Finite Element Analysis of Stress in Bone Adjacent to Dental Implants

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Understanding how clinical variables affect stress distribution facilitates optimal prosthesis design and fabrication and may lead to a decrease in mechanical failures as well as improve implant longevity. In this study, the many clinical variations present in an implant-supported prosthesis were analyzed by a 3-dimensional finite-element method. The anterior segment of a human mandible treated with 5 implants supporting a curved beam was created to perform the tests. The variables introduced in the computer model were cantilever length, elastic modulus of cancellous bone, abutment length, implant length, and framework alloy (AgPd or CoCr). The computer was programmed with physical properties of the materials as derived from the literature, and a 100-N vertical load was used to simulate the occlusal force. Images with fringes of stress were obtained, and the maximum stress at each site was plotted in graphs for comparison. Stresses tended to be concentrated at the cortical bone around the neck of the implant closest to the load, whereas stresses in cancellous bone were considered low. In general, the stress distribution was better with stiffer cancellous bone, longer abutments and implants, and shorter cantilevers. The use of a CoCr alloy framework appears to contribute to a better stress distribution.

Key Words: finite element analysis, implant prosthesis, bone stress

Introduction

Since the introduction of the technique of osseointegration by Branemark in 1969,[1] the widely accepted and clinically demonstrated prosthetic solution for edentulous patients who cannot adapt to a mandibular complete denture is the construction of a bridge supported by five 10-mm-long implants placed using a 2-stage surgery in the interforaminal region (zone 1).

Stage I surgery aims at implant placement in selected host bone sites that are unloaded during the healing period. After stage I surgery and prosthetic conclusion, the major cause of failure has been attributed to implant overload.[2,3] Since the occlusal load will be transferred to the implants and subsequently to the bone, it is believed that the biomechanics of the implant-supported prosthesis play an important role in the longevity of the bone around dental implants.[4] Under load, bone tissue undergoes a remodeling process, which ultimately influences the long-term function of a dental implant system.[5] The idea is to keep stresses below the failure stress of the bone.[4,6,7]

However, the amount of force that can be applied to a dental implant without jeopardizing the surrounding bone is unknown.[8] Factors such as the curvature of the mandible, the density of cancellous bone, the length of implants and abutments, the length of cantilever, the number of implants, and the framework stiffness influence stress distribution and have been objects of study.[5,9–11] Failure prevention demands testing and stress analysis of the implants and tissues in vitro as well in vivo.[9]

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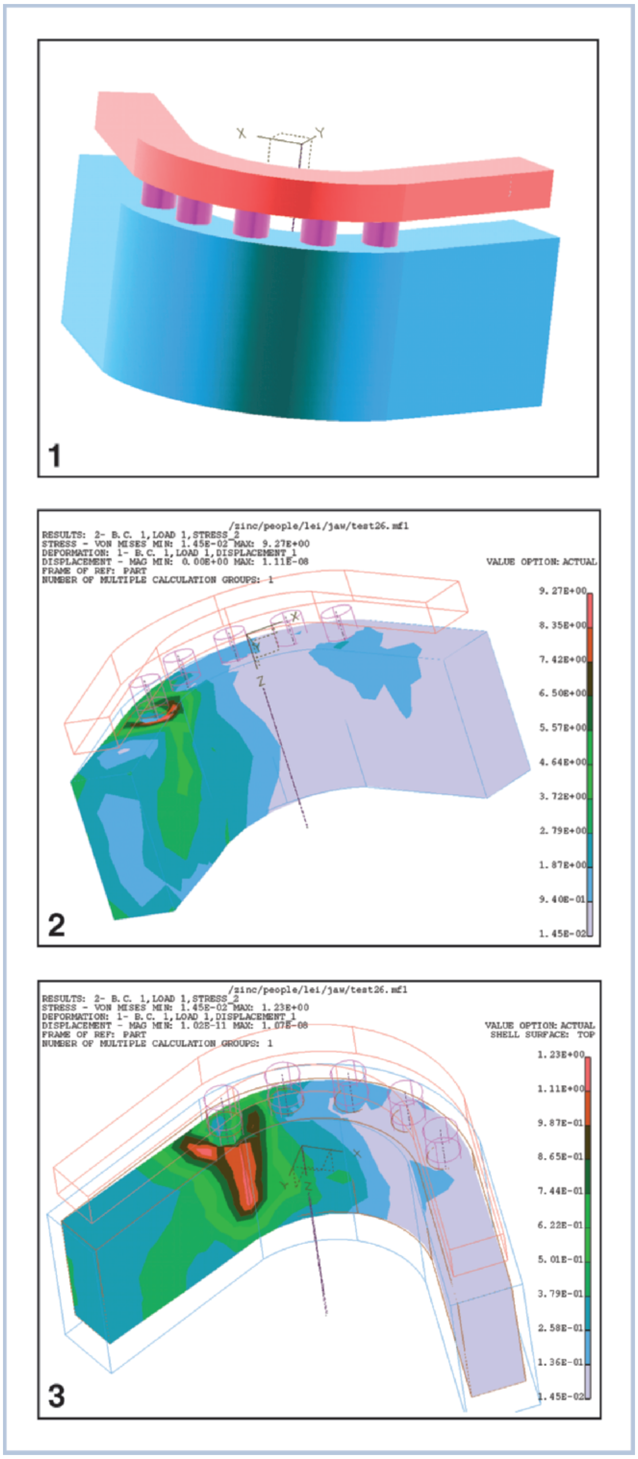


FIGURES 1–3. FIGURE 1. Graphic representation of the 3-D model used in the study. FIGURE 2. Fringes of stress in cortical bone (basic model). FIGURE 3. Fringes of stress in cancellous bone (basic model).

Reports indicate that finite element analysis (FEA) is a valuable tool for examining the stress distribution in bone adjacent to dental implants despite its inherent limitations.[12–16]

The creation of a computerized model of an implant-supported fixed prosthesis permits an analysis of the possible variation in stress distribution that is likely to occur with diverse prosthetic designs and occlusal load variables. A good understanding of how each component of the implant prosthesis behaves under load could facilitate optimal prosthesis design and fabrication. This could lead to a decrease in likelihood of mechanical failures as well as improve implant prosthesis longevity.

The purpose of this study was to observe stress distribution in bone adjacent to dental implants by means of FEA. Such observation could be used to make comparisons to clinical treatment outcomes.

MATERIALS AND METHODS

Model

A computerized 3-dimensional (3-D) finite-element model of the anterior segment of a human mandible provided with an implant-supported bridge was created (Figure 1). The basic model consisted of a curved beam with radius 15.0 mm and dimensions 69.0 mm long, 14.0 mm high, and 6.0 mm wide. This beam was covered with a 1.0-mm-thick layer on the buccal, occlusal, and lingual surfaces and a 3.0-mm layer at the base to simulate cortical bone. The final external dimensions were therefore $71.0 \times 18.0 \times 8.0$ mm, respectively. Five 10.0-mm cylinders 3.75 mm in diameter were placed at the center of the beam, their centers 7.0 mm apart from each other. The cylinders were provided with 3.0-mm-high extensions to simulate the abutments. A second beam ($71.0 \times 4.0 \times 6.0$ mm) was added in connection to the abutments to simulate a framework. This resulted in a model with 1714 nodes and 8062 elements. The model was fixed at both ends for the sake of the stress analysis. All materials, bone included, were assumed to be linearly elastic and isotropic. An FEA program (I-DEAS Structural Dynamics Research, Milford, Ohio) installed in a desktop computer was used to analyze the many possible variations in prosthetic design and occlusal load. The model was fed with elastic properties of the materials as derived from the literature (Table 1). Elastic modulus for cortical bone was assumed to be 13.7 GPa, while for cancellous bone it was assumed to be 1.5 GPa based on a density of 25%. A 100-N vertical load was applied at 15.0 mm distally to the terminal abutment to simulate the occlusal force (Table 2). Images with the fringes of stress were obtained for each of the components of the model, and maximum

TABLE 1

Elastic properties of materials used for the FEA model

Material	E* (GPa)	μ	Reference
Titanium	110	0.35	17
AgPd alloy	95	0.33	18
CoCr alloy	218	0.33	18
Cortical bone	13.7	0.30	5, 13
Cancellous bone	1.5 (d = 25%)	0.30	13
	4.0 (d = 50%)	0.30	13
	7.9 (d = 75%)	0.30	13

*E indicates elastic modulus; μ, Poisson's ratio; d, density.

TABLE 2

Clinical variables selected for FEA analysis

Cantilever length	10 mm	15 mm*	20 mm
Cancellous bone	E*† = 1.5 GPa	E = 4.0 GPa	E = 7.9 GPa
Abutment length	3.0 mm*	5.5 mm	7.0 mm
Implant length	10 mm*	13 mm	15 mm
Framework alloy	E = 95 GPa*	E = 218 GPa	

*Features of the basic model.
†E indicates elastic modulus.

Von Mises stress at each site was plotted in graphs for comparison.

Variables

Five clinical variables were chosen to be evaluated and are summarized in Table 2 for ease of reference. Each variable was introduced alternately on the basic model, all other conditions being equal.

Cantilever length was defined as the position where the load was applied relative to the center of the terminal abutment. The quality of the cancellous bone was expressed in terms of its modulus of elasticity (E) as derived from the literature. Abutment and implant lengths were based on the Branemark system. Framework alloy was selected according to the modulus of elasticity of the AgPd (E = 95 GPa) and CoCr (E = 218 GPa) alloys.

RESULTS

In the basic model, stress tended to be concentrated at the cortical bone on the disto-lingual aspect of the implant closest to the load (site #1). It gradually decreased as far as the implant is from the load and increased again on the terminal implant on the side opposite to the load (site #5). The same trend was observed on cancellous bone, although the magnitude of stress was considerably lower than on cortical bone. The peak von Mises stress on cortical bone was 9.27 MPa, whereas for cancellous bone it reached 1.23 MPa (Figures 2 and 3).

Cantilever length

Increasing the cantilever length had a remarkable effect on stress concentration (Figure 4). At each increment of 5 mm, the maximum von Mises stress raised by approximately 30% to 37% on the cortical bone around the implants. The same trend was seen on cancellous bone, but on a smaller magnitude.

Quality of cancellous bone

As the elastic modulus of cancellous bone increased, more stress was observed, causing an opposite effect on cortical bone (Figure 5). Stresses increased by 75% in cancellous bone when the elastic modulus was changed from 1.5 to 4.0 GPa and by 45% when changed from from 4.0 to 7.9 GPa. In cortical bone, stresses decreased by 25% and 14% when the elastic modulus was changed from 1.5 to 4.0 GPa and from 4.0 to 7.9 GPa, respectively. Therefore, the stiffer the cancellous bone, the more stress it took, which lessened the stress on the cortical bone.

Abutment length

The stress in cortical and cancellous bone around the implants on sites #1 and #5 remained basically the same when the abutments were changed from 3.0 to 5.5 to 7.0 mm. The stress decreased on site #2 and increased by 50% to 100% on sites #3 and #4, when the abutments were changed from 3.0 mm to 5.5 mm to 7.0 mm (Figure 6).

Implant length

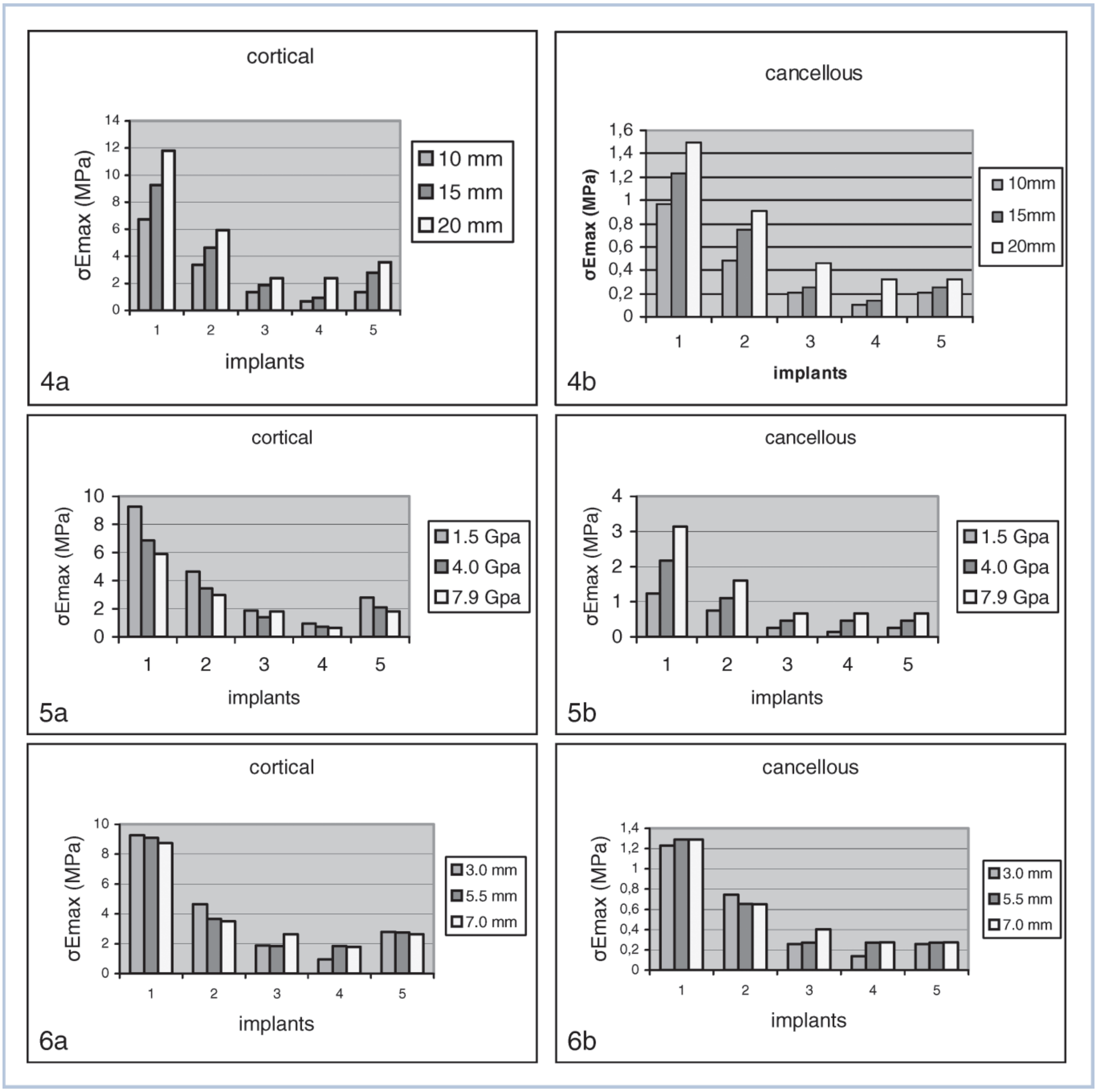
A slight decrease in stress around the implants was observed when the implant length was increased from 10.0 to 13.0 mm (10%) and from 13.0 to 15.0 mm (2%). This observation was valid for cortical and cancellous bone as well (Figure 7).

Framework alloy

Although not remarkably, the use of a CoCr alloy caused a slight decrease in the stress around the implants closest to the load. However, on sites #3 and #4, the stress on cancellous bone increased by up to 90% (Figure 8).

DISCUSSION

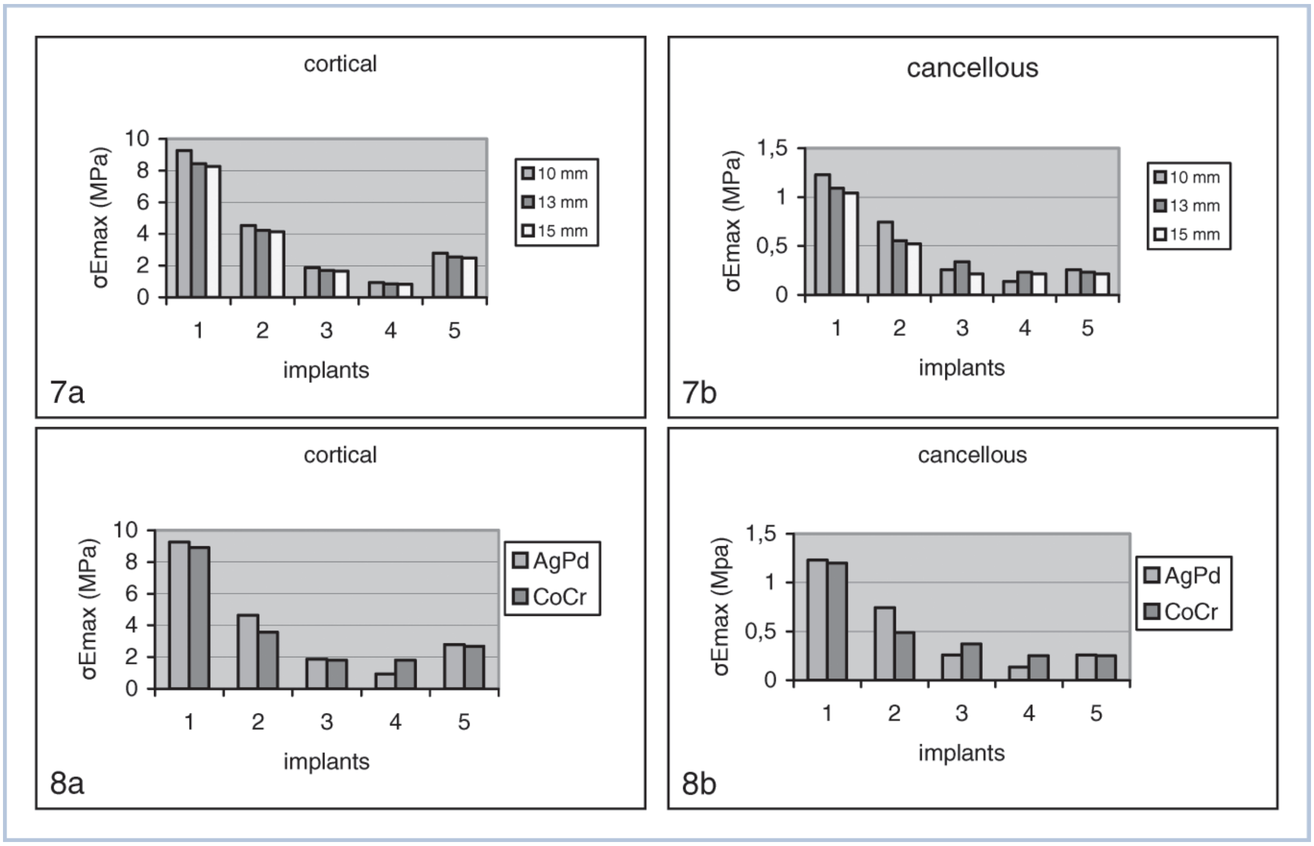
One of the most frequently asked questions about implant prosthesis is how much load implants can withstand without jeopardizing the surrounding bone.



FIGURES 4–6. FIGURE 4. Stress on cortical (a) and cancellous (b) bone when the force was applied at 10, 15, and 20 mm distally to the terminal implant. FIGURE 5. Stress on cortical (a) and cancellous (b) bone varying the modulus of elasticity of cancellous bone. FIGURE 6. Stress on cortical (a) and cancellous (b) bone when the abutment length is increased from 3.0 to 5.5 to 7.0 mm.

Although this question remains unanswered,[8,15,19–21] there is a general agreement that a well-planned and executed prosthesis is essential to avoiding excessive forces on bone and implant components. Implant dentistry would greatly profit if it were provided the means to predict how bone and implant components would behave considering each patient's unique jaw anatomy, quality of bone, amount of occlusal force exerted on the prosthesis, etc.

Finite-element analysis, with all its inherent limitations,[12,15] is a valuable instrument in pursuing that goal. When associated with clinical findings and accumulation of reliable data on implant loading, bone-implant contact area and other factors could



Figures 7 and 8. Figure 7. Stress on cortical (a) and cancellous (b) bone when the implant length is increased from 10 to 13 to 15 mm. Figure 8. Stress on cortical (a) and cancellous (b) bone with an AgPd and a CoCr alloy.

help us understand the problems encountered in daily practice.[19] For the reasons above mentioned, the results of this and other FEA studies have to be seen with a critical eye, and the values should not be taken as absolute but should rather be used to compare the possible magnitudes of stress that bone and implant components undergo during function.[5,15,22]

In trying to simulate a human mandible, it was found that a 3-D model not of the entire mandible but of the anterior segment (approximately 20 mm posterior to each foramen) would suffice. A previous study concluded that for comparison of the stress distribution around implants it is not necessary to build a model of the entire mandible. By doing so, one can have the advantage of reducing modeling and calculation time.[5]

The intention was to build the most versatile model possible to accommodate the largest number of variables with the least modifications. The model seen in Figure 1 was taken as a parameter for comparison. In every situation, the stress tended to be concentrated at the cortical bone on the disto-lingual aspect of the implant closest to the load. This is a common finding in the literature.[5,15,21,22] The magnitude of stress found on cortical bone was approximately ten times greater than that seen on cancellous bone. This can be explained by the higher modulus of elasticity of cortical bone, which therefore bears more stress.

Cantilever length

In this study, the effect of the cantilever length was studied by applying the load to 3 points distally to the terminal implant along the framework at 10.0, 15.0, and 20.0 mm. The conventional design of an implant-supported fixed bridge on the edentulous mandible results in bilaterally cantilevered framework extensions, which under load create torque and moment on the implants. Many suggestions have been made in the literature regarding the extension of the cantilever, but in general the various authors agree that according to the quality of bone, a range of 10 to 20 mm of cantilever extension is acceptable.[21,23–26] Perhaps one of the most compelling findings of this and other studies[17,21,27,28] is the fact that the increase in cantilever length has a remarkable impact upon the stress concentration around the implants. The longer the cantilever arm, the more stress was observed. Fortunately, this is one of the implant-supported prosthesis features most easily controlled by the dentist. The application of the shortened dental arch concept and the express recommendation to the dental technician to keep the length of the cantilever arm to a minimum are procedures that cannot be neglected.

Quality of cancellous bone

Since the elastic modulus of bone is related to its density, cancellous bone was assigned with 3 elastic modulli, 1.5, 4.0, and 7.9 GPa, to verify the influence on stress distribution.

Although the literature shows little variation in cortical bone density, many density ranges for cancellous bone have been found.[13,29,30] To guide the surgeon during placement of implants, attempts have been made to classify maxillary and mandibular bone according to the resistance to cut and the cortical:cancellous bone ratio.[31,32] The influence of such variation in cancellous bone density on stress distribution has not been the subject of many studies.[33]

Long-term clinical studies demonstrated that bone quality strongly correlates to implant success,[34–36] whereas micromotion has been regarded as disrupting of the healing process.[19] Once osseointegration is achieved and implants are loaded, there should be an effective load transfer from implant to bone. The load transfer is dependent upon the occlusal loads, implant shape and size, biomaterial properties, density of bone, and nature of the interface. Assuming complete contact between a cylindrical implant and bone, which does not necessarily correspond to the clinical situation, this study found that the stress on cortical bone decreased as the elastic modulus of the cancellous bone increased. The presence of a stiffer cancellous bone has the benefit of reducing the stress where it reaches its peak, namely the cervical area around the terminal abutment. At the same time, the stress in cancellous bone somehow increases, balancing the stress distribution. Of course, there is a limit to this benefit, because the more dense the cancellous bone, the more resistant it is to cutting and, therefore, the greater the heat generated during drilling.

Abutment length

Since patients will have different gingival thickness, the height of the abutments will vary accordingly.[37] Nevertheless, the higher the abutment, the more unfavorable is the "crown-to-root" ratio. The findings of this study have shown that varying the height of the abutments will have a different effect on bone around each of the 5 implants. Though on sites #1, #2, and #5 only small differences were noted, the stress increased by 50% to 100% on sites #3 and #4. Yet, given the low magnitude of the stress seen in those sites, it is possible that abutment length alone may not be a source of concern.

Implant length

The implant length selection is highly dependent upon the jaw anatomy and the need for implant stabilization. As the presence of micromotion early after implantation has been regarded as disrupting of the bone healing process,[19] the selection of an implant that reaches the basal compact layer of bone has been indicated.[38] Although there is indication for such procedure, this is seldom accomplished on the mandible zone I where the bone quality usually exempts the need for bicortical stabilization. The longest implant used in this study (15 mm) reached but did not engage the basal compact layer of bone. It has been demonstrated by clinical studies[34,39] that implant length does positively influence the outcome of implant therapy: longer implants are associated with higher success rates. This is particularly true when implants are placed in low-density bone, usually in the posterior maxillae (zone II). However, changing the implant length from 10 to 15 to 18 mm did not significantly affect the stress distribution in bone. Only a slight decrease was observed with longer implants. This trend has been observed before in 2-dimensional and 3-D FEA studies,[21,40] indicating that some factor other than stress distribution would be responsible for the better results found in clinical studies with longer implants.

Framework alloy

Frameworks for implant-supported prostheses have evolved from soldering a gold alloy framework to gold cylinders, to gold alloy frameworks cast directly onto the cylinders, to silver-palladium alloy frameworks.[26,39,41,42] The clinical results showed that the latter option gives consistently good results with respect to accuracy of fit, load-bearing capacity, reduced cost, and design versatility.[23] However, the use of an alloy with a higher modulus of elasticity, such as the CoCr alloy, would allow for a more evenly distributed load among implants with a less bulky framework, which would be an advantage if intraoral space is limited.[43–45] The structural rigidity of the prosthesis is dependent upon the elastic modulus, shape, and dimensions of the prosthesis, which can affect the force distribution among implants. The lower the elastic modulus, the greater the forces exerted on the abutments closest to the load. Therefore, if a hypothetical rubbery framework was used, the entire load would be concentrated on the implant closest to the loading point. The use of a CoCr alloy framework in this study favored a more even distribution of stress, as a slight increase was observed in cancellous bone around implants #3 and #4, which were opposite to the load. This finding is corroborated by others in the literature[4,27,45,46] who found that a stiffer framework, such as one made out of CoCr alloy, provides a more even distribution of forces among the bridge and abutments, decreasing the stress within the retaining screws as a result of the reduced bending of the framework.

Conclusions

Within the limits of the study, the following observations were found:

1. At each increment of 5 mm in cantilever length, stress increased by approximately 30% to 37% on the cortical bone around implants.
2. The stiffer the cancellous bone, the more stress it took and the less stress the cortical bone appeared to undergo.
3. A slight decrease in stress was observed with longer implants and abutments.
4. The use of a CoCr alloy contributed to a better stress distribution.

Acknowledgments

The authors wish to thank Professor George Zarb (University of Toronto) for providing the means to conduct this research and to Professor John Curran and his assistant Xiao Yang Lei (Mining Engineering, University of Toronto) for the FEA analysis. This study was supported by Fundação de Amparo a Pesquisa do Estado de São Paulo (FAPESP) and an Implant Prosthodontic Unit Research Grant – Faculty of Dentistry of Toronto.

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