Influence of Prosthesis Material on Stress Distribution in Bone and Implant: A 3-Dimensional Finite Element Analysis

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A 3-dimensional finite element analysis was conducted to assess stress distribution in bone, implant, and abutment when gold alloy, porcelain, or resin (acrylic or composite) was used for a 3-unit prosthesis. A unit force was applied axially and then buccolingually to the center of the pontic. For gold and porcelain, similar maximum equivalent stress was found in each part of the models. In almost all cases, stress in the model with the resin prostheses was similar to or higher than that in the models with the other 2 prosthesis materials. The highest increase in stress with the resins was found in the implant-abutment unit under axial load. The protective role of resin for the implant-bone interface could not be demonstrated under the conditions of this analysis.
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Key words: dental implants, fixed partial prostheses, prosthesis material, stress, 3-dimensional finite element analysis

Recently, treatment with implant-supported fixed partial prostheses has been established as an option for partially edentulous patients. Clinical studies conducted for various implant types have reported high rates of success and survival, but also variable rates of implant failures.[1–5] The most frequently cited reasons for implant failure are poor oral hygiene and biomechanical factors.[2,5–10] Infection and traumatic occlusion were confirmed as causes of implant failure by histopathologic and microbiologic studies.[11–13] Animal experiments and clinical studies have shown that bone loss around implants, in the absence of plaque-related gingivitis, was associated with unfavorable loading conditions.[1,7,10,14–16]

Because of intimate contact at the bone-implant interface, loads applied to the implant will be directly transmitted to the bone, and the bone's biologic reaction is linked with implant longevity.[17–20] Although a minimum amount of stress is considered to be necessary for bone remodeling,[21,22] very high amounts of stress could lead to microdamage and induce resorptive modeling and mechanical failure by exceeding the limits that the bone can tolerate.[19,22,23]

Implants themselves have to withstand stress induced by intraoral forces. Increased or abnormal loading, as well as fatigue under physiologic loads, could lead to fractures of the implant system components.[5,9,24,25]

Theoretical considerations[23] and an in vitro experiment[26] suggest that, under conditions of impact loading, acrylic resin occlusal surfaces of the prosthesis will protect the connection between implant and bone. However, clinically, when acrylic or composite resin is used on the occlusal surface, many complications are reported during the follow-up period of implants, including resin fracture, esthetic defects, occlusal screw loosening or fracture, abutment screw and implant fractures, and resin wear.[5,6,9,27,28] On the other hand, when porcelain is used instead of resin,

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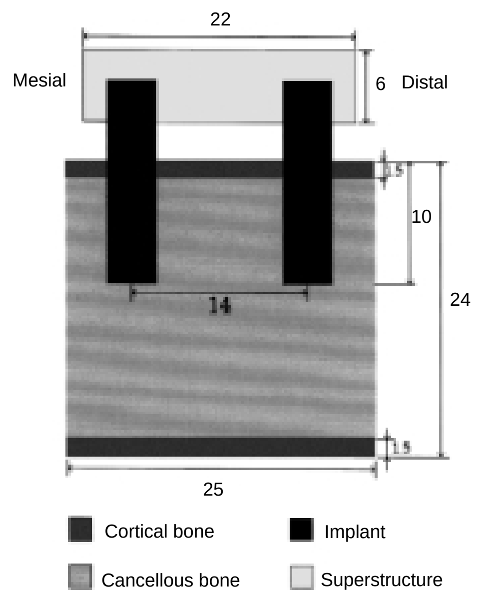
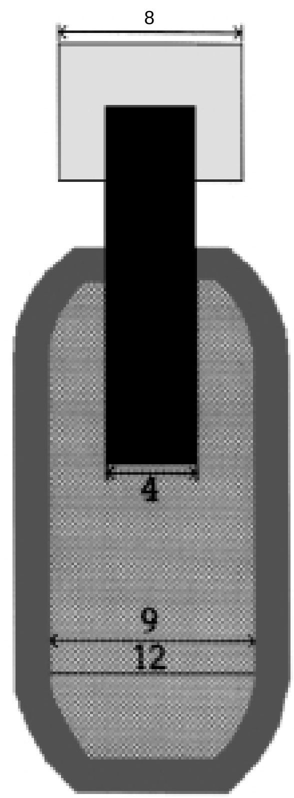




Figs 1a and 1b Mesiodistal *(above)* and buccolingual *(right)* views of the 3-dimensional finite element model. Two implants placed in a cancellous bone core surrounded by cortical bone support a 3-unit fixed partial prosthesis. All measurements in millimeters.

better esthetics, far fewer prosthetic complications,[1,28] and no statistically significant difference in marginal bone loss are found.[1] High resistance to fracture of metal-ceramic restorations has also been demonstrated in a mechanical failure test.[29]

The influence of different prosthesis materials on the stress in bone and implant components was also investigated by means of finite element analysis. Ismail et al[30] analyzed the influence of the occlusal material (porcelain, precious and non-precious alloy, acrylic or composite resin) on the stress in bone and implant, and they reported similar results for all the investigated materials. In models of single implant–supported prostheses[31] and implant-supported complete arch prostheses,[32] occlusal material did not influence bone stress, but in the model of the implant-supported complete arch prosthesis, it did influence retaining screw stress.[32,33] For fixed partial dentures, such data are lacking.

The present study used a 3-dimensional finite element analysis (3-D FEA) to investigate the stress generated in both bone and implant-abutment units when different materials were used for a 3-unit prosthesis supported at both ends by implants.

Materials and Methods

Model Design. A mandibular segment containing 2 implant-abutment units and a fixed prosthesis were modeled on a PC-H98 model V105 computer (NEC, Tokyo, Japan) using finite element software (ANSYS 5.0, Swanson Analysis System, Houston, PA). The bone was modeled as a cancellous core surrounded by a 1.5-mm cortical layer. The mesial and distal section planes were not covered by cortical bone (Figs 1a and 1b).

Two titanium implant-abutment units were modeled using a solid cylinder 16.5 mm long and 4 mm in diameter. Each implant-abutment unit was designed with 10 mm of embedment depth, 3 mm of neck, and a 3.5-mm abutment. These dimensions corresponded to the height of the middle-sized ITI implant and abutment, respectively,[34] but the shape was simplified to a cylinder. To simulate a fixed prosthesis, a superstructure was overlapped over the titanium abutments. This superstructure was in the shape of a block 22 mm long, 8 mm wide, and 6 mm high (Figs 1a and 1b). These dimensions were chosen to roughly correspond to the size of the posterior teeth, which were replaced by the implant-supported prosthesis.[35,36] A symmetrical model was designed to reduce the factors that could influence the outcome of the stress analysis. For the same reason, the geometry of the prosthesis was simplified to a block shape. The stiffness of the prosthesis was varied by changing the material from gold alloy to porcelain or to resin (acrylic and composite).

Material Properties. All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. Since in many cases the literature provides different values for the elastic properties of the same material, average values were chosen for this study. These values and the references consulted are listed in Table 1.

Interface Condition. To simulate ideal osseointegration, the implants were rigidly anchored in the bone model along their entire interface. The same type of contact was provided at the abutment-prosthesis interface.

Elements and Nodes. Because of its mesiodistal symmetry, only half of the model was meshed with 8-node hexahedron elements (Fig 2). A finer mesh was generated around the implant. Altogether, 3,328 elements and 3,846 nodes were created.

Constraints. Models were constrained in all directions at the nodes on the inferior border of the bone surface on one-fifth of the bone height. Since only half of the model was meshed, symmetry boundary conditions were prescribed at the nodes on the symmetry plane.

Loads. Unit static loads (1 N) were applied axially (AX) and buccolingually (BL), separately, to the occlusal key point corresponding to the center of the pontic (Fig 2). Since symmetry boundary conditions were prescribed to the model, each input load was doubled during the computations.

Solution. The analysis was performed for each load by means of the ANSYS software program, which was run on the aforementioned personal computer. The Von Mises stress (equivalent stress [EQV]) was used to display the stress in the bone and implant-abutment unit.

Table 1 Elastic Properties Ascribed to the Materials in the Models

Material	Young's modulus (GPa)	Poisson ratio
Cortical bone[23,37–40]	15	0.3
Cancellous bone[37,41]	1.5	0.3
Titanium[37,38,42]	110	0.35
Gold alloy[37,43–45]	90	0.3
Porcelain[46]	70	0.19
Acrylic resin[45,47–49]	2.4	0.35
Composite resin[48,50,51]	16.6	0.24

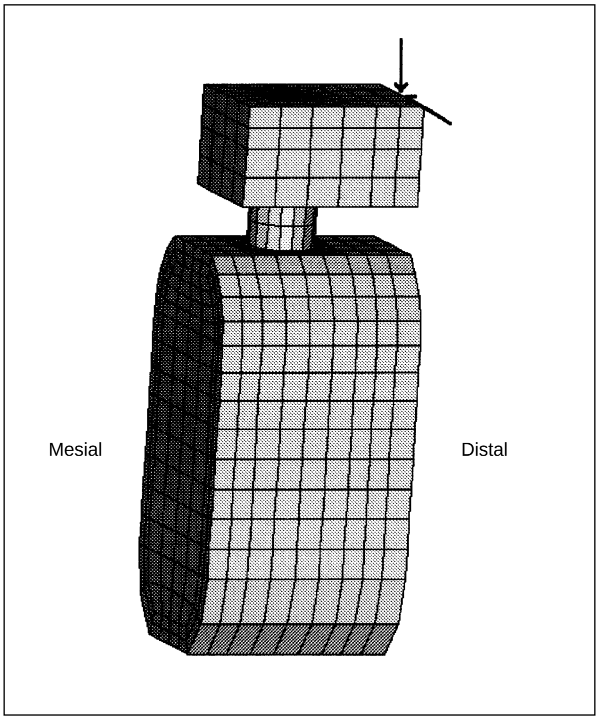


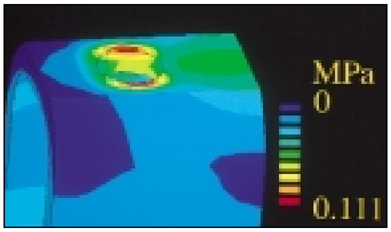
Fig 2 Half of the model was meshed and unit forces (1 N) were separately applied axially and buccolingually to the center of the pontic.

Results

Since only slight differences were found in the bone stress distributions with the 4 prosthesis materials, only the results from the acrylic resin prosthesis are presented (Figs 3 to 6).

EQV distribution in cortical bone is shown in Figs 3 and 4 for AX and BL loads, respectively. For convenience in presentation, the meshed half of the model corresponds to the mesial half when viewed anteroposteriorly. Regardless of prosthesis material and load direction, the highest stress in the cortical bone was located buccally and lingually around the implant neck. Under AX load, moderate stress was also found between the implants in the resin models (the green area on the crest, in Fig 3), but not in the porcelain and gold models.

Figures 5 and 6 display the EQV in a buccolingual section of the cancellous bone under AX and BL loads, respectively. Stress distribution in cancellous bone was significantly influenced by load direction, but it was not affected by prosthesis material. Under AX load, the highest stress was concentrated around the apical one-third of the implant and extended vertically beyond the end of the implant (Fig 5). Under BL load, the highest stress was concentrated buccally and lingually below the implant neck (Fig 6).



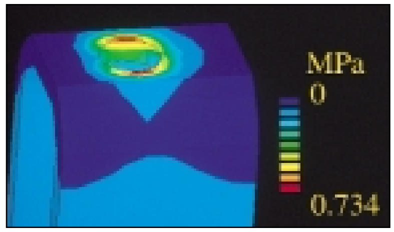
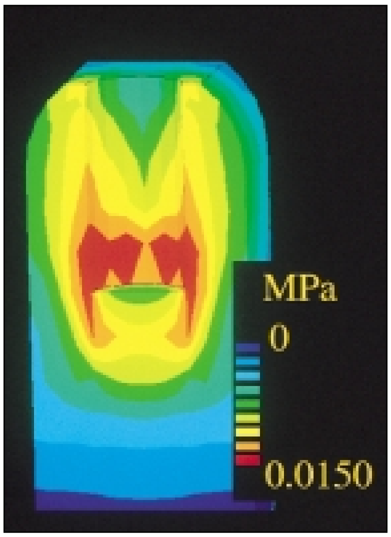


Fig 3 Close-up view of equivalent stress distribution in cortical bone in the acrylic resin prosthesis model under axial load.

Fig 4 Close-up view of equivalent stress distribution in cortical bone in the acrylic resin prosthesis model under buccolingual load.



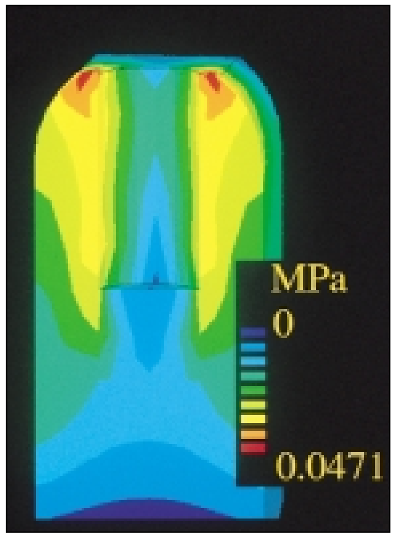


Fig 5 Distal view of a buccolingual section showing equivalent stress distribution in cancellous bone in the acrylic resin prosthesis model under axial load.

Fig 6 Distal view of a buccolingual section showing equivalent stress distribution in cancellous bone in the acrylic resin prosthesis model under buccolingual load.

Although the areas of high stress shown appear larger under AX load than under BL load, the values were much higher in the latter.

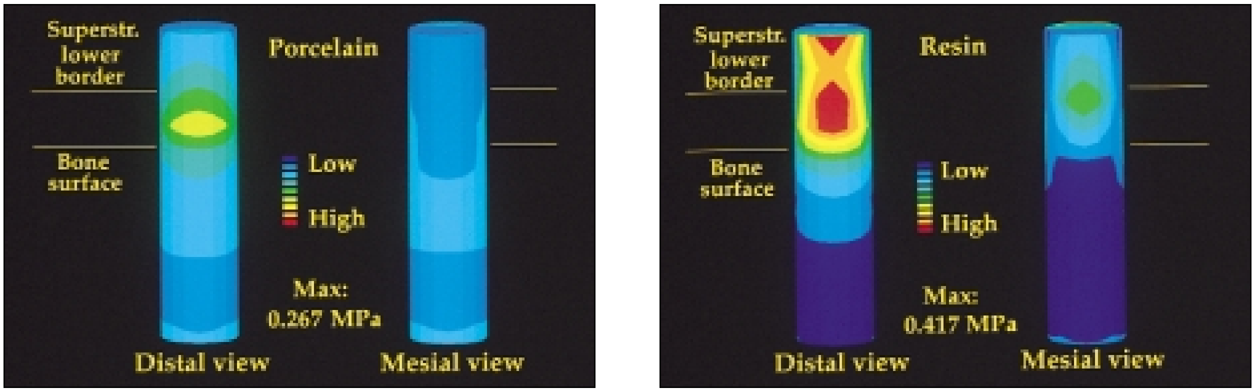
In the implant-abutment unit, stress distribution in the gold alloy prosthesis model was similar to that in the porcelain restoration model, but it was different from the stress distribution in the resin prosthesis model. Since the stress pattern in the composite resin model fell between those in the porcelain and acrylic resin models, only the results for the porcelain and the acrylic resin are presented in Figs 7 and 8.

Under AX load, stress was concentrated on the pontic side (Figs 7a and 7b) of the models. The highest stress in the porcelain prosthesis model was concentrated in a relatively small area which was located between the cortical bone surface and the lower border of the superstructure (Fig 7a). In the acrylic resin prosthesis this stress increased greatly in magnitude and extended over a very large area from the bone surface to the top of the abutment (Fig 7b). The area of highest stress in the composite resin prosthesis model was larger than in the porcelain model, but it did not extend to the abutment, as it did in the acrylic resin model. Under BL load, the highest stress was concentrated buccally and lingually, at the bone surface and above (Figs 8a and 8b). Although this stress was slightly lower in the resin models than in the porcelain and gold alloy models, it concentrated over a more extended area.

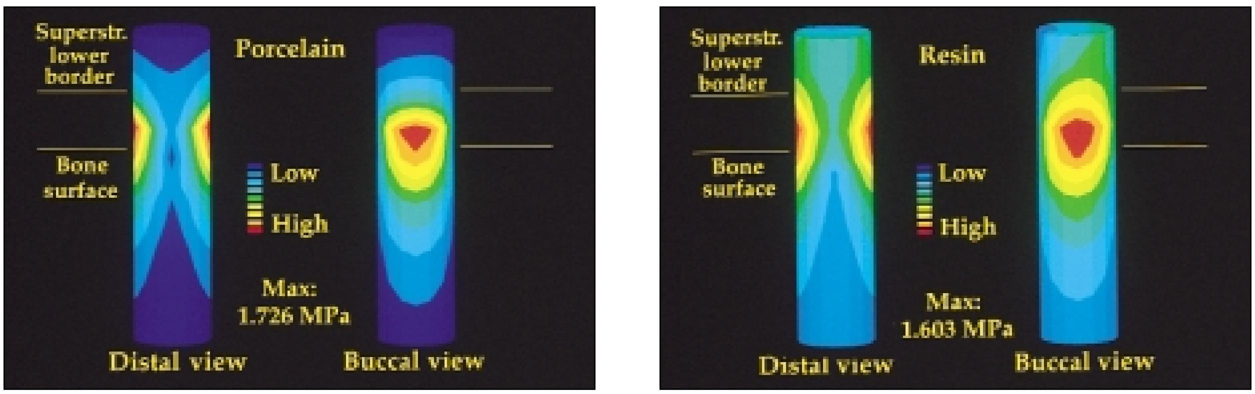
Maximum equivalent stress (max EQV) in cortical bone, cancellous bone, and implant-abutment units is shown in Tables 2 and 3 and in Fig 9, respectively. Each of these stresses was similar in the gold alloy and porcelain prostheses. Stress in every part of the models was much higher under BL load than under AX load.

Max EQV in cortical bone is detailed in Table 2. Under AX load, the lowest max EQV was found in the gold alloy and the porcelain models. In the composite resin model, max EQV was slightly higher, and in the acrylic resin model, it was about 10% higher than the max EQV in the gold prosthesis. Under BL load, max EQV was similar for all prosthesis materials.

For each loading condition, max EQV in cancellous bone (Table 3) was almost identical regardless of the prosthesis material. The stresses between max EQV in cancellous bone and that in cortical bone were 1:7 and 1:15 under AX and BL load, respectively.



Figs 7a and 7b Equivalent stress distribution in the implant-abutment unit under axial load *(left)* in the porcelain prosthesis model and *(right)* in the acrylic resin prosthesis model. Higher stress was concentrated over larger areas in the acrylic resin prosthesis model.



Figs 8a and 8b Equivalent stress distribution in the implant-abutment unit under buccolingual load *(left)* in the porcelain prosthesis model and *(right)* in the acrylic resin prosthesis model. Slightly higher stress was observed in the model with the porcelain prosthesis, but the areas of high stress were larger in the model with the resin prosthesis.

Table 2 Maximum Equivalent Stress (in MPa) in Cortical Bone

Prosthesis material	Axial load	Buccolingual load
Acrylic resin	0.111	0.734
Composite resin	0.104	0.728
Porcelain	0.100	0.729
Gold alloy	0.100	0.729

Table 3 Maximum Equivalent Stress (in MPa) in Cancellous Bone

Prosthesis material	Axial load	Buccolingual load
Acrylic resin	0.0151	0.0471
Composite resin	0.0149	0.0468
Porcelain	0.0149	0.0468
Gold alloy	0.0148	0.0468

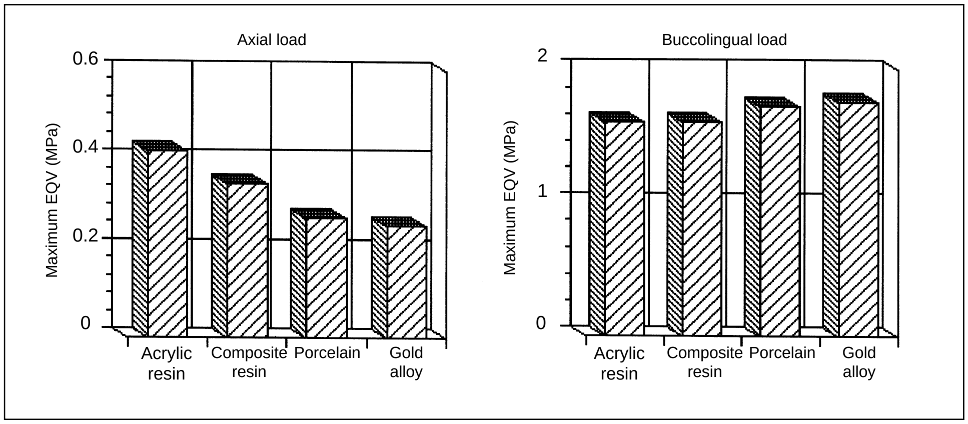


Fig 9 Maximum equivalent stress (max EQV) in the implant-abutment unit in all the models under *(left)* axial (AX) and *(right)* buccolingual (BL) load.

Max EQV reached in the implant-abutment unit is shown in Fig 9. Under AX load, max EQV in the acrylic resin prosthesis model was about 65% higher than in the gold alloy model; in the composite resin prosthesis model, max EQV was about 36% higher than in the gold alloy model. Under BL load, max EQV decreased by about 8% with the resin prostheses. Only in this case max EQV in the resin prosthesis was slightly lower than in the other two prostheses.

Discussion

This study used the 3-D FEA method to investigate the influence of different prosthesis materials on the stress in bone and implant-abutment units in a mandibular posterior segment model.

Various studies have reported similar results using the FEA and other experimental methods, such as in vivo strain-gauge measurements,[37] histologic studies,[14,21] and in vitro experiments.[52] FEA was considered to be an appropriate method for internal stress investigation and was used in the present study.

There are 2 types of FEA: static analyses and dynamic analyses. The estimates for the maximum closure speed of the mandible relative to the maxilla vary depending on the methods used for its measurements between 85 and 140 mm/s.[19,53,54] If higher mandibular velocities are involved, for example, during inadvertent biting of a hard object, a dynamic analysis may be required. A static analysis is considered suitable to simulate clenching, grinding, and most mastication conditions. Since bruxism is reported to be one of the main factors that can potentially damage bone and implants,[7,8,15,24] static loads were considered to be sufficient for the purpose of this study.

In each situation investigated, the present study focused on the locations and values of the highest stress—those that could put bone, implant, and abutment at risk. Thus the Von Mises stress, which estimates quantitatively the stress of a point in nonuniaxial stress state, was chosen to display the results of the computations.

Since all materials were considered to be linearly elastic, the stress in the model will increase proportionally with the force applied. Thus, knowing the EQV for unit loads, the stress generated by loads in the range of the occlusal forces can be deduced. However, a computer simulation operates with several simplifications related to material properties, geometry, load, and interface conditions. For this reason, when applying the results to clinical practice, a qualitative comparison between models is recommended, rather than focusing on quantitative data from FEA.[55,56]

Ratios between stress values remain the same, no matter the magnitude of the force, as long as the load applied allows only elastic deformation of the materials in the model. Thus, no attempt was made to use a particular bite force matching one of the various occlusal loads reported in the literature.[17] However, for very large loads, which produce plastic deformations, mechanical behavior of the model cannot be predicted by this method.

At the mesial end of the model, a low gradient of stress was observed. In a longer bone segment this stress would gradually decrease to the point of disappearing. However, since this stress was low, it is considered that its effect on the region of interest is negligible.

The highest bone stress was concentrated in the cortical bone in the region of the implant neck. The same tendency was also reported in other FEA of loaded implants with or without superstructure.[14,43,55–61] This location is consistent with findings from experiments and clinical studies that demonstrated that bone loss begins around the implant neck.[7,9,14,16,24,62] These results support the theory that high stress from inadvertent loading could lead to bone resorption around the implant neck.[1,7,10,14–16]

AX load generated a vertical displacement of the prosthesis and implant-abutment units in the mesiodistal plane. This led to high stress in the cortical bone, located in the thin bone plates buccally and lingually to the implant. Similar results were reported in other studies.[43,63] AX loading also produced a bending of the prosthesis in the mesiodistal plane. Consequently, the implants were bent toward the pontic, which produced a moderate stress in the cortical bone between the implant-abutment units. In the cancellous bone, this produced high stress around the apical one third of the implants, especially on the distal side. The high stress in cancellous bone buccally and lingually to the implant, which extended beyond its apical end, is a combined result of the bending and vertical displacement of the implant-abutment units. The lower elastic modulus of both resins, compared to gold and porcelain, produced a larger bending of the prosthesis and consequently greater bending of the implants toward the pontic. In the acrylic and composite resin models, this led to higher stress and larger areas of concentrated stress in the implant-abutment unit and, to a lesser extent, in the cortical bone. The geometry of the model (a slender implant-abutment unit rigidly fixed in a sizable bone mass) allowed bending rather than tilting of the implant-abutment units. Thus, the deformation was located mainly in the abutment portion, and it decreased toward the implant portion that corresponded to the bone level. Therefore, the stress differences were much larger in the implant-abutment unit than in the cortical bone, and they were absent in the cancellous bone.

BL load produced a tipping of the prosthesis in the buccolingual plane, which caused tilting of the implant-abutment units and bone toward the lingual plane. This led to lingual compression and buccal tension in the neck region of the implant and the cortical bone. The cancellous bone was under a combined stress state with compressive and tensile components.

The BL load also produced prosthesis bending in the horizontal plane, which led to a tendency of the implant-abutment unit to twist around the longitudinal axis. The degree of prosthesis bending, and thus the stress in the implant-abutment unit and the surrounding bone, would increase with a reduction in prosthesis stiffness. However, since the application point of the load was 3 mm above the longitudinal axis of the superstructure, a torque with the potential of twisting the prosthesis was also generated. Gold alloy and porcelain have high shear moduli, so almost no twisting occurred in these pontics, and most of the torque dissipated to the implant-abutment units and bone. On the contrary, since resin's low shear modulus allowed prosthesis twisting, less of the torque was transmitted to the implant-abutment units and bone. Thus, the stress from the prosthesis twisting would decrease in a less rigid prosthesis. As a combined effect of prosthesis bending and twisting tendencies under BL load, only a slight decrease in max EQV of the implant-abutment unit was found in the acrylic and composite resin prostheses and no difference in the bone stress was observed.

Bone stress was higher under BL load than under axial load in all the investigated situations. Similar results were reported in other studies[23,57] and are consistent with the findings of an in vivo experiment that demonstrated the damaging potential of lateral loads that are applied to implants.[10] However, intraorally, vertical components of occlusal forces are much larger than horizontal components.[17,64,65] Since in this study equal loads were applied axially and laterally, the ratio of stress from AX load to stress from BL load may be altered in clinical situations with higher stresses from axial force components.

Prosthesis design is considered to be one of the factors that influence the stress distribution in the bone around implants.[57,66] A previous 3-D FEA study[57] showed that the prosthesis design (namely, the presence or absence of a pontic and its location between the implants or as a cantilever extension) would significantly alter the bone stress. Other prosthesis features, such as the geometry of the superstructure, the type of connection between prosthesis and abutment, and the prosthesis material, are also believed to influence bone stress. In the present study, the prosthesis was simplified to a block whose size corresponded roughly to that of the posterior mandibular tooth crown. The prosthesis was rigidly connected to the abutments without allowing a separation between the contacting surfaces. This corresponds to the most severe conditions of load transmission, when no stress relief is allowed between the prosthetic components.

In a theoretical model[23] and an in vitro experiment,[26] acrylic resin was found to absorb shock when an impact force was applied to the occlusal material of a single implant–supported crown. While in the former study it was suggested that impact forces may occur during inadvertent biting of hard food, the latter study left the question about impact force in clinical situations unanswered.

In a static 3-D FEA, Ismail et al[30] could not demonstrate the assumed protective role of resin, since different occlusal materials did not influence the stress in bone and implants. In their abstract, however, no details were given about superstructure type, including the presence or absence of a metal framework. Since Sertgöz[32] found similar bone stress, regardless of framework and veneer material, it could be concluded that a framework would lessen any differences in the stiffness of various occlusal materials. Therefore, as a first step in investigating the influence of prosthesis materials on the stress in bone and the implant-abutment unit, no framework was modeled in the present study.

However, even in the absence of a metal framework, the use of acrylic or composite resin instead of gold or porcelain did not decrease the stress values in bone. On the contrary, a small increase in cortical bone max EQV was found under AX load. Therefore, the protective role of resin for the implant-bone interface could not be demonstrated in the conditions of this static analysis. This result concurs with those obtained in similar analyses for other prosthesis types.[31,32]

Since most of the load-related failures and significantly more bone loss occur in the early stages of loading,[1,5,67,68] acrylic-resin provisional prostheses are recommended during this period for testing the bone-implant interface (they may also be used to test esthetics and hygiene).[69] However, as the results of the present study show, acrylic resin prostheses could not be expected to lower the stress in the bone around implants.

Under AX load, using resin instead of porcelain or gold left open the possibility of bending the implant and abutment, which would consequently increase the stress in these components. This could be a potential problem with a provisional resin prosthesis in function over an extended period of time.

The stress distribution in the implant-abutment unit concurs with clinical studies that have reported that implant fractures were located mostly at the bone surface.[1,9,24,25] In those studies, fractures were often reported in conjunction with bone loss around implants, especially when this bone loss reached the implant's point of low resistance to bending, ie, the level of the abutment screw end. However, in the present study, no attempt was made to simulate these aggravating factors of implant biomechanics.

Since the load was applied to the pontic center, maximum prosthesis deformation was obtained. A load applied to any other point of the prosthesis or to an additional supporting implant would allow less bending and thus lower implant-abutment stress would be expected.

Furthermore, clinical studies show similar or even better results with a porcelain veneer on a gold framework than with a resin veneer on the same kind of framework.[1,28] Porcelain and gold alloy have comparable Young's moduli, which explain why similar stress was found in these 2 materials. Also, it may be inferred from this study that a model with gold framework and porcelain veneer would show stresses similar to a model with a gold alloy prosthesis. The stress values found in the porcelain and resin models may encourage the use of porcelain as an occlusal material, when occlusal forces are in the porcelain elastic range.

Conclusions

Similar stress was found in bone and the implant-abutment units in the gold alloy and porcelain prosthesis models. The protective role of resin for the implant-bone interface could not be demonstrated under the conditions of this analysis. Considering the intraoral predominance of axial loads, the use of acrylic or composite resin instead of porcelain or gold may increase stress in the implant and the abutment, in the absence of a metal framework.

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