APPLICATION OF FINITE ELEMENT ANALYSIS TO EVALUATE IMPLANT FRACTURES

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Statement of problem. Higher fracture rates were reported for Branemark implants placed in the maxilla and for 3.75 mm diameter implants installed in the posterior region.

Purpose. The purpose of this study was to investigate the fracture of a fixture by finite element analysis and to compare different diameter of fixtures according to the level of alveolar bone resorption.

Material and Methods. The single implant and prosthesis was modeled in accordance with the geometric designs for the 3i implant systems. Models were processed by the software programs HyperMesh and ANSA. Three-dimensional finite element models were developed for; (1) a regular titanium implant 3.75 mm in diameter and 13 mm in length (2) a regular titanium implant 4.0 mm in diameter and 13 mm in length (3) a wide titanium implant 5.0 mm in diameter and 13 mm in length each with a cementation type abutment and titanium alloy screw. The abutment screws were subjected to a tightening torque of 30 Ncm. The amount of preload was hypothesized as 650 N, and round and flat type prostheses were 12 mm in diameter, 9 mm in height were loaded to 600 N. Four loading offset points (0, 2, 4, and 6 mm from the center of the implants) were evaluated. To evaluate fixture fracture by alveolar bone resorption, we investigated the stress distribution of the fixtures according to different alveolar bone loss levels (0, 1.5, 3.5, and 5.0 mm of alveolar bone loss). Using these 12 models (four degrees of bone loss and three implant diameters), the effects of loading offset, the effect of alveolar bone resorption and the size of fixtures were evaluated. The PAM-CRASH 2G simulation software was used for analysis of stress. The PAM-VIEW and HyperView programs were used for post processing.

Results. The results from our experiment are as follows:

1. Preload maintains implant-abutment joint stability within a limited offset point against occlusal force.
2. Von Mises stress of the implant, abutment screw, abutment, and bone was decreased with increasing of the implant diameter.
3. With severe advancing of alveolar bone resorption, fracture of the 3.75 and the 4.0 mm diameter implant was possible.
4. With increasing of bending stress by loading offset, fracture of the abutment screw was possible.

Key Words

Finite element analysis, Preload, Abutment screw fracture, Implant fracture, Alveolar bone resorption

To maintain the success of an implant, biomechanical problems should be prevented such as screw loosening, screw fracture, prosthesis fracture, and implant fracture. Implant fracture is an uncommon but significant complication that occurred in 96 of 6560 implants (1.5%) followed for 3 to 15 years in nine studies.[1] Implant fracture is a mechanical failure that seriously compromises the longevity of the treatment. Higher fracture rates were reported for Branemark implants placed in the maxilla and for 3.75 mm diameter implants installed in the posterior region.[2,3] Most fractures occurred between the third and fourth implant threads, which correspond to the last thread of the abutment screw.[4,5] Implant fractures may be the result of implant design and manufacturing defects, nonpassive fit of the prosthetic framework, or physiologic or biomechanical overload.[6,7] A higher incidence of implant fractures has been reported in fixed partial dentures supported by only two implants.[3,7-14]

Bending overload was probably created by a combination of effects from parafunctional forces, bone resorption, posterior location of the implants, and implant diameter. All observed fractures occurred with commercially pure titanium 3.75 mm diameter threaded implants. Prosthetic or abutment screw loosening preceded implant fracture in the majority of the implants.[2,3]

The SEM study showed the presence of fatigue striations, which constituted the crack front under cyclic loading. These striations were, according to Morgan et al., the pathognomonic mark of fractures resulting not from overload, where a dimpled surface related to plastic deformation is present, but from fatigue failure.[15-18]

A specific bone loss pattern has been described as a primary cause of implant fracture. An alternative view is that implant fracture involves progressive fatigue failure until the implant lacks adequate strength to maintain integrity, culminating in a catastrophic failure. During the progression of the fracture process, an infective process may be involved in the observed pattern of bone loss. In the former situation, bone loss is an etiologic factor for the fracture, while in the latter instance, it is the fracture that causes the bone loss. At this point, it is unclear which event precedes the other. If bone loss is a predisposing factor, then early intervention to reduce occlusal forces to the implant seem to be justified. Conversely, if the initial tearing of the implant could result in bone loss because of a secondary infection, then occlusal adjustment would be of no value, since the weakened implant is destined to fracture.[16]

Bone loss is thought to be the result from the magnitude and/or the direction of the load incorrectly oriented along the long axis of the implant.[19,20] Peri-implant bone resorption and implant fracture are said to be related to excessive bending moments.[21,22] Coronal bone resorption produces a higher bending stress on the implant because of the loss of supporting bone. In addition, this type of bone resorption usually extends to the level corresponding to the end of the abutment screw, and in this region the resistance to bending is diminished. An area of stress concentrated could be produced at the root of a thread, resulting in crack initiation and propagation. Metal fatigue seems to be the most common cause of structural failure. The cracks grow from the site of maximum stress and can produce a sudden failure.[16,23-25]

Occlusal forces (magnitude and location) are usually the major factors that directly affect the load transfer and stress distribution.[26] The mechanical improvement of implants and the application of wide diameter fixtures reduce fixture fracture. However, it is occasionally impossible to use wide diameter fixtures, and there is uncertainty about the relationship of critical loading offset point to the amount of alveolar bone loss. Overload can cause bone resorption or fatigue failure of the implant.

Sufficiently high screw joint preloads are required to maintain screw joint integrity and confer clinical longevity to implant prosthetic components in order to prevent such complications as screw loosening and screw fracture.[27,28] Griffith suggested that the optimal preload for a given screw is 75% of the force required to exceed its ultimate breaking strength. This is a totally arbitrary criterion, which in no way takes into consideration material properties or the possibility of the screw having to resist any further external loading.[29]

However, it is difficult to measure preload in any implant system, because of the complex nature of the implant design. A method for the direct measurement of preload in the oral environment has yet to be developed. The in vitro determination of preload has relied on strain gauges or force transducers positioned adjacent to the implant body to measure forces within the implant complex as the screw joint assembly is performed. Other studies have assumed a preload value and then applied this value to an implant system model to determine implant biomechanical performance using finite element analysis.[30]

The aim of this study was to perform three-dimensional finite element analysis to investigate the fracture of fixtures and to compare 3.75 mm, 4.0 mm, and 5.0 mm diameter fixtures according to the level of alveolar bone resorption.

MATERIALS AND METHODS

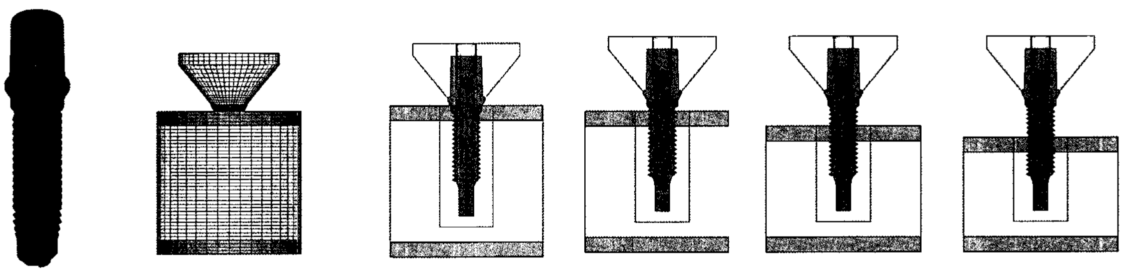
The single implant and prosthesis were modeled in accordance with the geometric designs for the 3i implant systems(3i Implant Innovations, Inc., Palm Beach Gardens, FL, U.S.A.). Models were processed by the software programs HyperMesh version 7.0(Altair Engineering, Inc., Troy, MI, U.S.A.) and ANSA(BETA CAE Systems, version 11.2.4). To obtain a detailed model, the threads of the implant and the abutment screw were represented with in their spiral characteristics. Three-dimensional finite element models were developed for: (1) a regular titanium implant 3.75 mm in diameter and 13 mm in length (2) a regular titanium implant 4.0 mm in diameter and 13 mm in length (3) a wide titanium implant 5.0 mm in diameter and 13 mm in length, each with a cementation type abutment and titanium alloy screw. In the wide implant, the abutment and prosthesis are constructed the same as in the regular implant for identification of loading conditions. The prostheses were designed as rigid bodies to avoid a deformative effect. The nodes and elements are shown in table I, and the values assumed for Poisson' s ratio and Young' s modulus are given in table II. All materials used in the models were considered to be isotropic, homogenous, and linearly elastic. The bone was modeled as a cancellous core surrounded by a 2.0 mm cortical layer. The mesial and distal section planes were not covered by cortical bone. Boundary fixation included restraints for all six degrees of freedom, including rotation and translation in three coordinate axes for correspondent nodes located at the most external mesial and distal planes and the bottom. The abutment screws were subjected to a tightening torque of 30 Ncm. In the finite element model, a nonlinear contact zone with friction was defined between the implant, abutment screw and abutment. According to the study by Martin et al.[31], titanium alloy abutment screws have a preload from 434.8 ± 310.6 N to 636.1 ± 336.6 N. Standardization of preload is difficult, so we employed Lisa' s experimental value for preload using ABAQUS(version 5.8; HKS Inc, Pawtucket, RI). When the friction coefficient is 0.12, the preload value of Unigrip and TorqueTite screws is approximately 650 N at 30 Ncm.[30] For real stress distribution in the preload, we rotated the abutment screw without contact conditions and then applied a contact condition for preload stress distribution. Maximum bite force for the cyclic fatigue fracture of implant was estimated at 600 N, which is 55% of the yield strength of titanium alloy abutment screw. The fatigue strength is roughly equivalent to 50 % of the tensile strength. Maximum bite force is generated when ipsilateral chewing of food causes mean maximum forces in centric occlusion and chewing and grinding.[32] For the removal of eccentric contact, the occlusal table was a flat type and was simplified for the evaluation of changes in loading offset alone. A bending moment was generated by altering the loading offset by 0, 2, 4 or 6 mm. The amount of preload was hypothesized as being approxima- tely 650 N and 12 mm diameter, 9 mm high round and flat type prosthesis were loaded to 600

Table I. The elements and nodes

Model Name		Crown		Abutment		Abutment screw		Implant		Cortical bone		Trabecular bone		TOTAL	
Implant	Cortical depth	node	element	node	element	node	element	node	element	node	element	node	element	node	element
3.75mm diameter implant	0	6289	5472	5472	4128	11384	56361	21463	99026	7125	14711	34192	130154	77303	309852
	1.5							21321	98113	7440	16202	31582	120191	75292	234506
	3.5							21481	99048	7504	16568	27570	104000	72169	219616
	5.0							21450	98869	7612	16264	24570	91806	69586	206939
4.0mm diameter implant	0							22964	107715	7083	14484	29091	100383	73660	222582
	1.5							22836	106989	7238	15020	26921	93079	71935	215088
	3.5							22915	107369	7242	15014	24106	83784	69873	206167
	5.0							22970	107680	7248	15081	22142	77606	68470	200367
5.0mm diameter implant	0	7968	6912	7776	6336	11384	56361	27817	136313	6419	10902	25644	81124	74639	228339
	1.5							28003	137434	6507	11291	23714	74996	73420	223721
	3.5							27856	136545	6494	11201	21174	67084	71320	214830
	5.0							27856	136570	6521	11361	19241	61010	69894	208941

Table II. Young' s modulus and Poisson' s ratio

	Bulk modulus (GPa)	Shear modulus (GPa)	Young' s modulus (GPa)	Poisson' s ratio
Titanium	87.50	40.38	105.0	0.30
Cortical bone	11.67	5.38	14.0	0.30
Trabecular bone	5.0	0.52	1.5	0.45



a) b) c) d) e)

Fig. 1. a) finite element model; b) 0 mm of bone loss; c) 1.5 mm of bone loss; d) 3.5 mm of bone loss; e) 5.0 mm of bone loss.

N(Fig. 1). In three-dimensional models, stress is generally represented by a stress tensor, which has six components. The von Mises equivalent stress (which expresses all these components in one value) was computed for the length and diameter variations mentioned above using PAM-CRASH 2G (ESI Group, version 2004, France). To investigate the effect of the preload, we compared the stress model A (3.75 mm diameter implant, 0 mm of bone loss with preload) to the stress model B(3.75 mm diameter implant, 0 mm of bone loss without preload).

To evaluate fixture fracture by alveolar bone resorption, the fixtures were investigated according to the degree of alveolar bone loss. Bone loss was classified as 0, 1.5, 3.5, or 5.0 mm. The 0 mm and 5 mm distances were controls, while the clinically important distance between the platform and the first thread is 1.5 mm, and the distance between the third and fourth threads, where the end of the abutment screw is postulated to be, is 3.5 mm. The bone was modeled as a cancellous bone surrounded by a 2 mm layer of cortical bone. Using these 12 models (four degrees of bone loss and three implant diameters), the effects of the loading offset, the extent of alveolar bone resorption, and the size of the fixture was evaluated. The PAM-CRASH 2G simulation software was used for analysis of stress. The PAM-VIEW(ESI Group, version 2004, France) and HyperView programs (Altair Engineering, Inc., Troy, MI, USA) were used for post processing. The fracture of the abutment screw and implant is determined by von Mises stress patterns considering the S-N curve and yield strength of commercially pure titanium and titanium alloys. A tensile strength of commercially pure titanium is ranging from 275 to 590 MPa, and a yield strength of 170 to 485 MPa, which strength is controlled primarily through the content of oxygen and iron content. The higher the oxygen and iron content, the higher the strength. Titanium alloys have a tensile strength of 860 MPa, and a yield strength of 795 MPa.[16,30]

RESULTS

When a 3.75 mm diameter implant with 0 mm of bone loss, a fracture of the abutment screw is possible starting from a 4 mm offset(Fig. 2).

When a 3.75 mm diameter implant has 1.5 mm of bone loss, a fracture of the abutment screw is possible from a 4 mm offset, and a fracture of the implant is possible from a 6 mm offset(Fig. 3).

When a 3.75 mm diameter implant has 3.5 mm of bone loss, the fracture of abutment screw and implant is possible from a 4 mm offset(Fig. 4).

When a 3.75 mm diameter implant with 5.0 mm of bone loss, a fracture of abutment screw and implant is possible from 4 mm offset(Fig. 5). The amount of stress concentrated in the implant increases with alveolar bone resorption(Figures 2 to 5).

When a 4.0 mm diameter implant has 0 mm of bone loss, a fracture of the abutment screw is possible from a 4 mm offset(Fig. 6).

When a 4.0 mm diameter implant has 1.5 mm of bone loss, a fracture of the abutment screw is possible from a 4 mm offset(Fig. 7).

When a 4.0 mm diameter implant has 3.5 mm of bone loss, fractures of abutment screw and implant are possible from a 4 mm offset(Fig. 8).

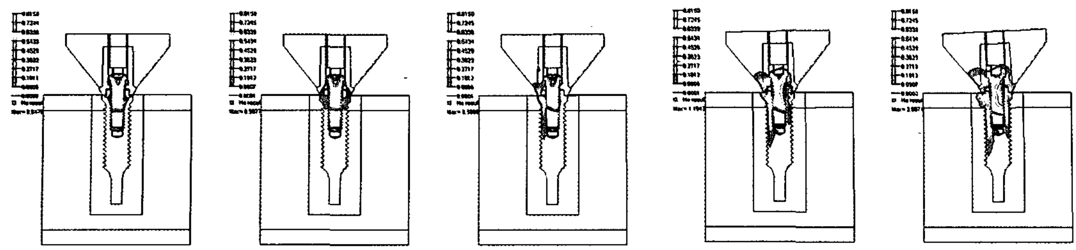
When a 4.0 mm diameter implant has 5.0 mm of bone loss, fractures of abutment screw and implant are possible from a 4mm offset(Fig. 9).

When a 5.0 mm diameter implant has 0 mm of bone loss, a fracture of the abutment screw is possible from a 4 mm offset(Fig. 10).

When a 5.0 mm diameter implant has 1.5 mm of bone loss, a fracture of abutment screw is possible from a 6 mm offset(Fig. 11).

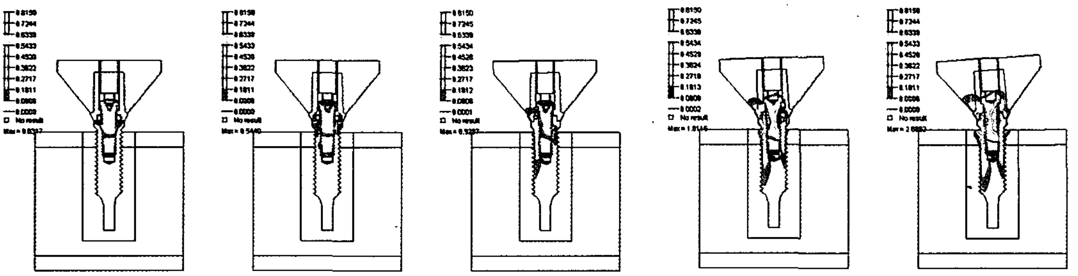
When a 5.0 mm diameter implant has 3.5 mm of bone loss, a fracture of the abutment screw is possible from a 6 mm offset(Fig. 12).

When a 5.0 mm diameter implant has 5.0 mm of bone loss, a fracture of the abutment screw is possible from a 6 mm offset(Fig. 13). In a 5.0 mm diameter



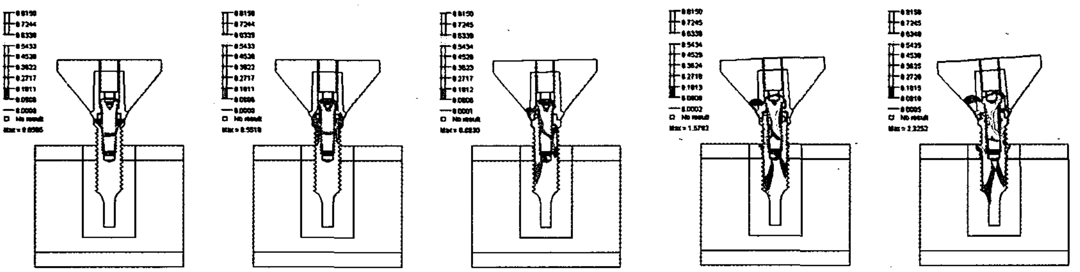
a) b) c) d) e)

Fig. 2. The 3.75 mm diameter implant with 0 mm of bone loss : a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



a) b) c) d) e)

Fig. 3. The 3.75 mm diameter implant, 1.5 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



a) b) c) d) e)

Fig. 4. The 3.75 mm diameter implant, 3.5 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.

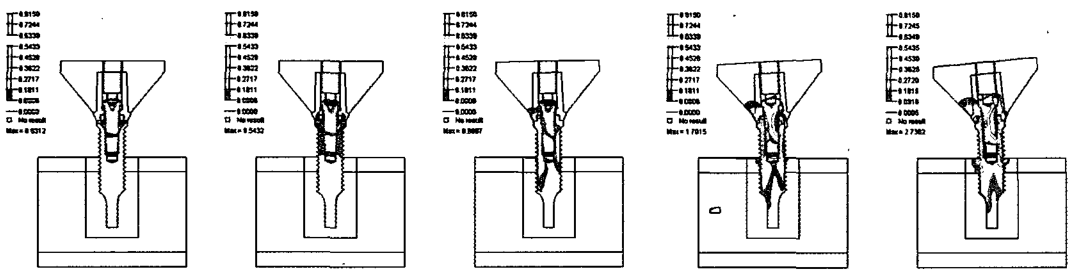


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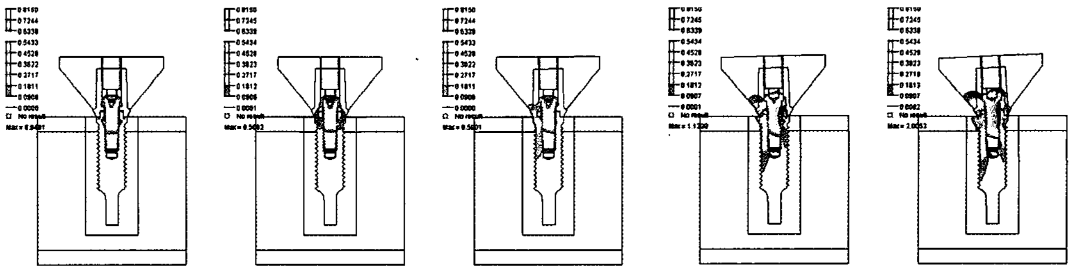
Fig. 5. The 3.75 mm diameter implant with 5.0 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



a) b) c) d) e)

Fig. 6. The 4.0 mm diameter implant with 0 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.

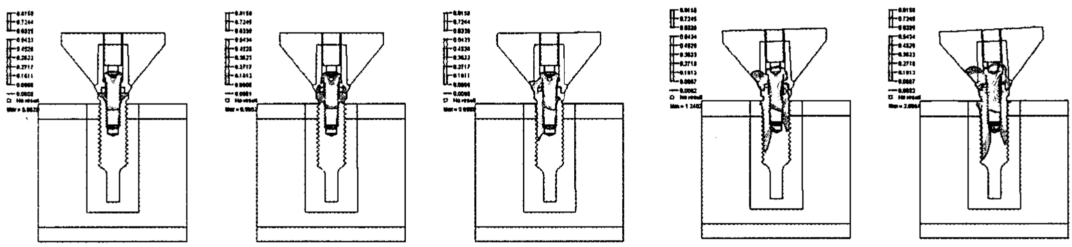


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Fig. 7. The 4.0 mm diameter implant, 1.5 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.

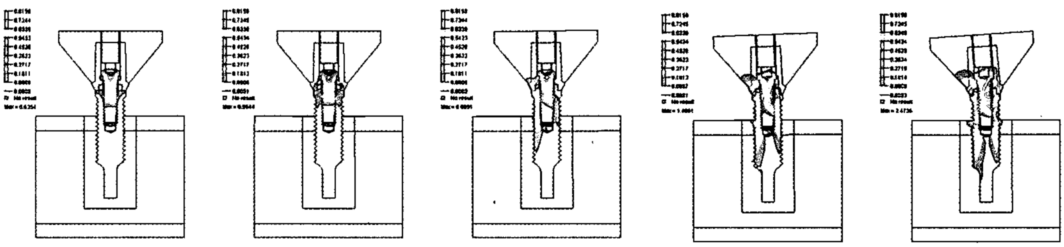

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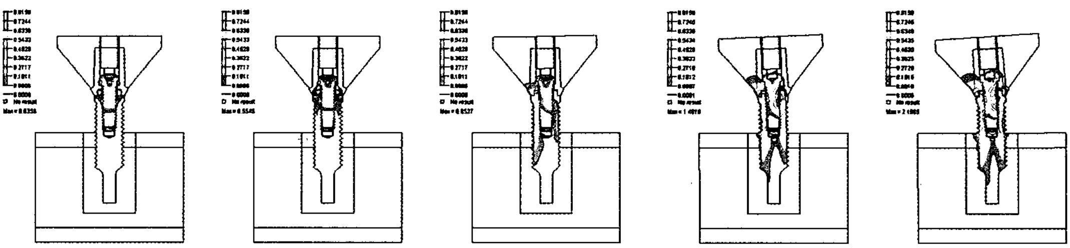
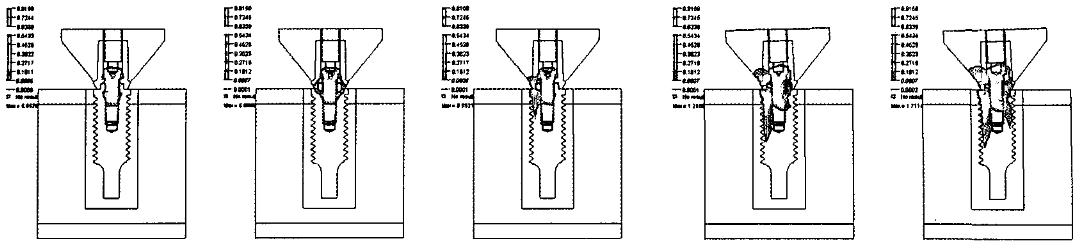
Fig. 8. The 4.0 mm diameter implant with 3.5 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



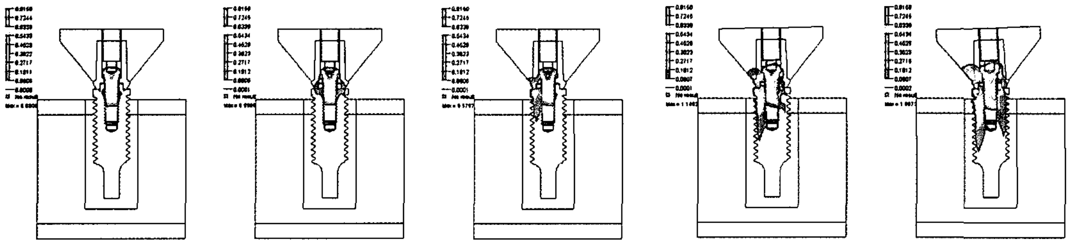
a) b) c) d) e)

Fig. 9. The 4.0 mm diameter implant with 5.0 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



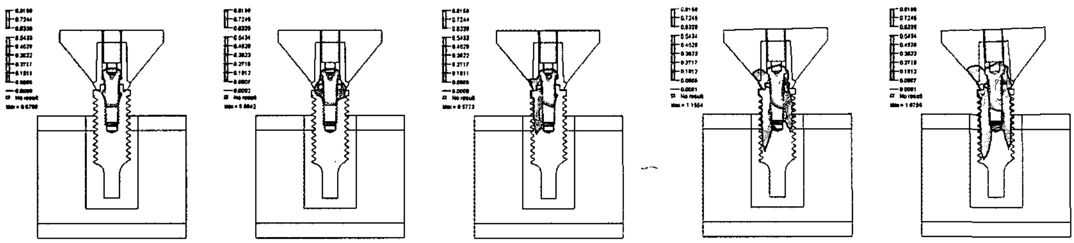
a) b) c) d) e)

Fig. 10. The 5.0 mm diameter implant with 0 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



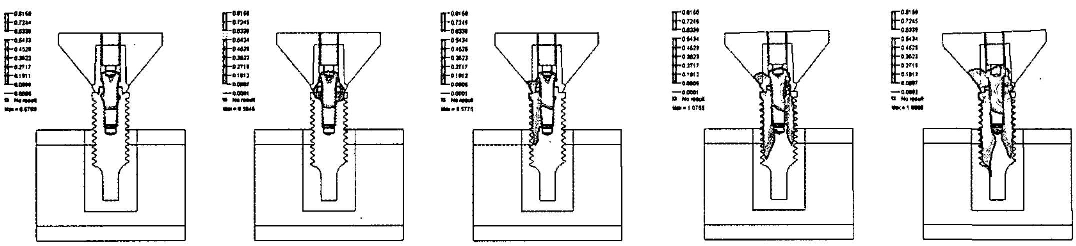
a) b) c) d) e)

Fig. 11. The 5.0 mm diameter implant with 1.5 mm bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



a) b) c) d) e)

Fig. 12. The 5.0 mm diameter implant with 3.5 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.



a) b) c) d) e)

Fig. 13. The 5.0 mm diameter implant with 5.0 mm of bone loss: a) stress distribution after preload; b) 600 N loading at the center of prosthesis; c) 2 mm offset loading; d) 4 mm offset loading; e) 6 mm offset loading.

implant with bone loss of 0 mm, a fracture of abutment screw is possible from 4 mm offset, but when bone loss is 1.5, 3,5, or 5.0 mm, an abutment screw fracture is possible from a 6 mm offset. The fracture of a 5.0 mm diameter implant is impossible(Figures 10 to 13).

1. Effects of preload on von Mises stress (stress model A versus B)

With the application of preload, the von Mises stress is maintained until an offset of 2 mm, but the stress is increased over a loading offset of 2 mm. Without preload, the von Mises stress increases with the amount of loading offset. However the stress patterns are similar to each other in the bone(Figures 14 to 17).

2. Effects of bone loss on implant stress

Without bone loss, the stress patterns are similar among the implants, but with the progression of bone loss, the 3.75 mm diameter implant showed the highest von Mises stress and the 5.0 mm diameter implant showed the least stress. The von Mises stress of the 4.0 mm diameter implant was similar to that of the 5.0 mm diameter implant at 1.5 mm of bone loss. However, while the stress pattern of 4.0 mm diameter implant is parallel to that of 3.75 mm diameter implant with progression of bone loss, the stress is less than 3.75 mm diameter implant. In general, the von Mises stress is increases with loading offset, and the amount increases faster above an offset of 2 mm(Figures 18 to 21).

3. Effects of bone loss on abutment screw stress

Without bone loss, the stress pattern of the abutment screw is similar among the implants. The von Mises stress of the abutment screw decreases slightly with a loading offset of less than 2 mm. The stress increases faster with a loading offset of over 2 mm. The 5.0 mm diameter implant has the least stress, and the 4.0 mm diameter implant is similar to 5.0 mm diameter implant at 1.5 mm of bone loss, but the stress pattern is parallel to that of 3.75 mm diameter implant with progression of bone loss(Figures 22 to 25).

4. Effects of bone loss on abutment stress

The von Mises stress of the abutment increases steeply for a loading offset of over 2 mm in all implants. The stress patterns are more similar among

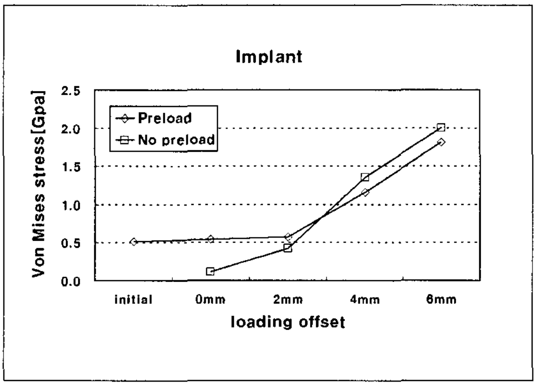


Fig. 14. Von Mises stress of implant with and without preload.

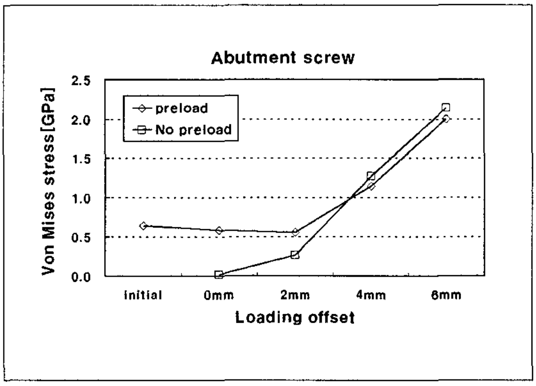


Fig. 15. Von Mises stress of abutment screw with and without preload.

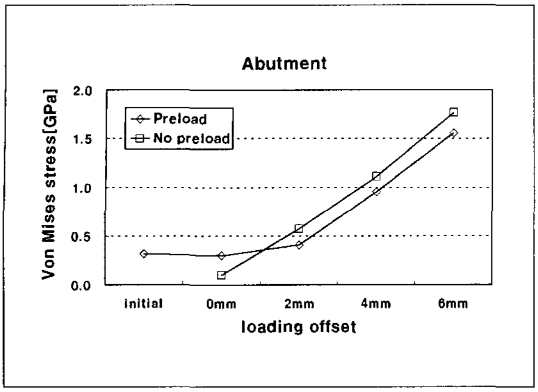


Fig. 16. Von Mises stress of abutment with and without preload.

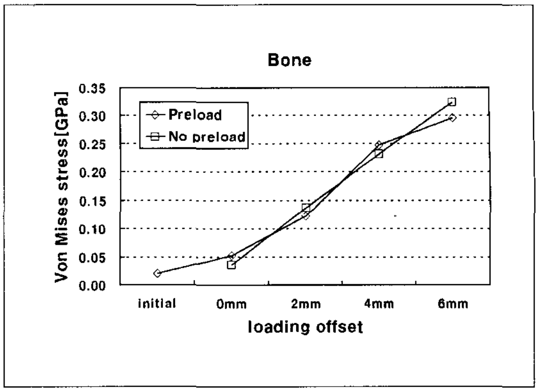


Fig. 17. Von Mises stress of bone with and without preload.

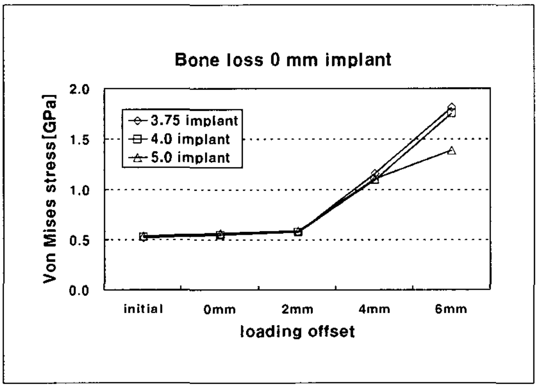


Fig. 18. Von Mises stress of implants with no bone loss.

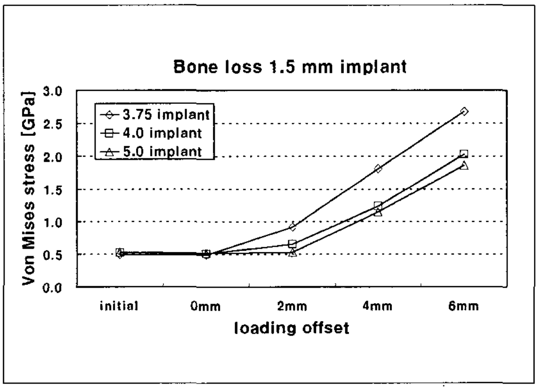


Fig. 19. Von Mises stress of implants at 1.5 mm of bone loss.

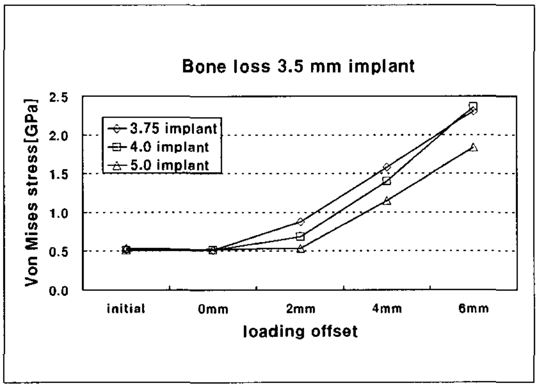


Fig. 20. Von Mises stress of implants at 3.5 mm of bone loss.

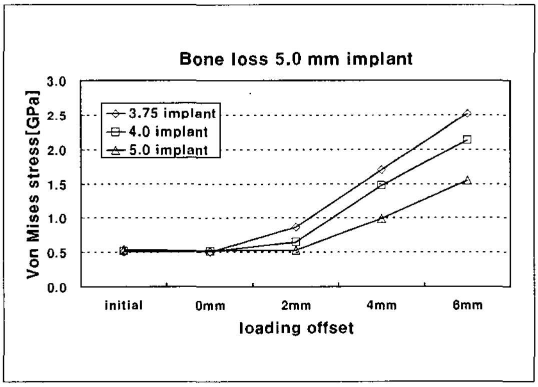


Fig. 21. Von Mises stress of implants at 5.0 mm of bone loss.

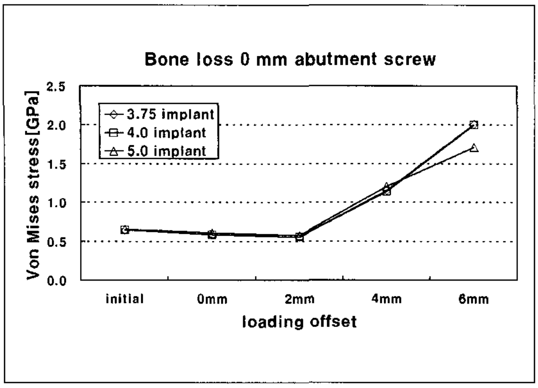


Fig. 22. Von Mises stress of abutment screw at 0 mm of bone loss.

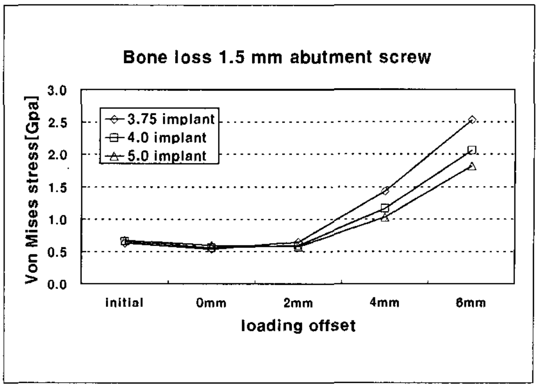


Fig. 23. Von Mises stress of abutment screw at 1.5 mm of bone loss.

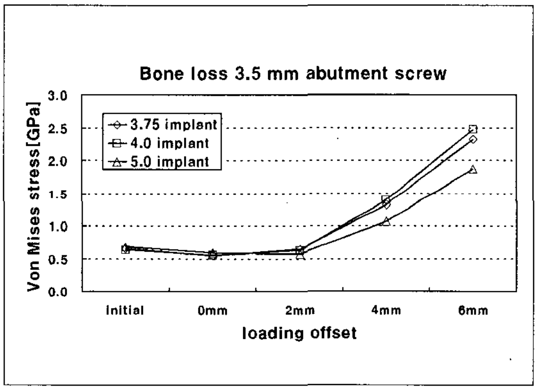


Fig. 24. Von Mises stress of abutment screw at 3.5 mm of bone loss.

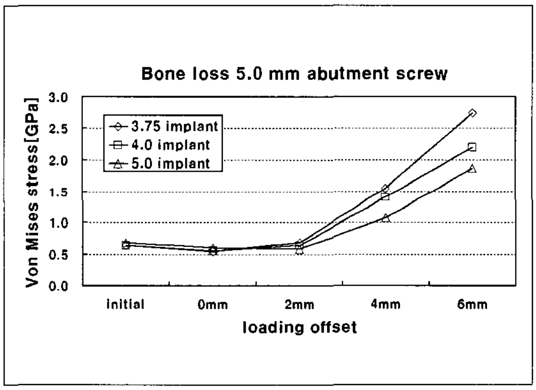


Fig. 25. Von Mises stress of abutment screw at 5.0 mm of bone loss.

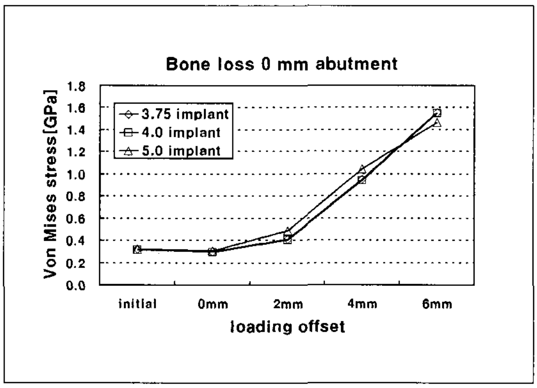


Fig. 26. Von Mises stress of abutment at 0 mm of bone loss.

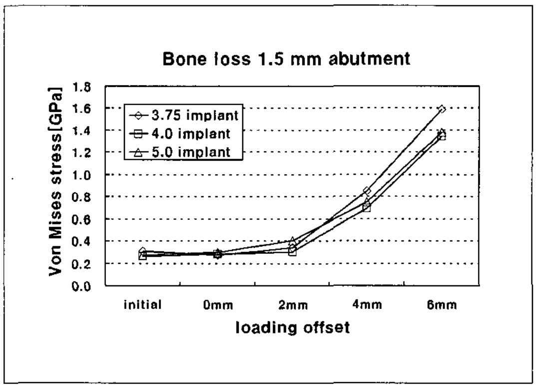


Fig. 27. Von Mises stress of abutment at 1.5 mm of bone loss.

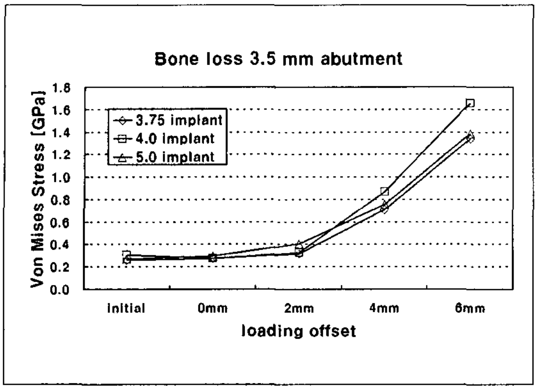


Fig. 28. Von Mises stress of abutment at 3.5 mm of bone loss.

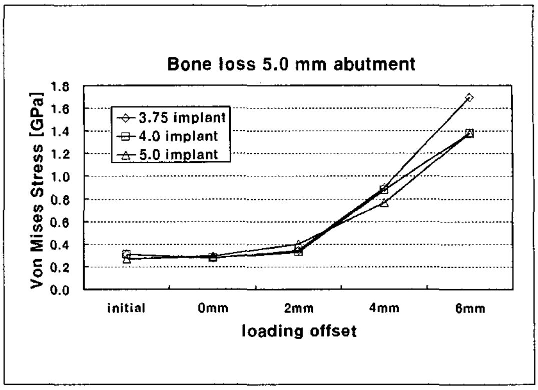


Fig. 29. Von Mises stress of abutment at 5.0 mm of bone loss.

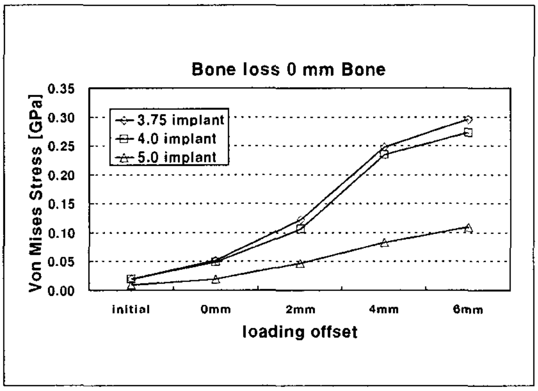


Fig. 30. Von Mises stress of bone at 0 mm of bone loss.

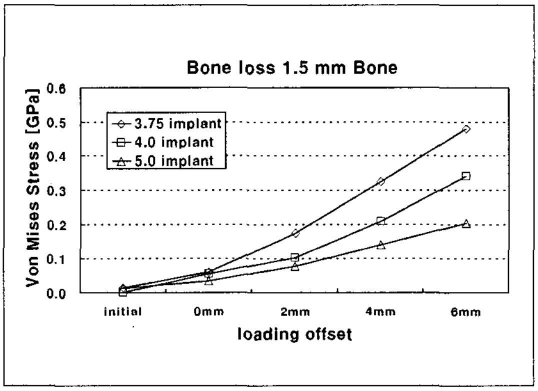


Fig. 31. Von Mises stress of bone at 1.5 mm of bone loss.

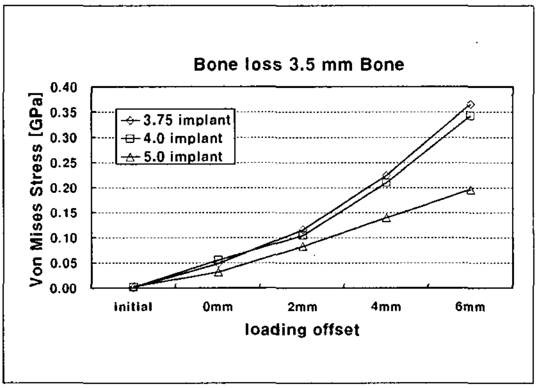


Fig. 32. Von Mises stress of bone at 3.5 mm of bone loss.

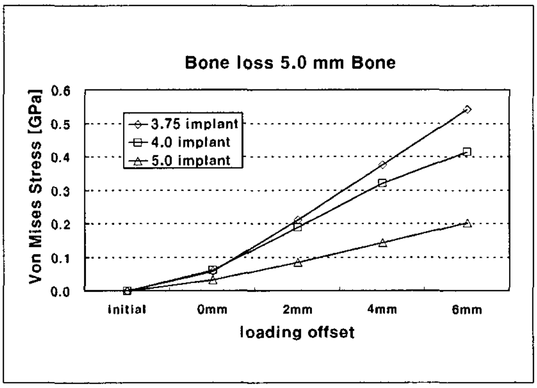


Fig. 33. Von Mises stress of bone at 5.0 mm of bone loss.

the abutments of the three implant types than among other components(Figures 26 to 29).

5. Effects of bone loss on bone stress

The bone of the 5.0 mm diameter implants has the least von Mises stress, and the stress increases with loading offset. The bone of the 4.0 mm diameter implant have stress levels between those of the 3.75 mm and 5.0 mm diameter implants at 1.5 mm of bone loss, but the bone stress levels of the 4.0 mm diameter implants are similar to those of 3.75 mm diameter implants at 3.5 mm of bone loss(Figures 30 to 33).

DISCUSSION

1. Fatigue

Commercially pure titanium, classes 1 to 4, has tensile strengths ranging from 275 to 590 MPa, with yield strengths ranging from 170 to 485 MPa. Titanium alloys have a tensile strength of 860 MPa and a yield strength of 795 MPa.[33] The most intensive stress is concentrated at the abutment screw, the fixtures have the next lower level of stress, and the abutment has the least stress of the implant components. Clinically, abutment screws are made of titanium alloy or gold alloy. The abutment is made of grade III commercially pure titanium. In contrast, fixture requires a better biological response for osseointegration, so the Branemark fixture was made of grade I commercially pure titanium. However, grade I titanium has a lower strength(tensile strength 275 MPa, yield strength 170 MPa) than titnaium grades III or IV, and the incidence of fracture is higher for grade I titanium. An additional alternative is to use a different implant material to provide a higher threshold for fatigue failure, since fatigue appears to have some relationship with the tensile strength of the implant materials. We tested a grade IV titanium implant (tensile strength 590 MPa, yield strength 485 MPa). The mechanical weakness of implant systems is a problem in clinical situations. Increased knowledge of the mechanical properties of implant systems is therefore essential.

To determine the exact timing of fractures of the implant and abutment screw, specific S-N curve experiments for the implant and abutment screw are needed for notch design. Metallurgical concentrators, such as superficial defects and notches, poor surface polishing, inclusions, and porosity are particularly dangerous, and can lead to catastrophic failure and a reduced number of cycles.[34]

We found a peak von Mises stress that is higher than the tensile strength of titanium. However, this value is restricted to a small notch area of the implant and abutment screw. Torsional relaxation of the screw shaft, embedment relaxation, and localized plastic deformation of the gold alloy and the opposing titanium threads were the most likely explanations for the exclusion of this peak von Mises stress.[35]

Commercially pure titanium has a tensile strength ranging from 275 to 590 MPa, and this strength is controlled primarily by the oxygen and iron contents. The higher the oxygen and iron contents are, the higher the strength. The fatigue strength (10^7 cycles) is roughly equivalent to 50% of the tensile strength. The fracture toughness of titanium alloys ranges from 28 to 108 $MPa.m^{1/2}$, and is in negatively correlated with tensile yield strength.

Fatigue is the progressive, localized, and permanent structural change that occurs in a material subjected to repeated, or fluctuating, strains at nominal stresses with maximum values less than the static yield strength of the material. Fatigue may culminate into cracks and cause fractures after a sufficient number of cycles. Fatigue damage is caused by the simultaneous actions of cyclic stress, tensile stress, and plastic strain. The plastic strain resulting from cyclic stress initiates the crack, and the tensile stress promotes crack growth (propagation). Although compressive stresses will not cause fatigue, compressive loads can result in local tensile stresses, and a fatigue crack may form even in a flow-free metal having a highly polished surface with no stress and no stress concentrators. One study reported shear cracks that initiated at the root of the thread and propagated into the inner section of the screw.[36]

2. Preload

An implant abutment is joined to the implant by an abutment screw. When the screw is tightened, a tightening torque is applied as a moment to the head of the abutment screw. The applied moment is transformed along the interface of the abutment screw thread surfaces and the implant threaded bore surfaces. The transformed force then induces a contact force in the interface between the abutment and the implant bearing surfaces that are being clamped together. This contact force clamping together the abutment and the implant is called the preload. As the tightening torque is increased above the level of the initial contact force, the preload stress in the abutment/implant interface is increased to a point. This point is within the material elastic range of the abutment screw. When the optimum preload is achieved, the abutment screw experiences the entire external load applied to the clamped parts. At this point, the screw joint is said to be protected against external force applications as long as these external loads do not exceed the preload. Thus, the accuracy of the preload reached during screw tightening and the clamping of the abutment and the implant together become a major and critical subject for studying the dynamic loading of the implant complex.[23]

Preload maintains implant-abutment joint stability within a limited offset point against occlusal force. Without preload, the joint stability can not be maintained. Bickford divided the process of loosening into two phases. The first involves the slippage of the joint surfaces, which is related to the application of transverse and axial forces that are sufficiently large to overcome the frictional and compressive forces that keep the contracting surfaces in a fixed relationship. The second phase occurs when the preload is reduced to such an extent that external forces and vibrations cause the mating threads to back off. Once this stage has been reached, the screw joint ceases to function and the clamped surfaces separate. In this state the screw is liable to be loaded in flexion and may fracture.[37]

The preload value could be acquired by ABAQUS, but there is no three-dimensional finite element analysis of the stress distribution of an implant with the application of preload. Preload maintains the stability of the implant components up to a loading offset of 2 mm when 600 N of biting force are applied. Without preload, stress is increased with loading offset. A loading offset beyond the effect of the preload will make the outbreak of many problems possible(Figures 14 to 17).

For the most part, finite element studies that have analyzed implant systems for stress development, magnitude, and displacement have not directly measured the development of the preload. This is most likely due to the difficulty in accurately creating a three-dimensional model of the implant complex with the exact dimensions of the thread helix for the abutment screw and screw bore.[30] With such a model, the preload generated by a torque application can be simulated. However, creating such a model is a difficult task and requires expertise in modeling and in finite element analysis. In the assembly of a screw joint, it is important to know the optimum preload that will maintain the components together. However, Lisa' s experiment using ABAQUS cannot rotate real screws, but only estimates the relationship between torque and preload.[30] In our experiments, we rotated the abutment screw without contact conditions, and applied the contact conditions for the stress distribution by preload.

In the external hex implant design, the abutment screw alone is primarily responsible for maintaining the implant abutment complex assembly under functional loads. Therefore, the axial preload of the abutment screw is a determining factor for the stability of the connection. The mechanical effect of the preload in implant-abutment complex was considered.[23,27-29]

In practice, the achievable preload is limited by the superpositioning of additional tension related to the external loading. For instance, Haack et al.[38], in an in vitro mechanical test, measured abutment screw elongation to determine preload during abutment/implant joint assembly. The abutment screw was measured before screw tightening and after preload development. The elongation of the screws after applying the manufacturer' s recommended tightening torques was within the elastics range. The induced stresses were 57.5% and 56% of the yield strengths for gold alloy and titanium, respectively. For the tightening of screws beyond the manufacturer' s recommended torques, the mean preload was 486.2(± 57.9) N using gold alloy screws and 381.5(± 72.9) N with titanium screws.

3. Maximum biting force

Maximum bite force is generated when ipsilateral chewing of food causes mean maximum forces in centric occlusion and chewing and grinding. The force created by ipsilateral chewing will cause a bending moment due to axial force.[32] The stress generated when there is no bone loss is the least, and the area of stress in the implant components is increased with bone loss. However, the increase of von Mises stress is not coincident with bone loss. We hypothesize that the axial force with loading offset causes a bending moment without oblique loading. This can apply bending and compressive forces to the bone and implant components. When oblique force is applied in chewing and grinding, the bone and implant components are exposed to shear and bending forces. In the fracture of an implant, the worst condition hypothesized is the combination of a hard food bite and axial loading with offset without oblique force. In a clinical situation, occlusal contact points are combined with each other, and the worst condition of loading offset is smaller than the expected value.

4. Fixture diameter

The 4.0 mm diameter fixture may substitute 3.75 mm diameter fixture in the prevention of implant fracture, but it is not as effective as 5.0 mm diameter fixture. Decreasing the loading offset by reducing of occlusal table size may be essential for avoiding fixture fracture.

Implant size influences the area of possible retention in the bone; factors such as occlusion, masticatory force, the number of implants, and the implant position within the prosthesis affects the forces on the bone adjacent to the implants.[39]

An increase in the implant diameter decrea- sed the maximum von Mises equivalent stress around the implant neck more than an increase in the implant length, by resulting in a more favorable distribution of the simulated masticatory forces. Holmgren[40] reported that implant diameter, shape, and load direction influence stress distribution. Alveolar bone reduction caused by extraction wound healing, the wearing of removable dentures, and bone injuries, together with anatomic structures (such as the canine fossa, antrum, nasal cavity or mandibular canal), may limit implant size or require implant placement into positions for which angled abutments are needed. Thus, such implants may be unable to distribute the masticatory forces effectively.[41]

The effects of fixture diameter showed that 5.0 mm diameter implant is advantageous. Large implant diameters provide for more favorable stress distributions. Finite element analysis has been used to show that stresses in cortical bone decrease in inverse proportion to an increase in implant diameter with both vertical and bending loads. Hence, in order to reduce the risk of implant fractures, it is recommended that the diameter of the implant be increased or the diameter of the abutment screw be decreased. However, Holmgren et al. showed that using the widest diameter implant is not necessarily the best choice when considering stress distribution to surrounding bone; within certain morphologic limits, an optimum dental implant size exists for decreasing the stress magnitudes at the bone implant interface.[40,42]

5. Cortical bone loss

Postulated boundary conditions restrain six degrees of freedom, involving rotation and translation in three coordinate axes for correspondent nodes located at the most external mesial and distal planes and bottom. We postulate boundary conditions for modeling at small area, but the stress pattern is different between the maxilla and mandible. A maxillary implant is fixed to the cranium, but a mandibular implant is located at a mobile position with the temporomandibular joint, and its stress pattern differs from the maxillary implant. The full arch modeling may be necessary for correct clinical data.

A higher predisposition of the maxilla for implant fracture could be related to the fact that the presence of a weaker bone can lead to bone loss at high loads and an increased bending moment on the implants.[2,3] With the progression of alveolar bone loss, the stress area of the implant increases, and the tendency to fracture also increases.

On the basis of clinical observations, some authors state that during the first year after implant loading, the marginal bone loss around the neck ranges from approximately 0.5 to 1 mm or 1.5 mm. Subsequently, the rate of bone loss is either considered to be stationary or significantly reduced (bone loss of approximately 0.1 mm), or else the resorption of the bone crest continues and the implant is ultimately lost.[19,43] These findings are in accordance with recent three dimensional mathematical models of dental implants under non-axisymmetric loading, indicating the maximum stress occurred around the implant neck.[44,45]

The manner in which bone is loaded may be essential to its response. Bone is usually subjected to cyclic loads, with results that differ from static loads. If a sufficient number of repetitive load cycles are applied, stress microfractures in the bone may occur. After bone microfractures occur, the microdamage caused by stresses greater than normal levels may stimulate osteoclastic activity to remove the damaged bone.[46] The highest stresses were concentrated in the cortical bone. Stresses under oblique loading were approximately 10 times greater than under axial loading. The type of veneering material and the size of the mandible had no effect on the stress levels under similar loading conditions. The ultimate strength of human cortical bone ranges from 72 to 76 MPa in tension and from 140 to 170 MPa in compression.[47]

We postulate alveolar bone resorption with constant cortical bone thickness.

In the resorption models, the bone defects were lined by a cortical shell to simulate an increase in bone density (lamina dura radiographically observed in clinically stable implants with bone resorption).[48,49] The tendency of most coronal bone to undergo remodeling adjacent to implants and to form a new cortical bone layer at a lower level was also found in a dog study.[50] In the pure vertical resorption models, stresses of the cortical and cancellous bone increased with resorption depth. In contrast, in the conical resorption models, the stress of the cortical bone was lower than in the non-resorption and corresponding pure vertical resorption models for all resorption depths. Thus, slight conical resorption may partially be the result of biomechanical adaptation of bone to occlusal loads in the successfully integrated implants.[51] Since the cortical bone had a much higher elastic modulus than the cancellous bone, it was the load-carrying member for all cancellous bone quantities, regardless of load direction.[52]

Van Oosterwyck et al.[53] carried out a finite element analysis of an axially loaded Branemark implant, in which the peak von Mises stress appeared at the point on the thread where the lower flank passed into the curved top, when the bone implant interface was assumed to resist only compressive stress. Clearly, with a very stiff implant, the stresses in the bone will be the same irrespective of whether the interface to the superstructure is located at the level of the marginal bone or more coronally.

That mechanical stimuli affect the architecture of bone has garnered wide acceptance.[54] Strains exceeding the physiologic tolerance threshold of bone (above 3000 microepsilons) may explain why bone loss is observed at the tip of threads in histological analyses of implants.[55,56] High marginal bone loss around conical Branemark implants has been attributed to the smooth implant neck, avoiding the generation of optimum stresses and strain in the vicinity of the implant.[57,58] The occurrence of marginal bone loss is often attributed to poor oral hygiene and biomechanical factors. Clinical studies have shown that bone loss around the implants that may have led to implant failure were associated in many cases with unfavorable loading conditions. Inappropriate loading causes excessive stress in the bone around the implant, and may results in bone resorption.[59,60]

The load transfer from implants to the surrounding bone depends on the type of loading, the bone-implant interface, the length and diameter of the implants, the shape and characteristics of the implant surface, the prosthesis type, and the quantity and quality of the surrounding bone.[61]

Because bending stress is increased by alveolar bone resorption, implant fracture is possible with alveolar bone resorption. But fracture of the 5.0 mm diameter implant is impossible with limited bite force of our study, it is a viable option to overcome implant fracture by altering implant diameter.

CONCLUSIONS

The fracture of 3.75 and 4.0 mm diameter implants is possible, according to finite element analysis, so the control of bone loss and loading offset by the occlusal table of the prosthesis should be considered. The conclusions from our experiment were as follows:

1. Preload maintains implant abutment joint stability within a limited offset point against occlusal force.
2. Von Mises stress of the implant, abutment screw, abutment, and bone was decreased with increasing of the implant diameter.
3. With severe advancing of alveolar bone resorption, fracture of the 3.75 and the 4.0 mm diameter implant was possible.
4. With increasing of bending stress by loading offset, fracture of the abutment screw was possible.

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