Finite-Element Analysis of Stress on Dental Implant Prosthesis

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ABSTRACT

Background: 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.

Purpose: In this study, the many clinical variations present in implant-supported prosthesis were analyzed by 3-D finite element method.

Materials and Method: A geometrical model representing the anterior segment of a human mandible treated with 5 implants supporting a framework 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 100N vertical load was used to simulate the occlusal force. Images with the fringes of stress were obtained and the maximum stress at each site was plotted in graphs for comparison.

Results: Stresses clustered at the elements closest to the loading point. Stress increase was found to be proportional to the increase in cantilever length and inversely proportional to the increase in the elastic modulus of cancellous bone. Increasing the abutment length resulted in a decrease of stress on implants and framework. Stress decrease could not be demonstrated with implants longer than 13 mm. A stiffer framework may allow better stress distribution.

Conclusion: The relative physical properties of the many materials involved in an implant-supported prosthesis system affect the way stresses are distributed.

KEY WORDS: finite-element analysis, implant prosthesis, stress distribution

INTRODUCTION

The peculiar characteristic of implant-supported prosthesis is the fact that its fixation is given by the connection of an alloplastic material (the implant) to a living tissue (the bone). This fixation has been defined as rigid and clinically assymptomatic and must be maintained during functional loading.[1] Under load, bone tissue will undergo a remodeling process, which ultimately influences the long-term function of a dental implant system.[2] Bone remodeling is a complex process that involves a sequence of chemical- and mechanical-mediated biologic events known as *mechanotransduction*.[3]

Because 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] It is commonly found in the literature that, for cantilevered implant-supported mandibular prosthesis, stresses tend to be concentrated at the cortical bone on the disto-lingual aspect of the implant closest to the load.[2,5–8] Many researchers have focused on the steps of force transfer to gain insight into the biomechanical effect of force directions, force magnitudes, prosthesis type, prosthesis material, implant design, number and distribution of supporting implants, bone density, and the mechanical properties of the bone-implant interface.[9] The resultant stresses must be kept below the failure stress of the materials involved.[4,10,11]

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TABLE 1 Elastic Properties of Materials Used for FEA Model

Material	E (GPa)	μ	Reference
Titanium	110	0.35	12
AgPd alloy	95	0.33	13
CoCr alloy	218	0.33	13
Cortical bone	13.7	0.30	2, 14
Cancellous bone	1.5 (d = 25%)	0.30	14
	4.0 (d = 50%)	0.30	14
	7.9 (d = 75%)	0.30	14

E = elastic modulus; FEA = finite-element analysis; μ = Poisson's ratio; d = density.

In a previous study,[8] the stresses generated at the bone level were assessed by means of finite-element analysis (FEA). The stresses on implants, abutments, and framework are as well important to evaluate because these structures, being the stiffest components of an implant prosthodontic system, bear a great amount of stress and are responsible for transmitting the load to the bone.

MATERIAL AND METHODS

A computerized three-dimensional finite-element model of the anterior segment of a human mandible provided with an implant-supported bridge was created. The basic model consisted of a curved beam with a radius of 15.0 mm and is 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 × 18.0 × 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 and were numbered 1 to 5 clockwise for identification. A second beam (71.0 × 4.0 × 6.0 mm) was added in connection to the abutments to simulate a framework. This resulted in a model with 1, 714 nodes and 8, 062 elements. The model was restrained 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, OH, USA) installed in a desktop computer was used to analyze 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. Images of the stress fields were obtained for each of the components of the model, and maximum von Mises stress (σEmax) at each site was plotted in graphs for comparison.

Five clinical variables were chosen to be evaluated and are summarized in Table 2 for easy reference. Each variable was introduced alternately on the basic model, all other conditions being equal. Cantilever length was defined as the position where load was applied relative to the center of the terminal abutment. Quality of 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 Brånemark system (Nobel Biocare AB, Göteborg, Sweden). Framework alloy was selected according to the modulus of elasticity of AgPd (E = 95 GPa) and CoCr (E = 218 GPa) alloys.

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 = elastic modulus; FEA = finite-element analysis.

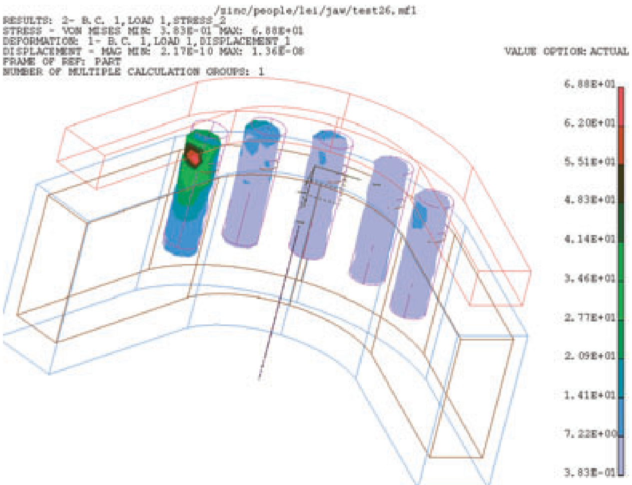

Figure 1 Stresses on implants/abutments (lingual view).

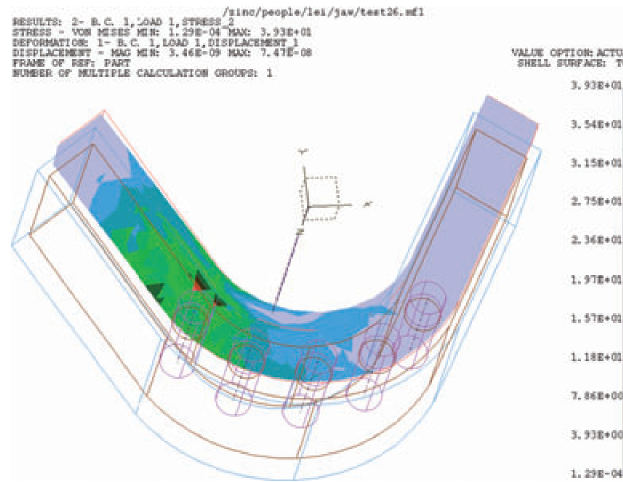

Figure 2 Stresses on the framework (bottom view).

RESULTS

Lingual views of the model were captured showing the areas of abutment/implant under higher stress levels. Images of the framework were analyzed from a bottom view because maximum von Mises stresses were observed in this area. In the basic model, stresses tended to be concentrated along the disto-lingual aspect of the terminal implant and abutment closest to the loading point (site #1). Stresses gradually decreased as moving away from the load and rose again on the opposite side (site #5). Maximum stresses on implants/abutments reached 68.8 MPa on site #1 and 39.3 MPa in the framework (Figures 1 and 2).

1. Cantilever Length

Stresses at the implant-abutment interface (site #1) raised by 45% when the cantilever length was increased from 10 to 15 mm and by 30% from 15 to 20 mm, almost twice the stress seen with a 10 mm cantilever. This trend was also observed in the other sites. Increasing the cantilever length from 10 to 15 mm also resulted in significant stress increase in the framework (Figure 3). Nevertheless, when the cantilever was increased from 15 to 20 mm, there was a decrease in stress on sites #1, #2, and #3 and an increase on sites #4 and #5.

2. Quality of Cancellous Bone

The stiffer the cancellous bone, the less stress was observed in the framework (Figure 4). On the other hand, varying the elastic modulus of cancellous bone had no effect in the stress observed on abutments/implants, which was kept at the same level as the stress observed in the basic model (σEmax = 68 MPa) regardless of the elastic modulus used.

3. Abutment Length

Stresses on implants and framework decreased by 20 to 35% on site #1, while the stress increased between 10 and 30% on site #5 (Figure 5).

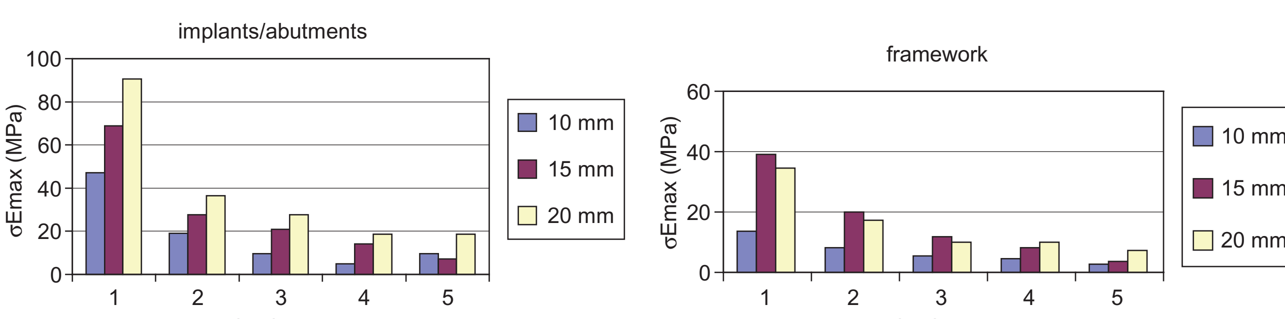

Figure 3 Stresses on implants/abutments (*left*) and framework (*right*) with varying cantilever lengths.

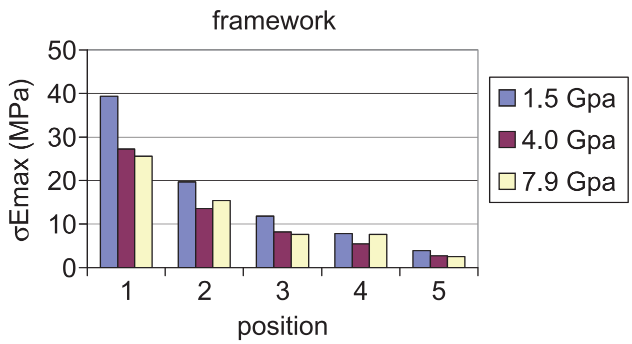

Figure 4 Stresses in the framework varying the elastic modulus of cancellous bone.

4. Implant Length

Stresses on implants and abutments decreased by approximately 14% when implant length increased from 10 to 13 mm and then stabilized at this level at the 13 to 15 mm change. The framework presented 30% less stress when changing implant length from 10 to 13 mm, remaining stable at the 13 to 15 mm change (Figure 6).

5. Framework Alloy

Changing the framework alloy from AgPd to CoCr resulted in a decrease of the stress seen on abutments/implants of approximately 20% on sites #1 to #4 and had an accentuated raise on site #5 (135%). The CoCr framework presented a generalized decrease in stress as compared with the AgPd framework, especially on site #1 (Figure 7).

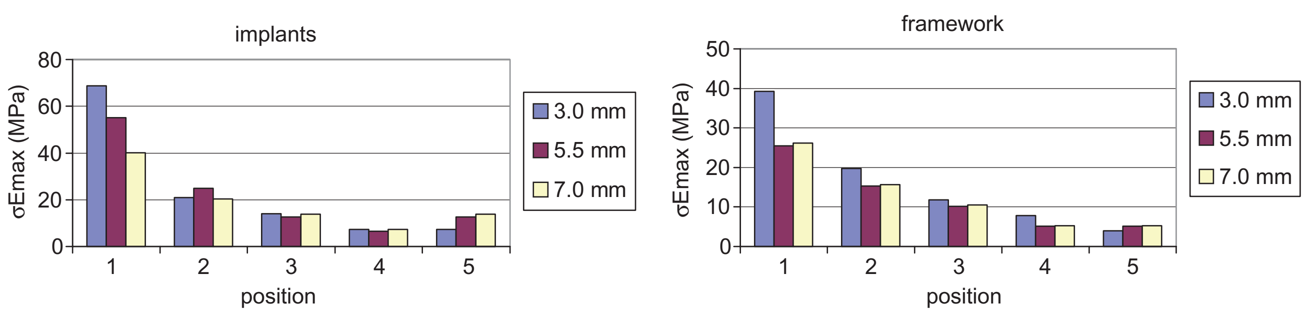

Figure 5 Stresses on implants and framework when abutment length was changed from 3.0 to 5.5 and 7.0 mm.

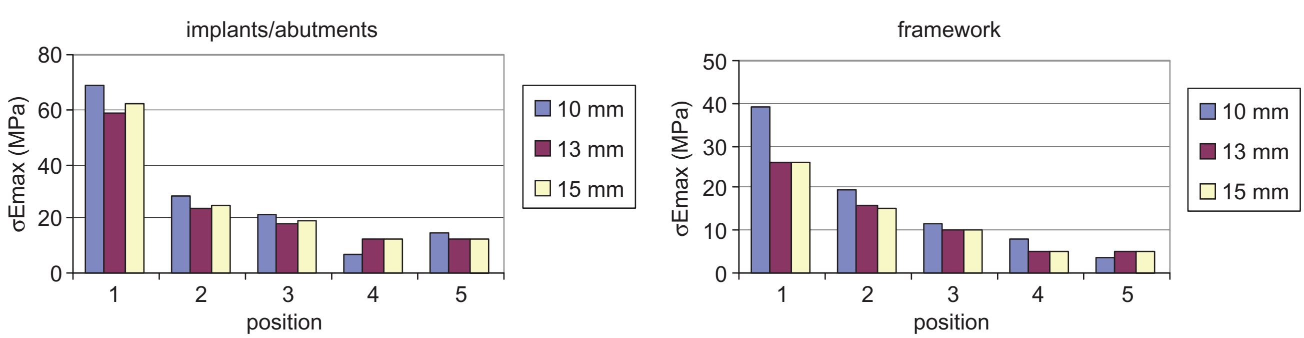

Figure 6 Stresses on implants/abutments and framework when implant length was changed from 10 to 13 and 15 mm.

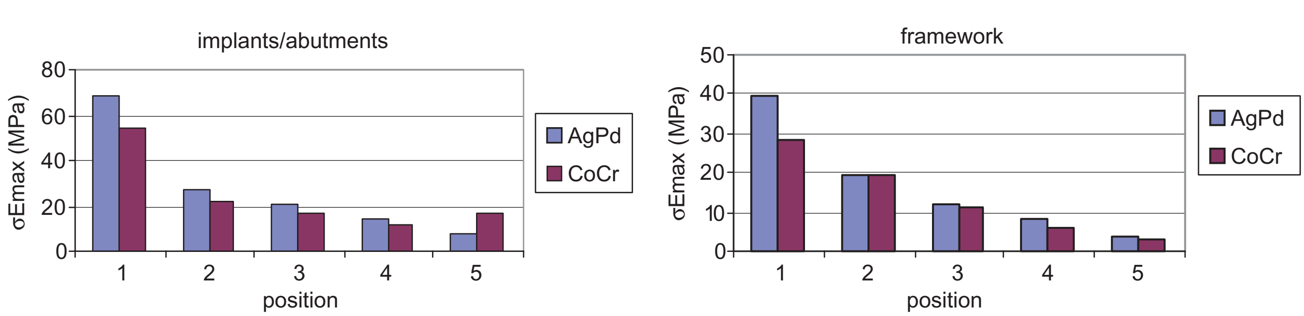

Figure 7 Stresses on implants/abutments and frameworks made with AgPd and CoCr alloys.

DISCUSSION

A well-planned and well-executed prosthesis is essential to avoid excessive and unnecessary forces on bone and implant components. Predicting how bone and implant components would behave, considering each patient's unique jaw anatomy, quality of bone, and amount of occlusal force exerted on the prosthesis, demands full comprehension of both mechanical and biologic events. Finite-element analysis, with all its inherent limitations,[14,15] is a valuable instrument in pursuing that goal. When associated to clinical findings and accumulation of reliable data on implant loading, bone-implant contact area, and other factors, it could help us understand the problems encountered in daily practice.[16] For the reasons mentioned above, 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 as a comparison of the possible magnitudes of stress bone and implant components undergo during function.[2,5,7]

In order to expedite the analysis and to make a model more versatile to the introduction of variables, a series of simplifications were performed to represent the clinical situation. These simplifications have been tested before and resulted in reduced modeling and calculation time with negligible difference to a more complex model.[2] Also, perfect passivity between the components was assumed to avoid the appearance of internal tensions that could confound the analysis. Such tensions can substantially increase the risk of failure even without external loads. A perfect fit between the components combined to the framework rigidity is therefore essential to the longevity of the prosthesis.[17,18]

1. Cantilever Length

In this study, the effect of cantilever length was evaluated by applying the load on three 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.[6,17,19–21]

Previous studies have demonstrated that the increase in cantilever length is directly proportional to the increase in stress concentration around the implants. Kunavisarut and colleagues[22] observed that the presence of a cantilever arm significantly increased the stress in the prosthesis, implant, and surrounding bone. Besides, when no proper fit is achieved, the stress is magnified by the cantilever. The increase in the cantilever arm resulting in more stress has also been demonstrated when abutment height is increased.[23] The effect of increasing the cantilever arm in this study was remarkable. Stress in the abutment/implant almost doubled when the force was moved from 10 to 20 mm along the cantilever. When analyzing the effect on the framework, a different pattern is observed. The stress increases from 10 to 15 mm but then decreases at the 15 to 20 mm change. According to Benzing and colleagues,[24] the load application in a framework for implant prosthesis produces deformation energy in the system that causes flexion. If a great amount of deformation energy is consumed by the framework, reduction of the transmitted energy happens, decreasing the stress in this structure.

2. Quality of Cancellous Bone

Long-term clinical studies demonstrated that bone quality strongly correlates to implant success,[25–27] while micromotion has been regarded as disrupting of the healing process.[16] Once osseointegration is achieved and implants are loaded, there should be an effective load transfer from implant to the bone. Load transfer is dependent upon occlusal loads, implant shape and size, biomaterial properties, bone density, and nature of the interface. The denser the cancellous bone, the more stress it bears, and less stress is expected to be seen in other structures.[8] Nevertheless, the stresses on implants/abutments remained unchanged, suggesting that the bone density increase was not significant to cause any effect in those structures. On the other hand, the stress in the framework decreased with denser cancellous bone. This behavior could be the result of the relative position of implants/abutments and framework to the underlying bone: the framework, being more distant to the bone and therefore presenting more leverage, was more affected by variations in bone elastic modulus.

3. Abutment Length

The findings of this FEA study show that varying the abutments height from 3.0 to 5.5 and to 7.0 mm resulted in a decrease of stress on implants and framework. On implants, the stress decrease seems to be the result of the dislocation of the point of maximal stress to a more coronal area of the abutment, away from the implant. A greater concentration of stress happens in the abutment with a consequent stress decrease in the framework. Because the stress increases in the abutments as they become higher, it could be hypothesized that a larger incidence of abutment screw failures would occur, but there is no clinical evidence to support this theory. Therefore, the structural strength of the abutments seems to be adequate to bear the stress increase resultant of the longer lever arm.

4. Implant Length

It has been demonstrated by clinical studies[25,28] that implant length does influence positively the outcome of implant therapy; longer implants are associated to higher success rates. The longest implant used in this study (15 mm) reached but did not engage the basal compact layer of bone. Although there is an indication for such procedure, this is seldom accomplished on the mandible zone I where the bone quality usually exempts the need for bicortical stabilization. Changing the implant length from 10 to 13 to 15 mm did not significantly affect the stress in these structures. Only a slight decrease was observed with longer implants, which is in agreement with other two-dimensional and three-dimensional FEA studies.[6,29] Factors other than stress distribution may be responsible for the better results seen in clinical studies with longer implants.

The study of Tada and colleagues[30] presented only slightly differences between the maximum equivalent stress/strains around screw- and cylinder-type implants. Therefore, as in the work of Kitamura and colleagues,[31] the implants and abutments were simplified to a cylinder for modeling convenience. Besides, a perfect bond between bone and implant was assumed. The results of Siegele and Soltez[32] showed that implant shape and interface type (perfect bond vs. no bond) significantly affect the stress fields. For the cylinder type with no bond, stresses concentrate near the apex, while, for perfect bond, near the bone crest. For a screw-shaped implant, there is stress concentration near each screw tread in both interface conditions. Assuming complete osseointegration and bone quality 1 and 2, Tada and colleagues[30] and Clelland and colleagues[33] observed no difference between cylindrical and screw-shaped implants. Therefore, the simplification of the geometric form of the implants neared the results that would be observed with screw-shaped implants as a result of the bone-implant interface adopted. In this study, the increase in implant length led to a decrease in framework stress (see Figure 6). This has been seen at the 10 to 15 mm change, but, at the 15 to 20 mm change, implant length had no influence in framework stress. A commom finding in FEA studies is that bone properties and interface condition have greater influence in stress state than implant variability.[34] Another relevant clinical fact is that the choice of implant length is a function of bone ridge resorption. In this regard, a clinical study, as well as a FEA analysis, agrees that, as greater the bone ridge resorption, as larger the likelihood of failures by tension accumulation.[29,26]

5. Framework Alloy

Structural strength is dependent on the elastic modulus, shape, and length of the prosthesis, which may affect the load distribution among the implants. The lower the elastic modulus, the greater the force applied to the abutment/implant closest to the load. Therefore, if an imaginary rubbery framework was used, nearly the totality of the load would be concentrated at the implant/abutment closest to the point of load application.[18,35] Likewise, an infinitely rigid framework would distribute the load equally among the implants. Brunski[34] compared two implant prosthesis frameworks of considerably different elastic moduli, one made out of acrylic resin and one made out of AgPd alloy, confirming the theory that most of the load concentrates on the terminal implant closest to the load. This finding opposes the original theory proposed by Skalak,[18] who underestimated the load on the implant closest to the load and super estimated the load on the implants on the opposite side, exactly because disregarded the mechanical properties of the structures.

Besides the materials properties, shape also influences tension. The influence of the cross section and materials properties on the flexural strength of a beam can be calculated by[36]

$$K = -M/E \times I \tag{1}$$

The equation demonstrates that the deformation of the long axis of a beam (K) is directly proportional to the torsion moment (M) and inversely proportional to the product of the elastic modulus by the cross section, ie, the structural rigidity of the beam ($E \times I$). By increasing the elastic modulus of the framework or its cross-sectional area, deformation decreases, and decreasing the torsion moment, deformation decrease also takes place. The law of beams establishes that multiplying the width by two doubles its rigidity, while multiplying the height by two, rigidity increases eight times. The cross-sectional area of the framework in this study was 24 mm^2. Obviously, alterations in shape and position of its long axis would result in different stress distribution.[36]

The use of a CoCr alloy allowed a better stress distribution. There was an increase of the stress in the abutment on site #5 (opposite to the load) of the order of 135%, with a decrease of the stress in the remaining positions (see Figure 7). This finding is corroborated by others in the literature,[4,24,37,38] who found that a stiffer framework provides a more even distribution of forces among the abutments, decreasing the stress within the retaining screws, as a result of the reduced bending of the framework. Left aside the difficulties in the use of CoCr alloys, like casting shrinkage (around 2.3%) and melting point differences relative to gold cylinders, its use has been clinically tested with no complications that could be regarded to the alloy itself.[38] The analysis of the stress in the framework with different cantilever length revealed an interesting fact relative to the AgPd alloy: the increase in cantilever length is not followed by a proportional increase in stress levels of the framework (see Figure 7). The stress in the framework increases when the cantilever changes from 10 to 15 mm, but this effect does not repeat at the 15 to 20 mm change. In fact, there is a decrease in the stress levels, denoting that, with longer cantilevers, the framework begins to bend, yielding to the load. Benzing and colleagues[24] observed that the resistance to the deformation of a golden framework was two-thirds of that of a nonprecious alloy framework. Therefore, it can be suggested that, in cases of long cantilevers, there would be a benefit in the use of a more rigid framework. Nevertheless, excessive long cantilevers must be avoided even with frameworks of high elastic modulus. Young and colleagues[39] observed deformation in CoCr frameworks with 26 mm of cantilever and loads of only 130 to 140 N.

CONCLUSIONS

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

1. The increase in cantilever length is proportional to the increase in stress concentration.
2. The increase of the elastic modulus of cancellous bone had little effect in the stress in abutments/implants and was inversely proportional to the stress in the framework.
3. Increasing the abutment length (increases the lever arm) resulted in a decrease of the stress on implants and framework.
4. Increasing implant length from 10 to 13 mm resulted in less stress in the framework, but the same has not been demonstrated at the 13 to 15 mm change.
5. The more rigid the framework is, the better the distribution of stress among the abutments/implants, and less stress is seen in the framework.
6. The relative physical properties of the materials substantially affect the way stresses are distributed.

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