Influence of Material of Overdenture-Retaining Bar with Vertical Misfit on Three-Dimensional Stress Distribution

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Abstract

Purpose: This study evaluated the effects of different bar materials on stress distribution in an overdenture-retaining bar system with a vertical misfit between implant and bar framework.
Materials and Methods: A three-dimentional finite element model was created including two titanium implants and a bar framework placed in the anterior part of a severely reabsorbed jaw. The model set was exported to mechanical simulation software, where displacement was applied to simulate the screw torque limited by 100-μm vertical misfit. Four bar materials (gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy) were simulated in the analysis. Data were qualitatively evaluated using Von Mises stress given by the software.
Results: The models showed stress concentration in cortical bone corresponding to the cervical part of the implant, and in cancellous bone corresponding to the apical part of the implant; however, in these regions few changes were observed in the levels of stress on the different bar materials analyzed. In the bar framework, screw, and implant, considerable increase in stress was observed when the elastic modulus of the bar material was increased.
Conclusions: The different materials of the overdenture-retaining bar did not present considerable influence on the stress levels in the periimplant bone tissue, while the mechanical components of the system were more sensitive to the material stiffness.

Implant-retained overdentures can be attached in two ways. The first is with the use of resilient attachments on freestanding implant abutments. The second is with the use of resilient attachments to attach the denture to a rigid bar assembly that interconnects the osseointegrated implants.[1] Most authors agree that a passive fit between the prosthesis framework and osseointegrated dental implants is required.[2-5] The resiliency of the periodontal membrane found in natural dentition is absent in the case of osseointegrated dental implants;[6] thus, they are unable to adjust to the misfits. When there is poor fit between structures, tensile, compressive, and bending forces may be introduced into an implant-supported restoration and may result in failure of the components.[5,7-9] Moreover, a poor-fitting framework may transfer unwelcome stress onto the bone-implant interface, which may induce loss of osseointegration.[4,10-12] Nevertheless, several studies have shown some biologic tolerance of osseointegrated dental implants to certain levels of misfit;[13-16] however, there is difficulty in determining these states due the limitations of these studies and ethical principles involved with in vivo studies.

Some authors have attempted to define an acceptable level of implant denture fit.[17,18] In 1983, Branemark was the first to define passive fit, and he proposed this should be at the level of 10 μm to enable bone maturation and remodeling in response to occlusal loads.[17] In 1991, Jemt defined passive fit as the level that did not cause any long-term clinical complications and suggested misfits smaller than 150 μm were acceptable.[18] Although the preceding values have been reported and used as reference, they are of empirical origin.

Potential distortion can be created at any step of the fabrication process. The error is mostly due to the volumetric inconsistency and linear expansion of the fabrication materials used, which include impression material, gypsum products, waxes, investments, and casting metal.[19-25] Several postcasting techniques have been developed to correct inaccuracies of fit resulting from the fabrication process;[26-29] however, denture misfits are a clinical reality.

Several alloys and metals have been used to make denture frameworks. The first implant-supported denture frameworks fabricated of gold alloy began to be used in oral rehabilitations in the early 1970s.[30] Nevertheless, the high cost of noble alloys has led to a search for substitutes: cobalt-chromium alloys,[31] silver-palladium alloys,[32] and titanium alloys.[33] A study evaluated the effect of four framework materials on the stress distribution in a six-implant-supported fixed denture and periimplant bone tissue;[34] however, the authors did not consider the misfits present in implant dentures. There is limited information about the influence of overdenture-retaining bar material with misfit on biomechanical behaviors. In addition, another study suggested that by deforming the framework-implant system to close horizontal misfits, the resiliency of the framework could have a significant effect on the stress distribution,[35] increasing the importance and significance of the present study.

Numerical analysis can help overcome the limitations of traditional experimental methods by offering accurate and reliable information about the biomechanical efficiency of multiple implant prostheses with regard to bar, implant, and bone response.[36] Thus, the aim of this study was to evaluate the effects of different bar materials (gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy) on three-dimensional (3D) Von Mises stress distribution in the bar framework, periimplant bone tissue, screw, and implant of a rigid bar assembly widely used at present to attach overdentures on two osseointegrated implants with 100-μm vertical misfit.

Materials and methods

The 3D model was defined starting with clinical data taken from a common situation. An anterior part of a severely resorbed jaw and an overdenture-retaining bar system over two osseointegrated implants were modeled using a 3D parametric solid modeler (Rhinoceros 3.0 software; McNeel, Seattle, WA). The geometry of the jaw portion modeled was obtained starting with CT data from a type III bone condition.[37] Two 3.75-mm diameter × 10-mm length titanium dental implants (Nobel Biocare, Yorba Linda, CA) with external hexagon were selected. A circular bar (2-mm diameter) and two calcinable UCLAs of an overdenture-retaining bar system (Conexão Sistema de Prótese, São Paulo, SP, Brazil) were also modeled, with a distance of 18.5 mm between the UCLA centers. The FE model was obtained by importing the solid model into mechanical simulation software (NEiNastran 9.0; Noran Engineering Inc., Westminster, CA) using STEP (*.stp) format. The corresponding elastic properties, such as Young's modulus and Poisson ratio, were determined from values obtained from the literature[38-42] (Table 1).

The following assumptions were made: All materials were presumed to be linear elastic, homogenous, and isotropic. Because of the lack of precise information regarding the material properties of bone, the cortical and cancellous bone were assumed to have these properties.[43] The implant thread and cancellous and cortical bone were removed, because after several convergence tests, they were found to be not relevant to the analysis and provided a relevant reduction in elements. Complete adhesion was considered between bone and implant, and bar and implant, provided by osseointegration and screw torque, respectively. Screw and implant were considered a single structure, because of not being relevant to the purpose of the analysis. The model stability was carried out to obtain a reliable model, which was regarded as relevant to engineering and clinical aspects.

Table 1 Material properties

Material	Young's modulus (GPa)	Poisson's ratio (v)
Cortical bone[38]	13.7	0.3
Cancellous bone[38]	1.37	0.3
Titanium (implant)[40]	110	0.33
Titanium (screw)[39]	110	0.28
Type IV gold alloy[42]	80	0.33
Silver-palladium alloy[41]	95	0.33
Commercially pure titanium[39]	110	0.28
Cobalt-chromium[41]	218	0.33

A 3D FE model was constructed using 10-node tetrahedral elements. The volumes were redefined in the new environment and meshed, finally resulting in a model with 11,718 elements and 21,625 nodes. The investigated model showed the configurations presented in Fig 1. All nodes on the external bone surface were constrained in all directions to allow application of the displacement condition and stress to be created in the models. One displacement was performed at the end of the bar framework to simulate the vertical misfit. Thus, at the end of the bar framework a displacement limited by 100-μm vertical misfit was applied, simulating the screw torque (Fig 2). Control groups without misfits were created as baseline. A total of eight models were created with the bar materials investigated (gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy). Stability of the model was checked, and particular attention was paid to the refinement of the mesh resulting from the convergence tests at the bone/implant interface.

The results for qualitative analysis were represented by figures and color gradients of Von Mises stresses, and presented in terms of the Von Mises stress values, because a higher Von Mises stress is a strong indication of a greater possibility of failure.

Results

Von Mises stresses that occurred in the bar framework, periimplant bone tissue, screw, and implant for all models, before and after applying 100-μm displacement simulating the screw torque, are presented in Figures 3 and 4. In the control groups, no stress was created in the models, regardless of the bar material.

Figure 3 shows Von Mises stress distribution in the bar framework and periimplant bone tissue for different bar materials. The models showed stress concentration in the cortical bone corresponding to the cervical part of the implant, and in the cancellous bone corresponding to the apical part of the implant; however, different bar materials showed little influence on the stress distribution in the periimplant bone tissue. In the bar framework, there was considerable increase in stress in the bar materials with higher stiffness.

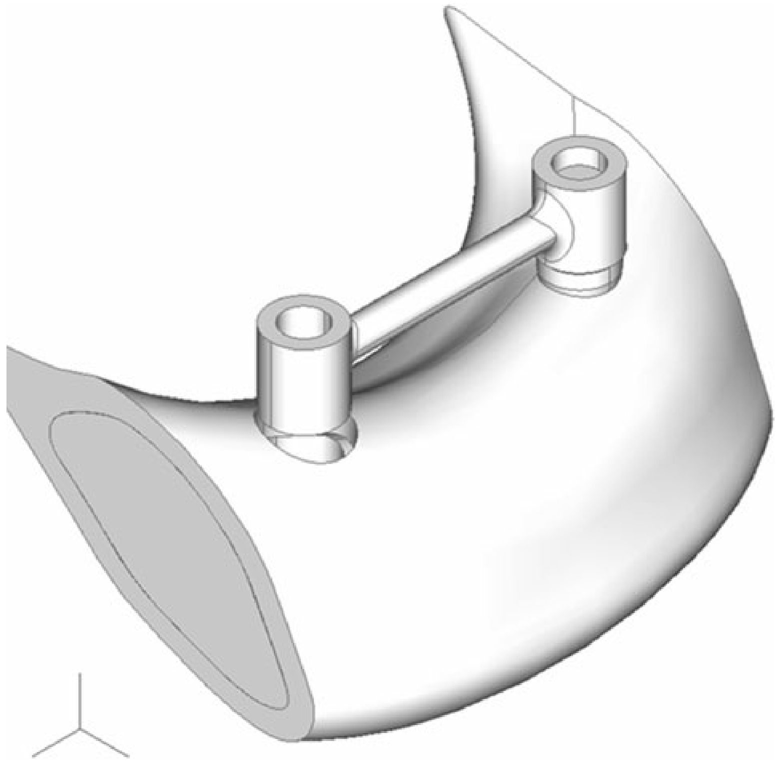

Figure 1 Design of the investigated model.

Figure 4 shows Von Mises stress distribution in the screw and implant for different bar materials. The models showed concentration in the screw neck, implant platform and neck. The different bar materials showed little influence on the stress levels in the implant neck; however, in the screw neck and implant platform, there was considerable increase in stress in the bar materials with higher stiffness.

Discussion

The model used in the present study involved several assumptions regarding the simulated structures. The structures in the model were all assumed to be homogeneous, isotropic, and to have linear elasticity. The proprieties of the materials modeled in this study, particularly the living tissues, however, are different. For instance, it is well documented that the cortical bone of the mandible is transversely isotropic and inhomogeneous. In addition, a 100% implant/bone interface was established, which does not match clinical situations. The effect of the bone/implant contact ratio at the bone/implant interface on stress distribution in the periimplant bone has been argued. One study presented a new mimic FE model simulating the entire structure of the periimplant cancellous bone showing a more homogeneous stress distribution when compared with conventional bone used in other studies.[44] In contrast, another study showed that the degree of osseointegration did not affect stress distributions by FEA.[45] Thus, the inherent limitations of the FEA as regards to stress distribution should always be taken into consideration.

The FEA showed considerable changes in the stresses induced in the bar framework, screw neck, and implant platform for the different bar materials investigated with higher stress levels in the cobalt-chromium alloy, whereas in the periimplant bone tissue this variable had little influence on stress distribution, suggesting these components of the system are more sensitive to stiffer materials. Natali et al,[35] evaluating the effects of horizontal misfits, suggested that higher framework resiliency could reduce stress levels transferred to the periimplant bone tissue. The present study is in disagreement with this hypothesis, since the vertical misfit did not create considerable changes in stress levels in the periimplant bone tissue for the different bar materials; however, a study on horizontal misfit needs to be conducted to elucidate this disagreement.

The results of the present study are partially in agreement with Sertgoz,[34] who evaluated the effects of four framework materials (gold, silver-palladium, cobalt-chromium, titanium alloys) and three occlusal surface materials (resin, resin composite, porcelain) on the stress distribution in a six-implant-supported fixed denture and periimplant bone tissue and related that the use of a framework with lower elastic modulus did not produce a significant change in the stress levels in the periimplant cortical and cancellous bone; these data are agreement with the present study. However, Sertgoz stated that the lower elastic modulus of the framework material increased the potential risk of denture failure, in disagreement with the present study. This difference in the results may be explained by the hypothesis that the misfit changed the biomechanical behavior in relation to stress distribution in the bar framework, but no occlusal load was applied in the present study. The torque

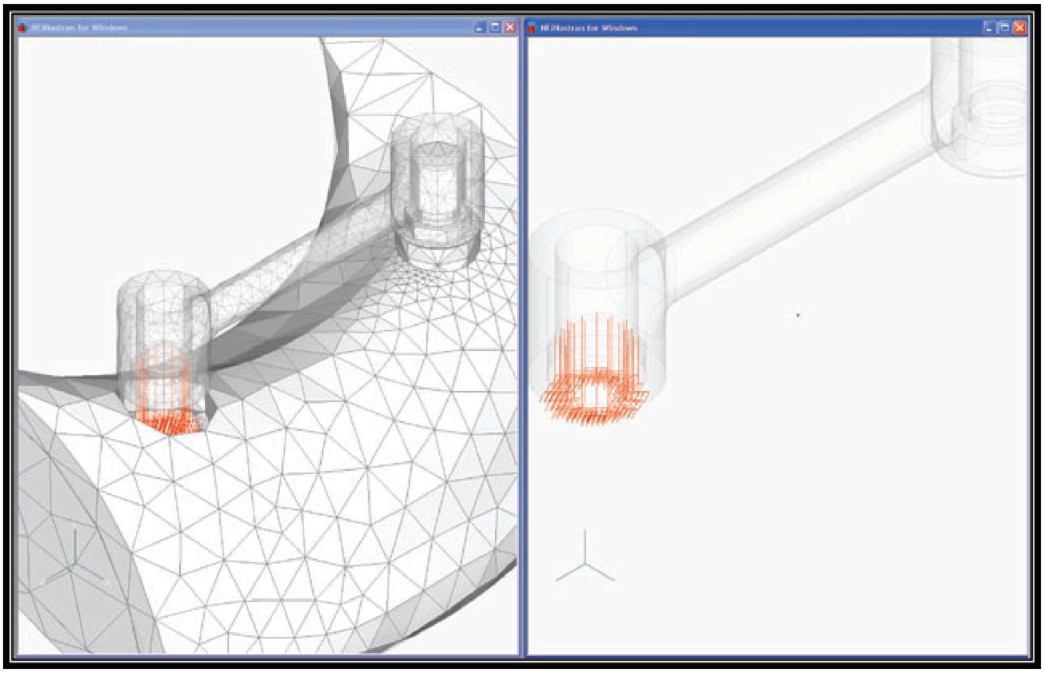

Figure 2 The displacement created to simulate the closure of the misfit.

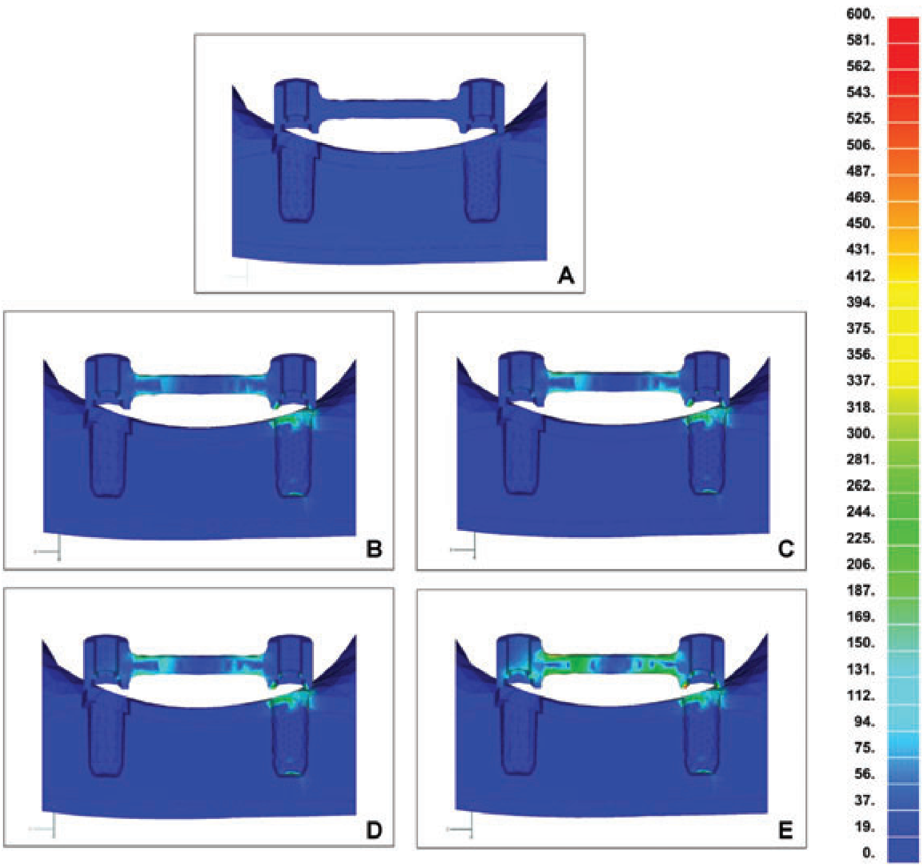

Figure 3 Von Mises stress (MPa) distribution in the framework and periimplant bone tissue for the different bar materials: (A) no misfit, regardless of the material (control group); (B) gold alloy; (C) silver-palladium alloy; (D) commercially pure titanium; (E) cobalt-chromium alloy.

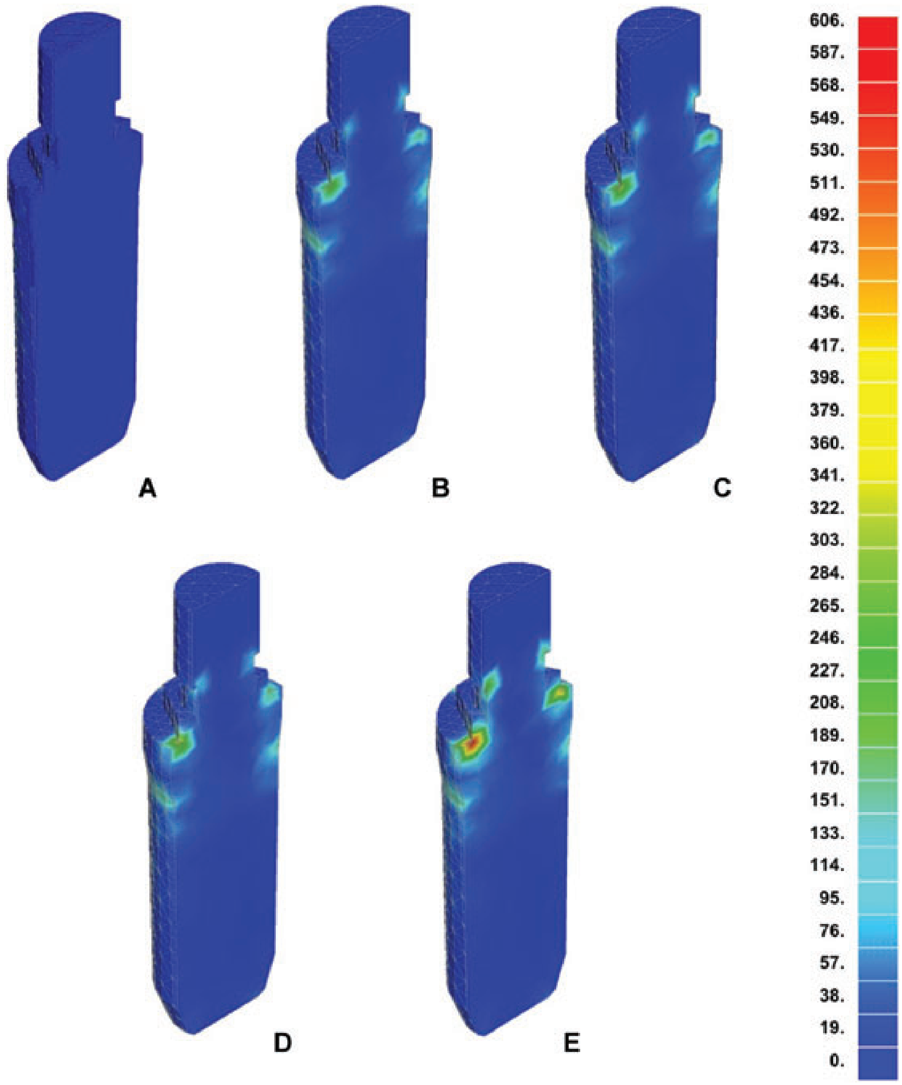

Figure 4 Von Mises stress (MPa) distribution in the screw and implant for the different bar materials: (A) no misfit, regardless of the material (control group); (B) gold alloy; (C) silver-palladium alloy; (D) commercially pure titanium; (E) cobalt-chromium alloy.

created on the overdenture-retaining bar from a nonfitting denture could increase the stress levels, increasing the negative effect of the high stiffness of the overdenture-retaining bar. Thus, the effects of the occlusal load need further evaluation to confirm this hypothesis.

Several studies have indicated a certain biological tolerance for denture misfit in the living bone.[13-16] A longitudinal study verified mean marginal bone loss of 0.5 and 0.2 mm for screw-retained prostheses with misfit of 111 and 91 μm, respectively. The authors found no statistical correlation between marginal bone level changes and different denture misfit values. Moreover, the authors observed that the implants were stable and immovable after years in function, suggesting certain biological tolerance to denture misfits.[13] Some authors consider marginal bone loss between 0.4 and 1.6 mm in the first year, and around 0.1 mm of subsequent loss per year after the first year to be acceptable;[10,46,47] however, the consequences of a lack of fit include micromovement that may cause failure of the prosthetic components,[7] such as loosening or facture of the prosthetic or abutment screw, and fracture of the framework or veneers.[1,8,10,11,48] Another study showed a significant loosening of torque on prosthetic screws for 100 and 175 μm misfits introduced between an implant-supported fixed complete denture and terminal abutment, after cyclic load, suggesting that denture misfit of such dimensions should be considered clinically unacceptable.[5] Thus, vertical misfits associated with the stiffness of materials could predispose the prosthetic components to failure. Another study showing the effect of the misfit amplification should be conducted to verify these states.

These data may suggest a different level of tolerance to vertical misfits and material stiffness of the framework between the biologic and mechanical complications of implant-supported dentures; however, it is premature to affirm that vertical misfits are not prejudicial to the bone/implant interface, and to determine which misfit is clinically acceptable, based on the available literature. To acknowledge and support studies using FEA for evaluating stress in bone tissue, further studies are essential to show quantitative stress for positive remodeling in osseointegration. Moreover, other factors, such as loading caused by clips to attach an overdenture, horizontal misfits, and the increase in misfit of the bar framework on the implants, which could influence the stress distribution in the overdenture-retaining bar system, are already under investigation.

Conclusion

Within the limitations of this FEA, the following conclusions can be drawn:

1. The vertical misfit created concentration of stresses in the overdenture-retaining bar system.
2. The different bar materials with vertical misfit simulated showed a large influence on the stress levels in the bar framework, screw, and implant, since a lower elastic modulus decreased the stress levels.
3. The different bar materials with vertical misfit simulated had no considerable influence on the stress levels in the periimplant bone tissue and implant neck.

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