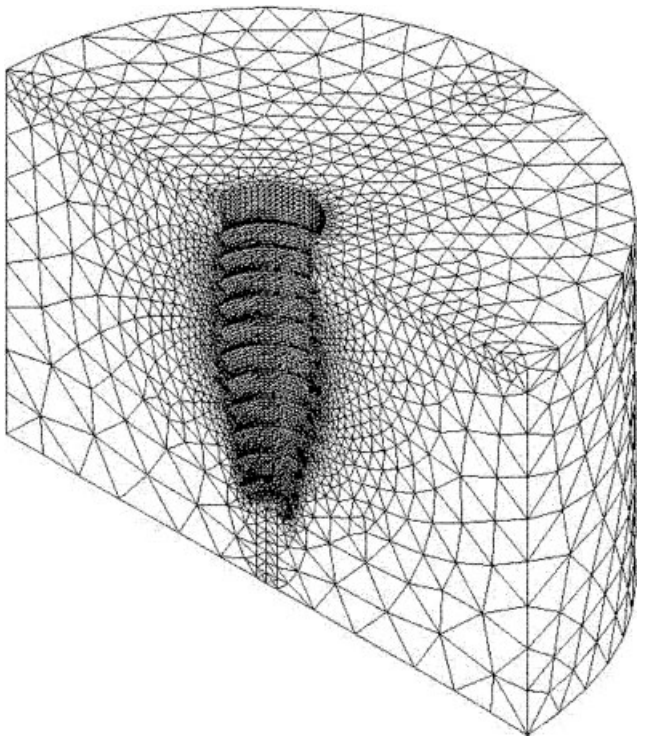

Figure 1. Three-dimensional, half-symmetry finite element model with 130,000 elements.

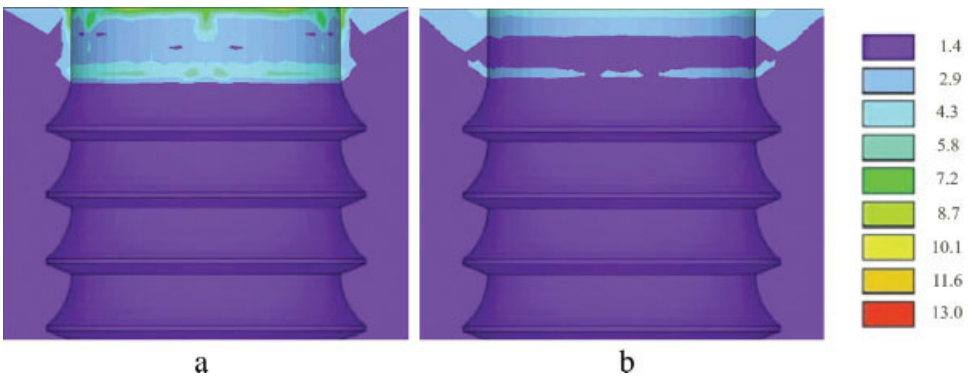

Figure 2. von Mises stress distribution around collar of the C (a) and LL (b) implants (80 N axial load). C, Max Stress 8.9 MPa; LL, Max Stress 4.0 MPa; Scale (MPa).

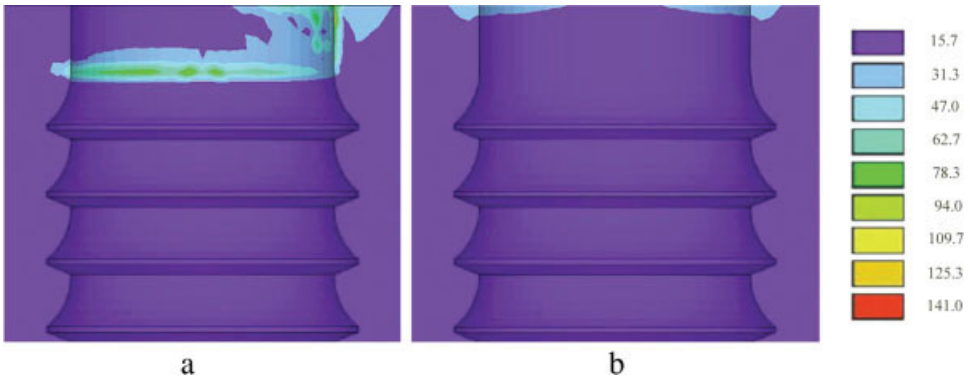

Figure 3. von Mises stress distribution around collar of C (a) and LL (b) implants (80 N side load). C, Max Stress 91.9 MPa; LL, Max Stress 22.6 MPa; Scale (MPa).

implant interface regions. A cylindrical outer boundary was assumed remote from the implant region. This boundary was restrained to prevent rigid body motion of the model under loading. On the symmetry plane, only translational displacements normal to the symmetry plane were fixed. The top surface of the model, which represents the free surface of the cortical bone, was left unrestrained.

In the model, a bonded (osseointegrated) interface was assumed between the threads and the cancellous bone. Bonding was achieved by merging coincident nodes at this interface so that the thread elements and adjacent bone elements shared common nodes. Two different interfaces were applied in the collar region; a nonbonded condition and a bonded condition. The nonbonded condition was used to simulate the "as machined" surface and the bonded condition was assumed to simulate the laser machined surface, since the *in vivo* testing in a canine model[4] yielded tensile separation stresses approaching those encountered in the human *in vivo* situation. The nonbonded condition required that interface elements be included in the model so that contact and separation was possible between the implant collar and the cortical bone. Interface element convergence was achieved via a nonlinear solution. The von Mises stress was computed as a "bone damage indicator." Detailed postprocessing of the analysis results was performed for the bone stresses around the top 5 mm of the implants in the collar region.

RESULTS

In a typical dental implant FEA stress analysis, one sees stress concentrations in the collar region and the base of the screw. The collar region stress concentration is accentuated in this case because of the effect of the pulling down of the cortical shell with a central hole.

In the collar region, when exposed to an 80 N axial load, the crestal bone adjacent to the C implant, shown on the left in Figure 2, is exposed to higher stresses than is the bone adjacent to the LL implant, shown on the right. However, these stresses are still in a very low range.

The major effect is seen in side loading (Figure 3). When exposed to the 80 N side load, the maximum crestal bone stress around the Control (a) (91.9 MPa) implant approaches maximum bone stress levels. However, the maximum von Mises stress in the cortical bone around the collar of the LaserLok implant (b) (22.6 MPa) is significantly lower. For this loading configuration, the maximum crestal bone distortional stress adjacent to the implant is lowered significantly by crestal bone attachment.

Figure 4 summarizes the results of this finite element stress analysis. In axial loading, the LL has a 55% maximum stress advantage over C. In side loading, the LL has a 75% maximum stress advantage over C.

DISCUSSION

The finite element mechanical analysis predicts that the maximum bone stress and strain occur in the collar region. This high relative motion from distortional stress overload can result in loss of crestal bone and fibrous tissue formation. Early in the history of dental implants it was believed fibrous encapsulation was optimal, therefore the loading timeline was not a concern. Fibrous tissue formation is

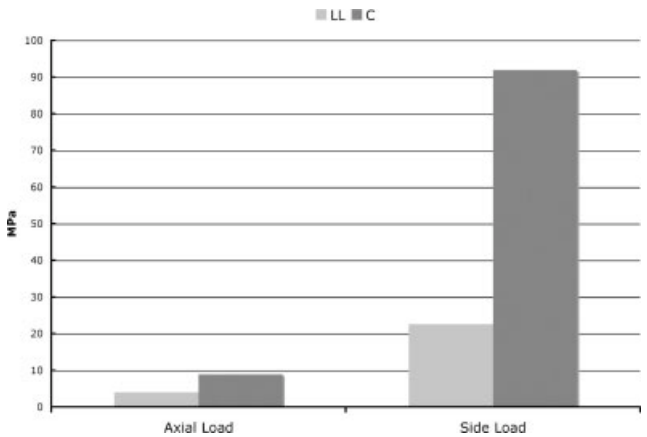

Figure 4. Summary of the results of finite element analysis demonstrating the stress decrease resulting from implant attachment.

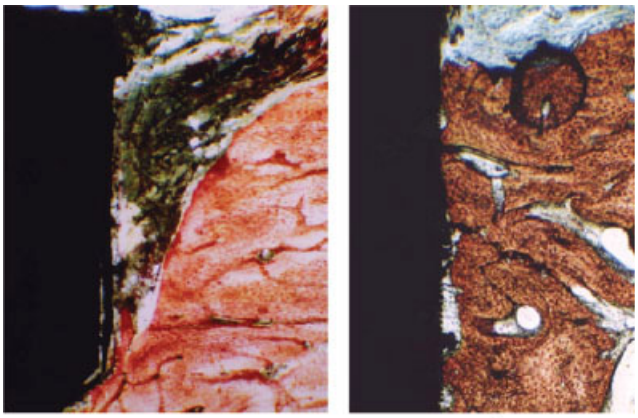

Figure 5. Tissue response to C and LL collars at 9 months postimplantation in a canine model (Reproduced from Ref. 20, with permission from Lippincott Williams and Wilkins). (a) C Histology, (b) LL Histology.

believed to occur via the following mechanism. During early bone healing, micromotion damages the tissue and vascular structure. ''Micromotion probably interferes with the development of an adequate early scaffolding from a fibrin clot, and disrupts the re-establishment of a new vasculature to the healing tissue, which in turn interferes with the arrival of regenerative cells. Eventually, the healing process is rerouted into repair by collagenous scar tissue instead of regeneration of bone.''[17] The resulting encapsulation is significantly inferior to adequate bone response at the bone–implant interface.

As understanding evolved, it was realized that the longevity of implants depended on the quality of bone fixation. As a result, implants were placed and given a long period of time, 3–6 months, under low stress to allow for sufficient osseointegration. In the 1970s, the concept of micromotion affecting bone response at the implant interface was introduced. In the years leading up to the present, many expansions on the concept of micromotion have been made. For example, it was first shown by Brunski in 1979 that early loading could result in fibrous encapsulation of the implant. Furthermore, it is now known that there is a threshold for excessive micromotion between 50 and 150 μm, above which bone formation turns to fibrous encapsulation.[18] It also appears that the threshold value for micromotion is a function of implant design and surface characteristics. To this date, research is still being done to investigate micromotion and eliminate the unfavorable condition of fibrous encapsulation. Recently, Pilliar et al.[19] also utilized finite element analysis to investigate the crestal bone stress state around porous-surfaced implants *versus* machined threaded dental implants. They concluded that the observed greater retention of crestal bone next to porous-surfaced implants was attributable to lower peak stresses developing in crestal peri-implant bone.

In the present study, aggressively attaching bone to the collar with the LaserLok^{TM} design is predicted to diminish crestal bone stress and, therefore, this fibrous tissue formation effect. This may be the explanation for crestal bone retention. Canine implantation study results reported by Weiner[20] appear to bear out this proposition. Figure 5 demonstrates the difference in crestal bone response between the two different implant collars. Since the canine uses the teeth to shear food rather than grind it, the comparison with the side loading state would appear to be appropriate. The higher stresses predicted with the unattached collar,

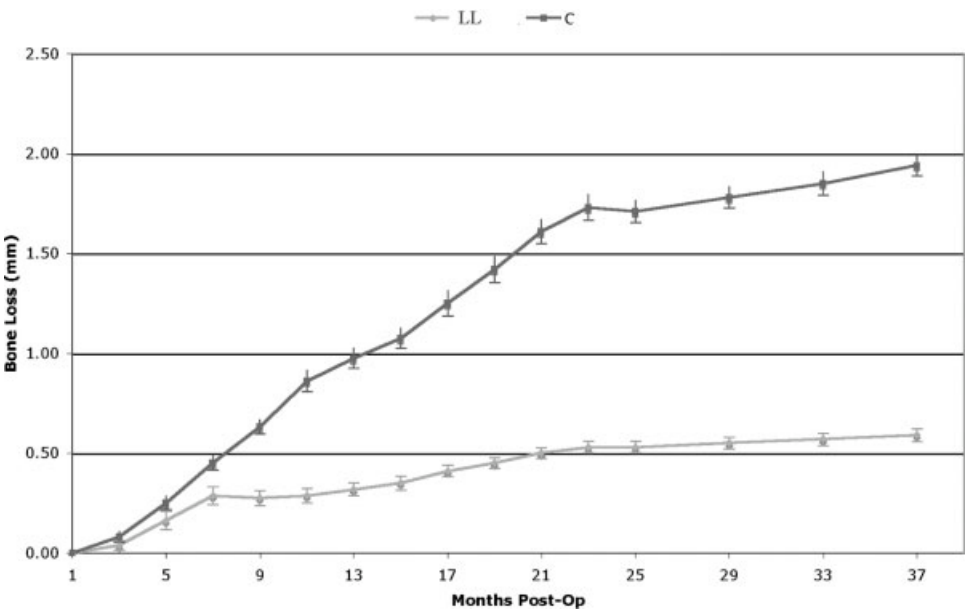

Figure 6. Crestal bone loss LL *versus* Control in a 3-year prospective, controlled clinical study. Error bars = standard error: $p < 0.005$ after month 5 (Reproduced from Ref. 7, with permission from Lippincott Williams and Wilkins).

should result in bone loss. This is born out by the histology shown in Figure 5(a). The low stress predicted with the attached collar should be crestal bone protective. This is also born out by the histology shown in Figure 5(b).

In the collar region, a critical bony area because of the higher potential for high bone stresses and relative motion, the laser microgrooving of the LL implant demonstrated its superiority. The finite element analysis prediction is for lower bone stress in this region.

Clinical testing was performed in a prospective, controlled 37-month study.[7] Each patient received two single tooth implants (LL *vs*. C). The study was performed with a total of 15 patients who received 20 sets of implants. The crestal bone loss data are the most dramatic result of this study. The differences between the LL and C implants were tested at each study visit by a paired *t* test resulting in *p*-values <0.005 for all time periods after 5 months post-op. As is shown in Figure 6, the LL bone loss is limited to the 0.6-mm range, while the Control data (C) demonstrates up to almost 2 mm of bone loss.

CONCLUSION

High surface-area and organized micro textures have been applied that encourage bone integration. This finite element analysis, investigating the local distortional stresses in the crestal bone area adjacent to the implant, provides a further mechanical explanation for the superiority of this laser micromachined surface in retaining bone in the critical crestal area of a dental implant.

REFERENCES

1. Szmukler-Moncler S, Perrin D, Ahossi V, Magnin G, Bernard J-P. Biological properties of acid etched titanium implants: Effect of sandblasting on bone anchorage. J Biomed Mater Res B 2004;68:149–159.
2. Ricci J, Charvet J, Frenkel S, Chang R, Nadkarni P, Turner J, Alexander H. Bone response to laser microtextured surfaces. In: Davies JE, editor. Bone Engineering. Toronto: Em2 Inc.; 2000. pp 282–294.
3. Wojciak B, Curtis ASG, Monaghan E, Macdonald K, Wilkinson CDW. Guidance and activation of murine macrophages by nanometric scale topography. Exp Cell Res 1996;223:426–435.
4. Frenkel SR, Simon J, Alexander H, Dennis M, Ricci JL. Osseointegration on metallic implant surfaces: Effects of microgeometry and growth factor treatment. Canine study of bone response to LaserLok surfaces. J Biomed Mater Res B 2002;63:706–713.
5. Wojciak B, Crossan J, Crutis ASG, Wilkinson CDW. Groomed substratia facilitate in vitro healing of completely divided flexor tendons. J Mater Sci Mater Med 1995;6:266–271.
6. Norton MR. Marginal bone levels at single tooth implants with a conical fixture design. The influence of surface macro- and microstructure. Clin Oral Implants Res 1998;9:91–99.
7. Pecora GE, Ceccarelli R, Bonelli M, Alexander H, Ricci JL. Clinical evaluation of laser microtexturing for soft tissue and bone attachment to dental implants. Implant Dent. Forthcoming.
8. Small PN, Tarnow DP. Gingival recession around implants: A 1-year longitudinal prospective study. Int J Oral Maxillofac Implants 2000;15:527–532.
9. Wolff J. Das Gesetz der Transformation der Knochen. Berlin: Hirschwald; 1892.
10. Mosley JR, Lanyon LE. Strain rate as a controlling influence on adaptive modeling. Bone 1998;23:313–318.
11. Prendergast PJ, Taylor G. Prediction of bone adaptation using damage accumulation. J Biomech 1994;27:1067–1076.
12. Cowin SC. Bone stress adaptation models. J Biomech Eng B 1993;115:528–533.
13. Cowin SC, Moss ML.Mechanosensory mechanisms in bone. In: Cowin SC, editor. Bone Mechanics Handbook. Boca Raton: CRC Press LLC; 2001. pp 29-1 –29-17.
14. Burger EH. Experiments on cell mechanosensitivity: Bone cells as mechanical engineers. In: Cowin SC, editor. Bone Mechanics Handbook. Boca Raton: CRC Press LLC; 2001. pp 28-1–28-16.
15. Burger EH, Klein-Nulend J. Mechanotransduction in bone-role of the lacuno-canalicular network. FASEB J 1999;13 (Suppl):S101–S112.
16. Ugural AC, Fenster SK. Advanced Strength and Applied Elasticity. New York: Elsevier; 1975. pp 105–106.
17. Brunski JB. In vivo bone response to biomechanical loading at the bone/dental-implant interface. Adv Den Res 1999;13: 99–119.
18. Szmukler-Moncler S, Salama H, Reingewirtz Y, Dubruille JH. Timing of loading and effect of micromotion on bone-dental implant interface: Review of experimental literature. J Biomed Mater Res B 1998;43:192–203.
19. Pilliar RM, Sagals G, Meguid SA, Oyonarte R, Deporter DA. Threaded versus porous-surfaced implants as anchorage units for orthodontic treatment: Three-dimensional finite element analysis of peri-implant bone tissue stresses. Int J Oral Maxillofac Implants 2006;21:879–889.
20. Weiner S, Simon J, Ehrenberg DS, Zweig B, Ricci JL. Advanced surface microtexturing techniques to enhance bone and soft tissue response to dental implants. Implant Dent. Forthcoming.