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Implant Design Considerations for the Posterior Regions of the Mouth

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The posterior regions of the mouth sustain greater forces, yet often present poorer bone density. A biomechanical approach, often presented to decrease risk factors in such regions, is to increase implant surface area. Most manufacturers provide implants in various lengths. The longest implants are typically inserted into the anterior regions of the mouth, where forces of less magnitude and superior bone quality are present. A finite element analysis supports the hypothesis that implant length is a secondary parameter for stress distribution. A common approach is to enhance implant surface area in the posterior regions primarily by focusing on diameter. However, this increases surface area by only 30% for conventional thread designs despite the fact that forces increase by >300% in the posterior regions. A change in implant diameter and thread design may increase surface area by >300%. Such increases in surface area may decrease stresses to the crestal bone regions and reduce both crestal bone loss and early loading implant failure. (Implant Dent 1999;8:376–386)

Key Words: design, crestal bone, early implant failure, functional surface area, bone density-quality

The posterior regions of the mouth present many unique conditions in implant dentistry. Bite forces are >300% greater in the posterior compared with the anterior regions.[1] The volume of bone is reduced in height as a consequence of the maxillary sinus or mandibular canal. Bone strength is directly related to its density, which is often poorer in the posterior regions.[2,3] The maxillary bone is often half as strong compared with the denser bone of the anterior mandibular regions.[4] A traditional method to decrease the risks of greater bite force and/or weaker bone has been to increase the implant surface area, which in turn decreases stress.[5] The most commonly used methods to increase implant surface area are by increasing the length and/or diameter of the implant body.

An implant treatment plan axiom has been to use the longest implant possible in the available bone.[6] The surface area of a root-form implant increases relative to its length. Almost every implant manufacturer provides implants in lengths ranging from 7 to 16 mm, often in 2 or 3 mm increments. For every 3 mm increase in length, the surface area of a cylinder-shaped implant increases an average of 20% to 30%[7] (Fig. 1). The use of the longest possible implant is most commonly recommended to maximize implant surface area. This length also allows engagement of the opposing cortical plate, a dense region that provides implant immobilization during trabecular bone interface remodeling. The anterior mandibular region serves as an ideal example of this approach.

Functional Versus Total Surface Area

For a given type of bone and implant volume, implant surface area should be optimized for the regions that sustain greater stresses during function. Thus, an important distinction is made between total surface area and functional surface area. Functional surface area is defined as the area that actively serves to dissipate compressive and tensile nonshear loads through the implant-to-bone interface during loading.[8] In contrast, total surface area may include a "passive" area that does not participate in load transfer. For example, the microscopic titanium balls from plasma spray coatings are often reported to provide increases up to 600% in total surface area. However, most of the surface geometry is too small to allow cell ingrowth. As a result, the amount of area that is actually exposed to the bone for functional loading may be <30% of the total surface area of the implant body.[9]

Interfacial loads to an implant should be compressive in nature because of bone's ability to best resist compressive loads, with a decrease in strength of 30% when subjected to tensile loading, and a decrease in strength of 65% in shear loading[10] (Fig. 2). Functional (or active) surface area does not include the portion of the implant that is passive or that transfers shear loads to bone. For example, if large-size balls are attached to the surface of an implant, only the bottom 1/3 of the sphere

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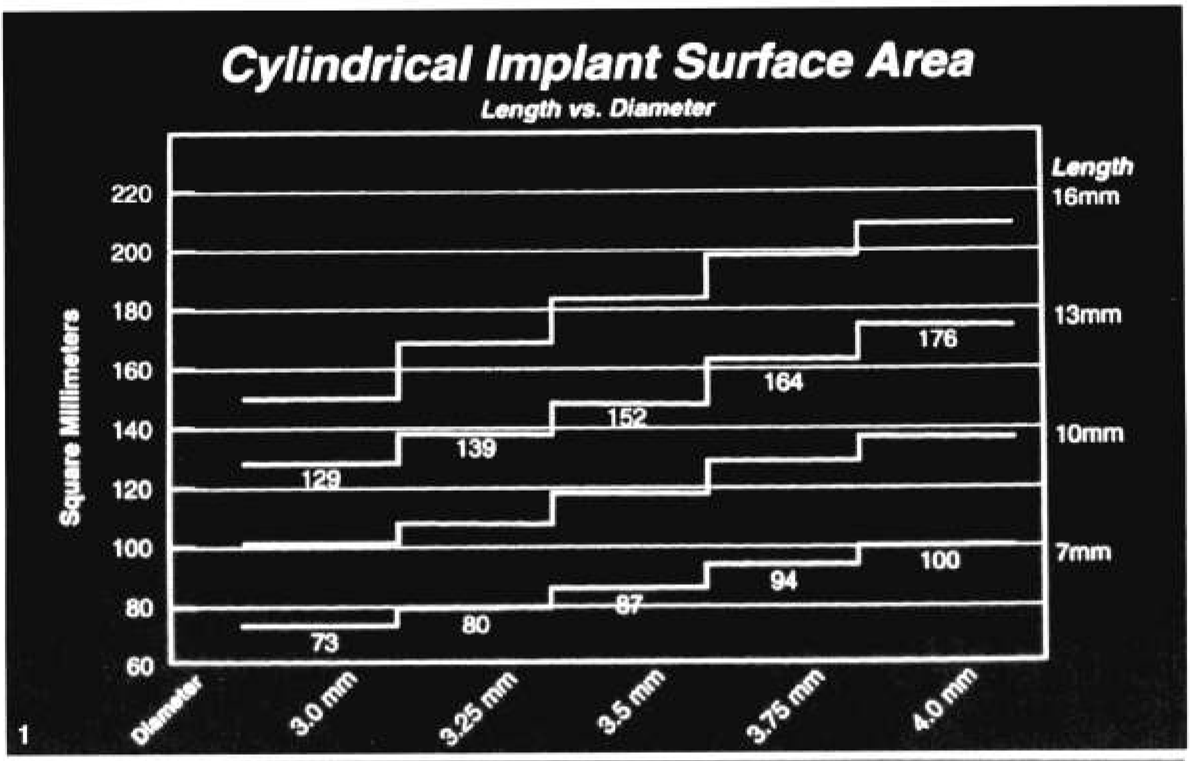

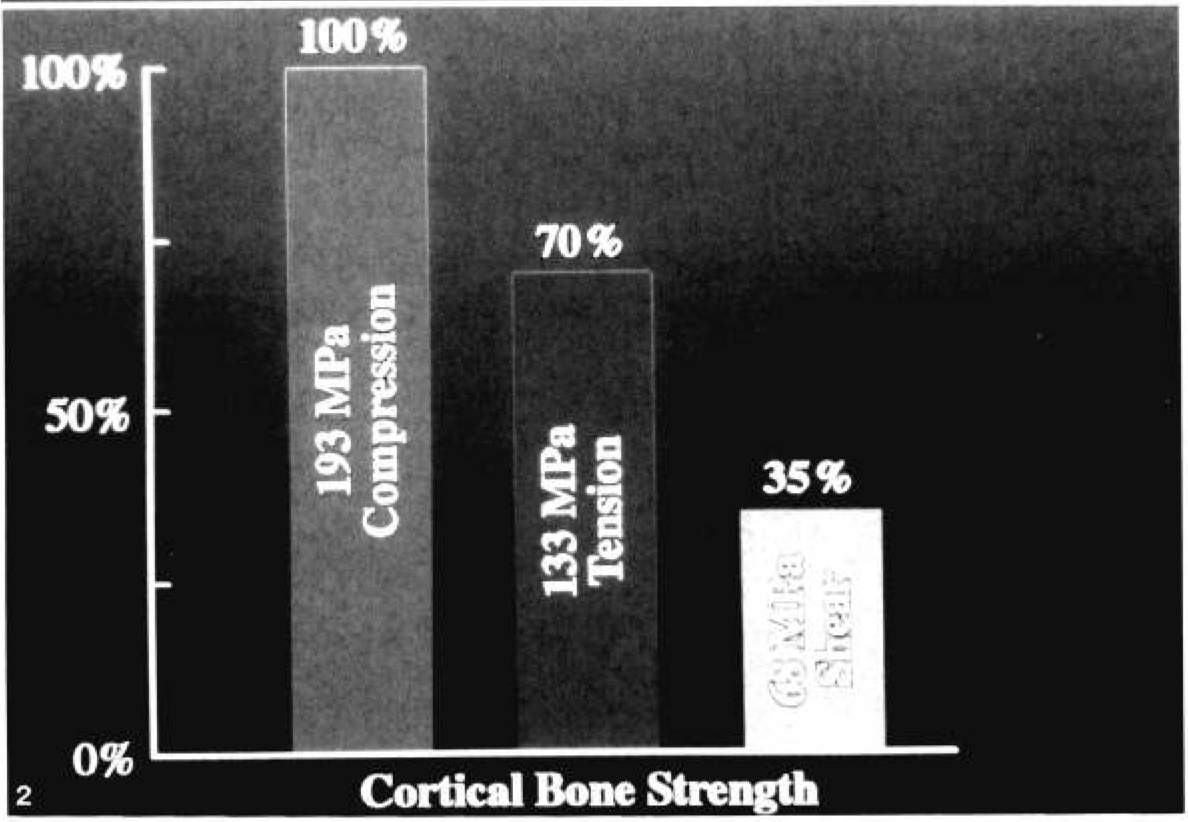

Fig. 1. The total surface area of a cylinder implant increases by 20% or more when a greater implant length is used for a similar diameter.
Fig. 2. Bone is strongest in compression, 30% weaker with tensile forces, and 65% weaker under shear load. Implants should be designed to load the bone under compression or tension, and attempt to limit shear forces.

can load the bone under an axial occlusal load. The top 1/2 to 2/3 of the sphere does not actively load the bone, but instead transfers passive or shear loads. Likewise, the functional surface area of a thread is that portion of the thread that participates in compressive or tensile load transmission under axial occlusal loads (Fig. 3).

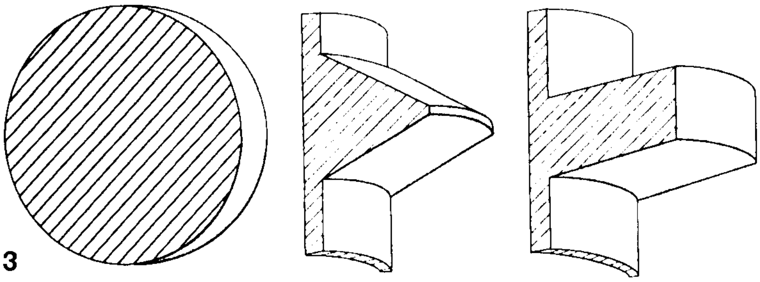

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Fig. 3. The surface area of the top of a sphere or thread does not participate in the functional load when an occlusal force is directed along the implant body long axis.
Fig. 4. Photoelastic study. Effect of torque on an implant placed in plastic with a modulus of elasticity similar to bone. The stress contours are greatest at the crest and minimal at the apex.

Rationale for Implant Length

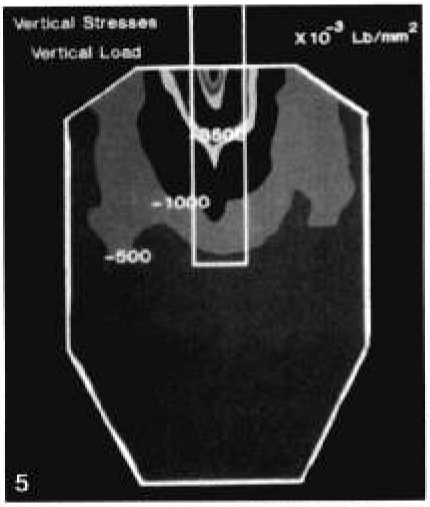
Implant length does affect the overall surface area through geometry and is, therefore, theoretically desirable. However, the functional surface area is more important than the overall surface area. A basic mechanical principle states that when two materials of different moduli are placed in contact with no intervening material and one is loaded, a stress concentration can be observed where the two materials first come into contact. These stress contours form a "V" or "U" shaped pattern, with greater magnitude near the point of first contact.[11] Most endosteal dental implants are fabricated from alloyed or pure titanium with a modulus of elasticity (stiffness) approximately five times greater than dense cortical bone.[12] Therefore, when an implant is in bone, the biomechanical mismatch results with an increased force where the implant body first contacts the bone. This contact area corresponds to the crest of the ridge. For an implant with a direct bone contact, the greatest magnitude of stress is concentrated in the crestal 5 mm of the bone implant interface. Therefore, unlike what occurs for a natural tooth surrounded by a periodontal membrane, stresses around implants during function and parafunction are typically concentrated in greatest magnitudes at the crest of the ridge, and stresses distributed to the apical third of an implant are minimal. The following study results substantiate this observation.

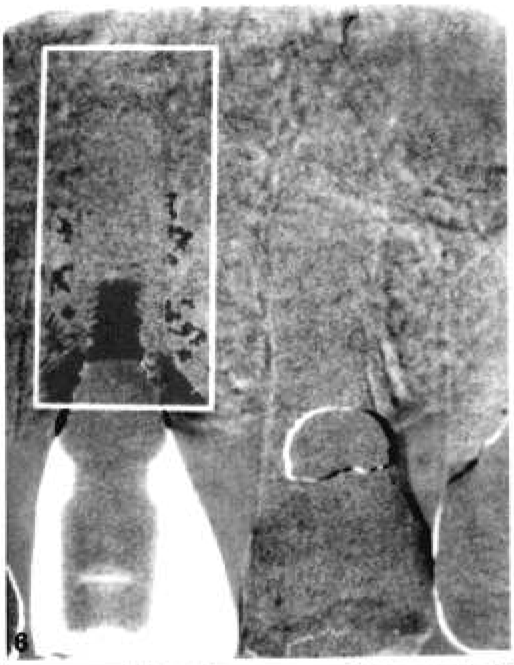
Photoelastic and Computer Generated Models

The phenomenon of higher crestal stresses next to an implant are confirmed in photoelastic and two- or three-dimensional finite element analysis studies when an implant is placed within a bone simulant and loaded[13,14] (Figs. 4 and 5). V-shaped patterns of stress can be observed in these models. Although the stresses are greatest at the crest compared with other regions of the implant body, the depth and geometry of the V-shaped pattern of bone loss around the implant varies in different bone densities.

Histomorphometric Studies and Digital Radiographs

Bone responds to greater crestal stresses by one of two extremes, either by increased density or resorption, based upon the amount of strain within the bone.[15] Digital subtraction radiography revealed bone density increases in the crestal third of a successful implant loaded within physiological limits[16] (Fig. 6). Histologic evaluation of loaded and nonloaded implants in monkeys demonstrated a greater bone increase in the crestal third of the loaded implant[17] (Figs. 7 and 8). An increase in bone density in the crestal third has also





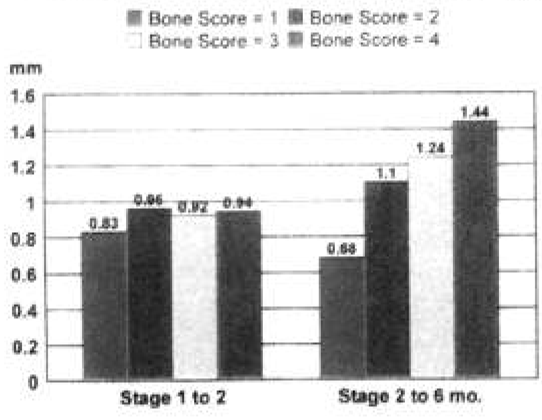


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Fig. 5. Three-dimensional finite element analysis of an implant in bone. A V-shaped pattern of stresses occurs at the crest, with minimal stresses at the apex. (From Misch[14])
Fig. 6. Digital subtraction radiograph demonstrates an increase in bone density in the crestal third of the implant (green) and illustrates the crestal bone loss (red). (From Appleton et al[16])
Fig. 7. A one-stage implant before loading exhibits equal bone distribution around it. (From Piatelli et al[17])
Fig. 8. A loaded implant exhibits an increase in bone density at the crestal third as a response to the increase in bone stresses within physiologic limits. (From Piatelli et al[17])
Fig. 9. The amount of early crestal bone loss seems related to the density of bone. The bone loss from surgery to Stage II is similar for all bone densities. During the first 6 months of loading, the bone loss is directly related to the density, the least bone loss occurs in D1 and the most in D4. (From Manz[23])

been reported in dogs when abutments were torqued into position four separate times over a period of several weeks.[18]

Early Crestal Bone Loss

Early crestal bone loss after loading, such as the "saucerization" of one-piece blade implants has been observed around the permucosal aspect of dental implants for decades.[19] Marginal bone loss during the first year of function after prosthesis placement was first quantified and reported by Adell et al.[20] This study observed the greater magnitude of bone loss during the first year of prosthesis loading, with an average bone loss of 3 mm from the original crest of bone during this timeframe (range, 0–3 mm from the first thread). The initial transosteal bone loss around an implant during the first year of loading formed a V- or U-shaped pattern. Bone loss occurs when the stresses are beyond the physiologic limit. The pathologic microstrain level of bone that results in resorption represents only 20% to 40% of the value required to cause fracture.[15] Such stresses cause resorption and bone loss at the crestal region. The stresses found at the crest when beyond physiologic limits of the bone may cause microfracture of bone. Such stress concentrations may impede blood supply in the region.[21,22] This localized decrease in blood may contribute to bone loss, impair remodeling, and make the environment more susceptible to the deleterious effects of anaerobic bacteria.

Clinical Observation

The amount of early crestal bone loss seems to be directly related to its density; the densest bone loses the least, and the softer bone types lose the most[23] (Fig. 9). A variable range of bone loss related to density allows us to refute the hypotheses of periosteal reflection, osteotomy preparation, and the biologic width as possible causes of initial bone loss after loading.[24] In weaker bone types, crestal bone loss may involve a large portion of the implant body and lead to implant failure. Early loading implant failure has been reported in good quality bone, with reports ranging from 2% to 6%.[25-29] However, when only the softest bone is evaluated, these early loading failures may range from 28% to 50%.[30,31] The modulus of elasticity is not a constant value for all bone densities. Denser bone exhibits higher values, and fine trabecular bone has the lowest value.[4] The decreased bone strength, increase in elastic modulus mismatch between implant and bone as density decreases, the decrease in implant-bone contact as bone density decreases,[32] and the stress contour differences in implant bone interfaces as bone density decreases increase the risk of early loading failure. As a consequence, the cause of early crestal bone loss is also often the causative element in total implant failure,[5] and also accounts for the greater bone loss and implant failure in the maxilla compared with the mandible.[20] When no early crestal bone loss after loading is observed, early implant failure after initial rigid fixation at Stage II uncovery is seldom observed. The management of stress is essential to reduce the incidence of implant failure and crestal bone loss.

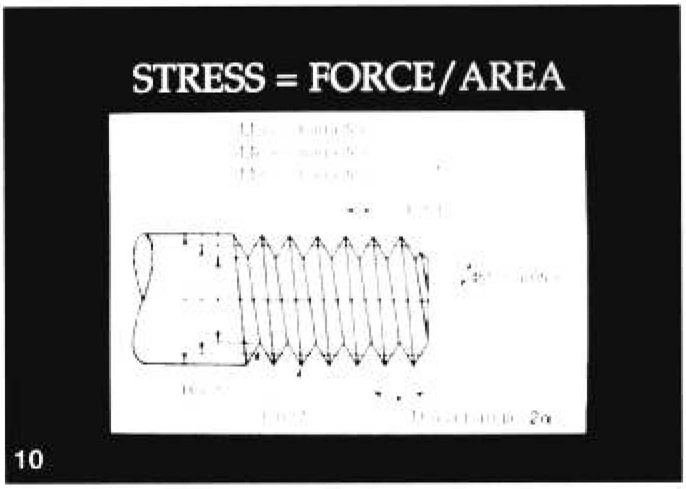
Most stresses to the implant bone interface are located in the crestal 1/3 of the implant. This is a critical area for stress distribution. Longer implants, although of greater total surface area, may transfer very little stress to the apical region and do not minimize stress in the most critical crestal regions. Hence, the greatest functional surface area is required in the crestal 5 mm of the implant body. In the posterior regions, longer implants often cannot be used without surgical manipulation such as nerve repositioning in the mandible or sinus grafts in the maxilla. In addition, the posterior regions do not have a dense cortical plate as the opposing landmark (sinus floor and inferior alveolar canal). Hence, the rationale and prescription to engage the opposing cortical plate with the apex of a root-form implant usually cannot be followed in these regions.

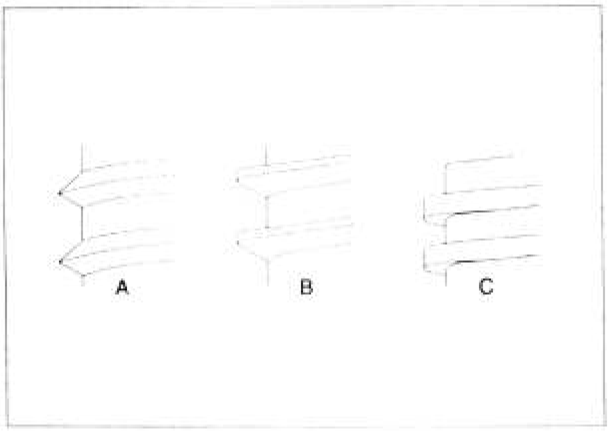
Implant Body Surface Area

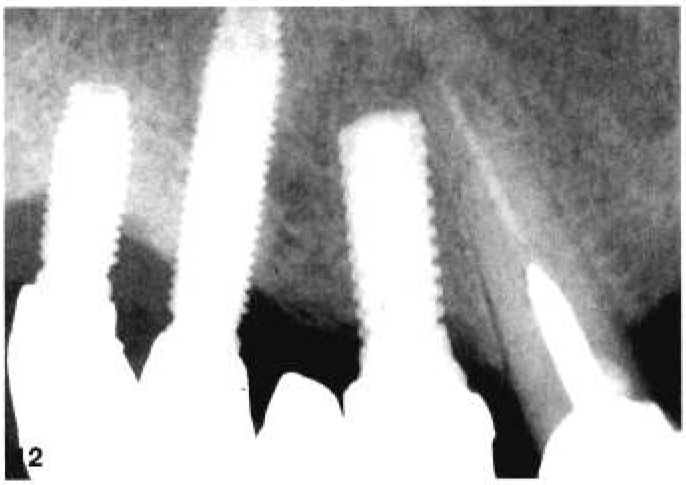
Thread Geometry

Biomechanical engineering can help improve the transosteal region, which can subsequently decrease the length of implants and minimize crestal stress around endosteal implants. One approach is to increase surface area at the crestal region, where the greatest stress conditions exist.[33] Functional surface area per unit length of the implant may be modified by varying three thread geometry parameters: thread pitch, thread shape, and thread depth. Thread pitch is defined as the distance between adjacent threads, or the number of threads per unit length in the same axial plane and on the same side of the axis[34] (Fig. 10). The smaller (or finer) the pitch, the more threads on the implant body for a given unit length and, thus, the greater surface area per unit length of the implant. Restated, a decrease in the distance between threads will increase the number of threads per unit length. For example, the thread pitch of the Steri-Oss implant body (Steri-Oss, Yorba Linda, CA) is 0.625 mm, and the thread pitch of the Nobel BioCare implant body (Nobel BioCare USA, Westmont, IL) is 0.60 mm. The latter exhibits a greater number of threads per unit length.[33] Therefore, if force magnitude is increased or bone density is decreased, the thread pitch may be decreased to increase the functional surface area. Traditionally, manufacturers have provided implant systems with a constant pitch and surface area per unit length regardless of the nature of forces applied to or the bone density of the recipient site.

The thread shape is another characteristic by which functional surface area can be altered. Thread shapes presently represented in dental implant designs include the following: V-shape, reverse buttress, and square (Fig. 11). In conventional engineering applications, the V-thread design (Nobel BioCare, [Westmont, IL], Paragon [Paragon Implant Co., Encino, CA] Lifecore [Lifecore, Chaska, MN], 3i [3i, West Palm Beach, FL]) is called a "fixture" and is primarily used for the fixation of







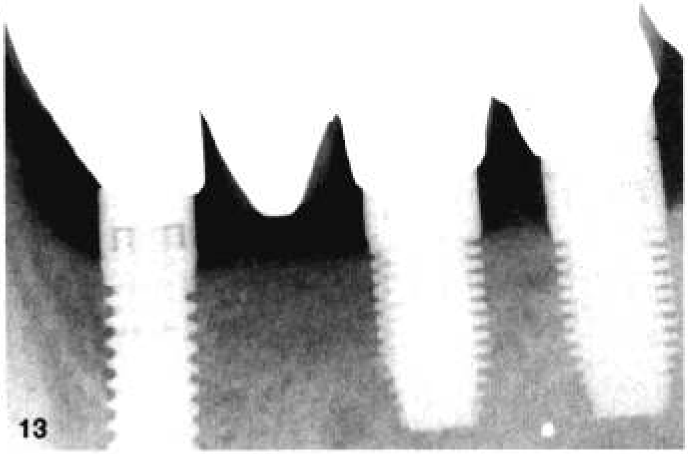


Fig. 10. The thread pitch describes the distance between the threads. The smaller the pitch, the greater the number of threads and the greater the surface area.
Fig. 11. Thread shape may be a V-shape (Noble BioCare, Paragon, LifeCore), reverse buttress (Steri-Oss), or square(BioHorizons).
Fig. 12. A V-shaped thread (the two implants on the left in the molar position) transmits a 10 times greater shear force than a square thread design (implant on the right in the first premolar position). Bone is weaker under shear loads. Forces are greater in the posterior regions. An implant with higher crestal stress may influence the amount of bone loss (18-month postoperative periapical x-ray).
Fig. 13. The anterior implant (far left) is 4 mm in diameter, whereas the posterior implants (center and right) are 5 mm in diameter (BioHorizons). The posterior implants benefit from a 300% greater surface area compared with conventional designs because of combined thread pitch and diameter increase (2-year postoperative periapical x-ray).

Table 1. Comparison of Functional Thread Surface Area (Without Coatings)*

	BioHorizons	Nobel BioCare and Similar V-Shape Designs	Steri-Oss
Type	(a) D3 (b) D4	Standard fixture	Threaded implant
Diameter (mm)	4	3.75	3.8
Thread surface Area (mm^2)	(a) 210 (b) 245	127	111
Type	(a) D3 (b) D4	Wide platform	Threaded implant
Diameter (mm)	5	5.5	5
Thread surface Area (mm^2)	(a) 419 (b) 468	183	134

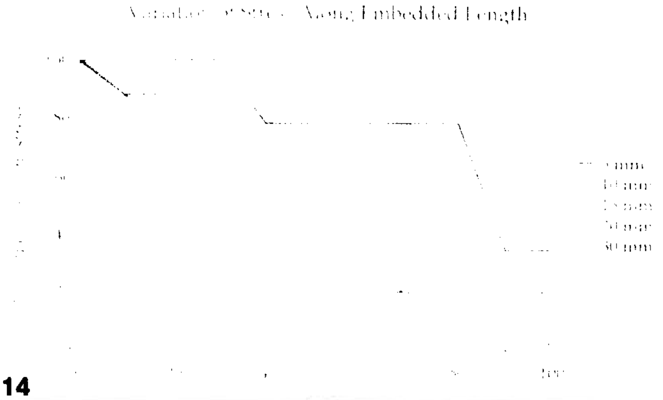
* The functional surface area of three 10-mm commercial thread designs is affected by design (does not include coatings or the crest module). The greater diameter implants increase surface area by increase in circumference, except BioHorizons, which also increases thread depth.

metal parts together because the 30 degree incline of the V-thread design causes the male component of the screw to stretch or elongate during a preload, which decreases the incidence of screw loosening.[33] The reverse buttress thread shape is flat on the top (Steri-Oss), and is optimized for pullout loads. This thread design originated from a German engineer, Krupp, and was used to prevent screws from pulling out of concrete bunkers used to hold artillery cannons.[34] The square or power thread (BioHorizons Implant System, Birmingham, AL) provides an optimized surface area for intrusive, compressive load transmission.[11] For example, over the last decade, nearly every car jack has used a square thread design to support the weight of the car when it is lifted off the ground. Shear loading is the most detrimental loading profile for bone.[10] The shear force transmitted by a V-thread face is ~10 times greater than the shear force transmitted by a square thread.[33] The shear component per unit length of a reverse buttress thread design is similar to a V-thread when subjected to an occlusal load. A reduction in shear loading at the thread-to-bone interface is particularly important in compromised bone strengths or under high occlusal loads, which are often seen in the posterior regions of the mouth (Fig. 12).

The thread depth refers to the distance between the major and minor diameters of the thread.[34] The greater the thread depth, the greater the surface area. The Steri-Oss thread depth is 0.28 mm, the Nobel BioCare thread depth is 0.375 mm, and the BioHorizon thread depth in the crestal region is 0.419 mm; therefore, each has a different surface area.[33] Conventional implants provide a uniform thread depth throughout the length of the implant. The thread depth may be varied over the length of the implant to provide increased functional surface area in the regions of highest stress (eg, the crestal region of alveolar bone).

The thread depth for conventional implants is similar for all implant diameters. As a result, an increase in surface area with greater diameter implants is primarily the result of an increased circumference, which is ~30%.[7] However, thread depth can be maximized in larger-diameter implants. Hence, the surface area in the larger-diameter implant body may result from a combined increase in circumference and thread depth, which could be increased up to 300% as a consequence[35] (Table 1). Thread geometry may modify surface area to such an extent that smaller-length implants may have greater surface area than implants of wider and/or longer dimensions, but of a different design. These factors are especially critical in the posterior regions of the mouth because the opposing anatomical and functional boundaries often limit the length of the implant (Fig. 13).

Teeth with longer roots represent a better biomechanical system to resist lateral loads. Hence, it has been suggested that longer implants be used to provide greater stability un-



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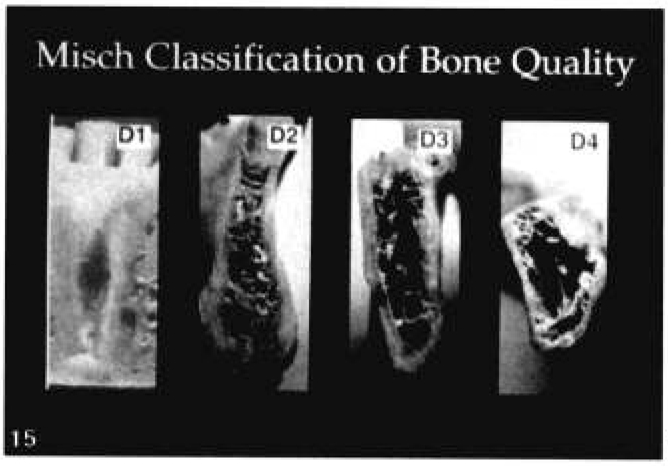

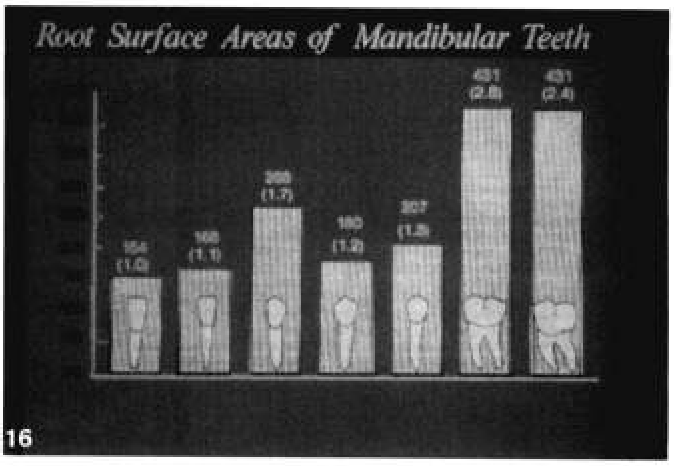

Fig. 14. The amount of stress by percent is plotted on the *y* axis and implant length is plotted on the *x* axis (also by percent). Regardless of length, the greatest stresses occur in the crestal third of the implant (finite element analysis).

Fig. 15. Bone density affects the amount of implant-bone contact before early loading. D1 bone benefits from the greatest contact whereas D4 bone has the least.

Fig. 16. The anterior teeth have three times less surface area than posterior teeth. Natural teeth do not increase in length in the posterior regions, but exhibit an increased diameter and different support design (increased number of roots).

der similar loading conditions. Finite element analysis provides an analytical means to investigate the influence of implant length relative to functional surface area under extreme lateral loading conditions. Idealized cylindrical implants of different lengths were placed in a computerized bone simulant model of good quality bone and embedded to lengths of 5, 10, 15, 20, and 30 mm.[8] For each implant sample, a 10-mm long crown was placed over the implant. Under these loading conditions, the percentage of maximum stress was plotted against the percentage of embedded length (Fig. 14). Stresses were given values from 0% to 100% (100% being the greatest).

Regardless of implant length, of stresses with values ranging from 70% to 100%, 70% were all concentrated in the crestal third of the implant. Peak stresses were not completely dissipated in the 5-mm long implant model, with approximately 30% of the maximum stress still present at the apex of the implant. The 5-mm implant model, therefore, does not provide adequate length for lateral force dissipation. In the 10-mm implant model, 80% of the stresses were dissipated in approximately 95% of the embedded length. For the 15- and 20-mm lengths, 80% of the stresses were dissipated in approximately 90% of the embedded length. For the 30-mm length, 80% of the stresses were dissipated in approximately 70% of the embedded length.

The results of this analysis point to the fact that the greatest stresses generated under lateral load are concentrated within 5 mm of the crest regardless of the implant length. Therefore, crestal stress contours are comparable in the 10-mm and 30-mm implant models. This study suggests that implants longer than 10 mm may not be necessary in good quality bone. In addition, the greatest need for functional surface area is in the crestal third of the implant.

The results of this computer bone study corroborate the animal and human studies, which exhibit bone density increases after loading. Histologic and radiographic results demonstrate an increase in bone density during the first 6 months of loading in the crestal third of the implant.[16-18] Stresses that reach the apical region seem to be of insufficient intensity to stimulate an increase in bone density in these regions (Figs. 6, 7, and 8). Physiological stresses can cause bone density to increase, which decreases the risk of crestal bone loss or early implant failures. This may be the reason for the decreased rate and occurrence of crestal bone loss and early loading failure after the first year of loading.

Implant Length Relative to Bone Density

Functional surface area also plays a critical role when addressing implant-to-bone contact in relation to bone density. D1 bone, the densest and strongest bone found in the jaws, also provides the most intimate contact with a threaded, root-form implant at initial implant loading (~80%). D2, D3, and D4 bone exhibit progressively decreasing percentages of bone at the implant interface, with D4 bone exhibiting ~25% of implant-bone contact after initial healing and uncovery of a titanium implant[32] (Fig. 15). Mechanical stress is equivalent to the applied load divided by the surface area over which the load is dissipated. D4 bone exhibits the weakest biomechanical strength and the lowest functional contact area to dissipate loads at the implant-borne interface. Thus an improved functional surface area per unit length of the implant (in contrast to total surface area) is indicated in this soft bone to reduce stresses. The modification of thread design parameters may cause such a significant increase in surface area that implants of shorter length and similar diameter may exhibit a greater functional surface area than conventional implant designs of greater length. The functional surface area requirements increase from a minimum in an implant designed for

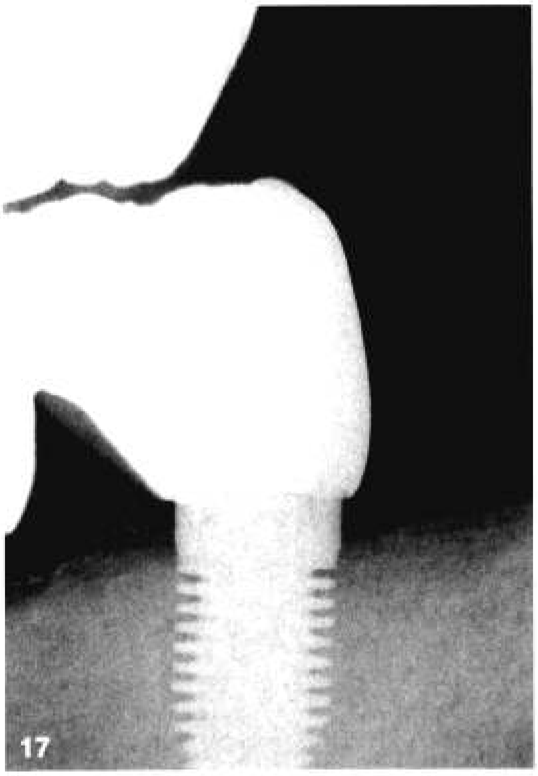

Fig. 17. A posterior implant, especially in soft bone, should have greater surface area to minimize crestal bone loss or early loading failure. The larger diameter, greater thread number, and thread depth may result in a functional surface area increase of >300% compared with the conventional V-shaped 3.75-mm diameter thread design.

D1 bone to a maximum in the D4 bone implants.

Conclusion

In the past, force reduction and surface area were difficult to balance in the posterior regions of the mouth. Studies clearly demonstrate forces are often 300% greater in the posterior compared with the anterior regions of the mouth.[1] Bone densities and strengths are 50% to 200% weaker in the posterior region. Yet, the implants with greater surface area (according to length) were inserted in the anterior regions. Natural teeth do not have longer roots in the posterior regions of the mouth, where stresses are greater. Instead, increased surface area is achieved by an increase in diameter and a change in root design (Fig. 16).

Methods to increase the functional surface area, notably at the crestal 5 to 8 mm (longer for less dense bone) are warranted, with emphasis in the posterior regions and areas of greater force. Conventional implant designs increase surface area by 20% to 30% when their diameters are increased by 2 mm. This may not be adequate to compensate for a force increase of more than 300% in the posterior regions. When thread depth is increased along with the diameter, the functional surface area may increase more than 300% (Fig. 17).

Implant dentistry has developed a greater understanding of biomechanics and the importance of stress reduction to minimize the risks of crestal bone loss and early implant failure. However, conventional root-form implants are limited to changes in implant length and diameter. These changes are much less effective surface area enhancers than thread designs. A modified implant design with increased functional surface area (instead of total surface area) allows shorter implants with greater surface areas to be used in all regions of the mouth. This variation is most important in the posterior regions where forces are greater, bone strengths are inferior, and available bone height is often less than in the anterior regions.

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Abstract Translations [German, Spanish, Portuguese, Japanese]

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ZUSAMMENFASSUNG: Die hinteren Mundraumbereiche sind gröβeren Kräften ausgesetzt, weisen dabei aber oft eine geringere Knochendichte auf. Ein zur Verringerung der damit verbundenen Risikofaktoren häufig vorgebrachter biomechanischer Ansatz besteht in einer Vergröβerung der Implantatoberfläche. Die meisten Hersteller bieten Implantate unterschiedlicher Länge an. Die längsten Implantate werden normalerweise in die vorderen Mundbereiche eingebracht, wo geringere Kräfte auftreten und die Knochenqualität gröβer ist. Eine Untersuchung mit Hilfe der finite Elemente-Methode stützt die Hypothese, daβ die Länge des Implantats nur einen sekundären Parameter für die Lastverteilung darstellt. Für gewöhnlich konzentrieren sich die Anstrengungen bei der Vergröβerung der Implantatoberfläche in den hinteren Bereichen hauptsächlich auf den Durchmesser. Damit werden jedoch nur Oberflächenvergröβerungen von etwa 30% für konventionelle Fadendesigns erzielt, was der Tatsache gegenübergestellt werden muβ, daβ die im hinteren Mundraum auftretenden Kräfte um über 300% zunehmen. Eine Änderung des Implantatdurchmessers und des Fadendesigns (*durch Implantatsysteme von BioHorizons®*) kann die Oberfläche um über 300% vergröβern. Eine solche Zunahme der Oberfläche kann die Belastung des Kammknochenbereichs verringern und sowohl den Verlust an Kammknochen sowie die Gefahr des Implantatverlustes durch frühe Belastung verringern.

SCHLÜSSELWÖRTER: Implantatdesign, endostale Implantate, Kammknochenverlust, früher Implantatverlust, funktionale Oberfläche, Knochendichte, Knochenqualität

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ABSTRACTO: Las regiones posteriores de la boca sostienen fuerzas más fuertes, pero a menudo presentan una densidad del hueso más pobre. Un método biomecánico, a menudo presentado para reducir los factores de riesgo, es aumentar el área de la superficie del implante. La mayoría de los fabricantes entregan implantes en varias longitudes. Los implantes más largos son típicamente insertados en las regiones anteriores de la boca, donde se encuentran fuerzas de menor magnitud y calidad superior del hueso. Un análisis de elementos finitos apoya la hipótesis que la longitud del implante es un parámetro secundario para la distribución de la tensión. Un método común es mejorar el área de la superficie del implante en las regiones posteriores principalmente al concentrarse en el diámetro. Sin embargo, esto aumenta el área de la superficie de solamente un 30% para diseños de roscas convencionales a pesar de que las fuerzas aumentan más de un 300% en las regiones posteriores. Un cambio en el diámetro del implante y diseño de la rosca podría aumentar el área de la superficie más de un 300%. Dichos aumentos en el área de la superficie pueden reducir las tensiones en las regiones crestales del hueso y reducir la pérdida del hueso crestal y la falla inicial de carga del implante.

PALABRAS CLAVES: implantes endosteales, pérdida de hueso crestal, falla inicial del implante, área de la superficie funcional, densidad – calidad del hueso.

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SINOPSE: As regiões posteriores da boca suportam forças maiores, e no entanto freqüentemente apresentam densidade óssea debilitada. Uma abordagem biomecânica, freqüentemente apresentada para diminuir os fatores de risco em tais casos, é aumentar a área da superfície de implante. A maioria dos fabricantes fornecem implantes em vários comprimentos. Os implantes mais compridos são inseridos geralmente nas regiões anteriores da boca, onde estão presentes forças de menor magnitude e qualidade óssea superior. Uma análise de elemento finito apóia a hipótese de que o comprimento do implante é um parâmetro secundário para a distribuição do estresse. Uma abordagem comum é a de ampliar a área da superfície de implante nas regiões posteriores principalmente através do enfoque no diâmetro. Contudo, isto aumenta a área da superfície somente 30% para desenhos de rosca convencionais, apesar do fato de que as forças aumentam mais de 300% nas regiões posteriores. Uma mudança no diâmetro de implante e no desenho da rosca *(isto é, sistemas de implante BioHorizons®)* pode aumentar a área da superfície mais de 300%. Tais aumentos na área da superfície podem diminuir os estresses para as regiões ósseas das cristas e reduzir tanto a perda óssea da crista como a falha no carregamento antecipado do implante.

PALAVRAS-CHAVES: desenho de implante, implantes endosteais, perda óssea da crista, falha antecipada de implante, área de superfície funcional, densidade óssea - qualidade

口腔奥部インプラントデザインの考察

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概要：
口腔奥部には大きな力がかるにもかかわらず、その骨密度は低いことが多い。生物機械的方法でこのようなリスクを低減するためには、インプラント表面積を大きく取ることが考えられる。インプラント製造元のほとんどが、各種の長さのインプラントを提供しているが、一番長い種類のインプラントは普通、かかる力が小さく骨質的に優れた口腔前部に使われることが多い。有限要素分析法を使うと、インプラントの長さは応力分布に関して第2に重要な要因であるとの仮説を立てることができる。よい強度を得るために普通に使われる方法に、インプラントの直径を大きく取り口腔奥部のインプラント表面積を拡げるものがあるが、口腔奥部では応力が300％以上の大きさに達するにもかかわらず、この方法で従来のスレッド方式インプラントを使った場合に表面積は30％程度拡げるのがせいぜいである。これに対しBioHorizons(R)インプラントシステムのように直径とスレッドデザインを共に変ええるものでは、表面積を300％以上拡大することが可能となる。表面積をこのように拡げた場合、crestal bone部への応力は低下しcrestal boneの損失とインプラント設置後の初期的失敗件数を減少させることができる。

キーワード：
インプラントデザイン、エンドスティールインプラント、crestal boneの損失、
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