UNIVERSIDADE ESTADUAL DE CAMPINAS
FACULDADE DE ODONTOLOGIA DE PIRACICABA

ATAÍS BACCHI

INFLUÊNCIA DO MATERIAL DE INFRAESTRUTURA E DO DESAJUSTE VERTICAL DE PRÓTESES PARCIAIS FIXAS IMPLANTOSSUPORTADAS NAS TENSÕES TRANSMITIDAS ÀS ESTRUTURAS PROTÉTICAS E AO TECIDO ÓSSEO PERI-IMPLANTAR

Dissertação apresentada à Faculdade de Odontologia de Piracicaba, da Universidade Estadual de Campinas, para obtenção do título de Mestre em Clínica Odontológica - Área de concentração Prótese Dental.

Orientador: Prof. Dr. Rafael Leonardo Xediek Consani

Este exemplar corresponde à versão final da Dissertação
defendida pelo aluno Ataís Bacchi e orientada
pelo Prof. Dr. Rafael Leonardo Xediek Consani

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Banca examinadora:

Rafael Leonardo Xediek Consani [Orientador]

Marcelo Ferraz Mesquita
Jarbas Francisco Fernandes dos Santos

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Rafael L. X. Consani

Prof. Dr. RAFAEL LEONARDO XEDIEK CONSANI

Prof. Dr. JARBAS FRANCISCO FERNANDES DOS SANTOS

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DEDICATÓRIA

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RESUMO

O objetivo neste estudo foi avaliar a influência do material de infraestrutura e diferentes níves de desajuste vertical na concentração de tensões em prótese parcial fixa implantossuportada (infraestrutura e porcelana de cobertura), parafuso de retenção e tecido ósseo peri-implantar durante o assentamento protético e frente à aplicação de carga oclusal. Um modelo tridimensional de elementos finitos de uma porção posterior de mandíbula contendo dois implantes osseointegrados nas posições de segundo pré-molar e segundo molar, suportando uma prótese parcial fixa foi construído utilizando *software* específico de modelagem (SolidWorks 2010). Modelos de elementos finitos foram obtidos pela importação do modelo sólido ao *software* de simulação mecânica (ANSYS Workbench 11). Os modelos foram separados em grupos de acordo com o material de infraestrutura (liga de ouro tipo IV, liga de prata-paládio, titânio comercialmente puro, liga de cobalto-cromo ou zircônia) e o nível de desajuste vertical (10 µm, 50 µm e 100 µm) criado na interface prótese-implante do segundo pré-molar. A concentração de tensões foi avaliada nas seguintes condições: (1) assentamento protético; e (2) cargas oclusais simultâneas de 110 N vertical e 15 N horizontal em cada dente. Os resultados obtidos mostraram que as infraestruturas mais rígidas apresentam maior concentração de tensões internas; entretanto, promoveram menores concentrações de tensão sobre a porcelana de recobrimento, em ambas condições avaliadas. Na análise do assentamento protético, materiais mais rígidos para infraestruturas aumentaram os valores de tensão no parafuso de retenção e não causaram diferença relevante nas tensões no tecido ósseo peri-implantar. Quando a carga foi aplicada, o uso de infraestruturas mais rígidas promoveu redução de tensões no parafuso de retenção e no tecido ósseo peri-implantar. Em ambas condições avaliadas um considerável aumento na concentração de tensões foi obsevado em todas as estruturas com a amplificação do desajuste. Nas diferentes simulações, o material de infraestrutura exerceu considerável influência nas tensões transmitidas às estruturas avaliadas, exceto ao tecido ósseo peri-implantar em condições de assentamento. Aumento de tensões em todas as estruturas pode ser observado com o aumento do desajuste.

Palavras-chave: Implantes dentais; prótese dentária fixada por implante; biomecânica; análise por elementos finitos.

ABSTRACT

The aim in this study was to evaluate the influence of the framework material and vertical misfit on the stresses created in an implant-supported partial prosthesis (framework and porcelain veneer), retention screw and peri-implant bone tissue during the settlement of the prosthesis and under load conditions. A 3-D Finite Element model of a posterior part of a jaw with two osseointegrated implants at the place of the right second pre-molar and second molar supporting an implant-supported fixed partial prosthesis was constructed using specific modeling software (SolidWorks 2010). Finite element models were obtained by importing the solid model into mechanical simulation software (ANSYS Workbench 11). The models were divided into groups according to the framework material (type IV gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy or zirconia) and vertical misfit level (10 µm, 50 µm and 100 µm) created at the second pre-molar implant-prosthesis interface. The stress concentration was evaluated in the following conditions: (1) settlement of the prosthesis; and (2) simultaneous loads of 110 N vertical and 15 N horizontal in each tooth. The obtained results showed that stiffer frameworks presented higher stress concentrations in it and led to lower stresses in the porcelain veneer, in both conditions. In the analysis of settlement of the prosthesis, stiffer framework materials increased the stress values in the retention screw and did not cause a relevant difference in the stresses values in peri-implant bone tissue. When the load was applied, the use of more stiffness frameworks led to lower stresses in the retention screw, and peri-implant bone tissue. In both conditions evaluated, considerable raise of stress concentration was observed in all the structures within misfit amplification. Comparing the results of the different simulations, the framework materials presented a considerable influence on the stress concentration in the structures evaluated, except on the peri-implant bone tissue during the settlement of the prosthesis, while a considerable increase of the stress in all the structures was observed with the increase of the misfit.

Key-words: dental implant; dental prosthesis, implant-supported; biomechanics; finite element analysis.

SUMÁRIO

INTRODUÇÃO

Com a inclusão dos implantes osseointegrados na Odontologia um considerável aprimoramento pode ser observado na reabilitação protética de pacientes edêntulos ou desdentados parciais. Avaliações clínicas longitudinais (acompanhamento por mais de 5 anos) relatam taxas de sobrevivências muito favoráveis dos implantes e da prótese em reabilitações com próteses totais fixas, próteses parciais fixas, *overdentures* ou coroas unitárias (Wennerberg & Albrektsson, 2011).

Em relação às reabilitações com próteses parciais fixas, uma alta taxa de sobrevivência dos implantes foi relatada (92 a 97%). Entretanto, a taxa de sobrevivência da prótese variou entre 86 e 100%, onde fratura do material da prótese e o afrouxamento dos parafusos foram as principais complicações, depois de 5 anos de acompanhamento em estudos clínicos (Jemt & Lekholm, 1993; Lekholm *et al*., 1994; Lekholm *et al.*, 1999; Wennstrom *et al*., 2004; Wennerberg & Albrektsson, 2011).

Esse tipo de prótese é usualmente composta de uma infraestrutura com um material de recobrimento (DeHoff *et al*., 2006; Erkmen *et al*., 2011). Incialmente, as ligas de ouro foram os materiais mais utilizados para a confecção da infraestrutura; porém, devido ao alto custo das ligas de ouro, algumas ligas não-nobres foram introduzidas na Odontologia, como as ligas de cobalto-cromo, prata-paládio e titânio (Abreu *et al*., 2010). Posteriormente, infraestruturas em zircônia foram propostas como outra alternativa às infraestruturas metálicas devido à baixa degradação química, baixo potencial à aderência de biofilme, biocompatibilidade, estética e propriedades mecânicas (DeHoff *et al*., 2006), superando alguns incovenientes observados nas ligas metálicas convencionais, como corrosão e limitações estéticas (Pietrabissa *et al*., 2000; Erkmen *et al*., 2011). Desde então, tem sido observado um exponencial aumento na utilização da zircônia como material de infraestrutura (Denry & Kelly, 2008; Abduo *et al*., 2011). Essas ligas metálicas para confecção de infraestruturas tem sido apontadas como biomecanicamente importantes, exercendo influência nas tensões propagadas para as estruturas protéticas e tecido ósseo (Meriç *et al*., 2011).

Para o sucesso e longevidade do tratamento protético implantossuportado, tem sido apontado como pré-requisito a presença da adaptação passiva da prótese sobre a plataforma do implante (Sahim & Cehreli, 2001). Branemark, 1983, preconizou como aceitável uma tolerância de até 10 μm de desajuste vertical entre as estruturas. Entretanto, durante a confecção da prótese, algumas distorções podem acarretar aumento desse valor, em decorrência de diversos procedimentos clínicos e laboratoriais, como na moldagem e na obtenção do modelo de trabalho (em função da técnica e material empregado), na confecção da infraestrutura (durante enceramento, fundição ou fresagem) e durante a aplicação do material de recobrimento (Wee *et al.*, 1999).

Algumas possíveis complicações têm sido creditadas à ausência de adaptação passiva, incluíndo falhas mecânicas como fratura do material de recobrimento e da infraestrutura, assim como também fratura e afrouxamento dos parafusos de fixação. Dentre as complicações biológicas são apontadas inflamação gengival, dor, fístula e perda de tecido ósseo peri-implantar (Torres *et al.*, 2007, Monteiro *et al.*, 2010).

Estudos prévios utilizando primatas (Carr *et al.*, 1996) e coelhos (Michaels *et al.*, 1997; Duyck *et al.*, 2005) avaliaram as consequências de diferentes níveis de desajuste vertical sobre o tecido ósseo peri-implantar. Entretanto, esses estudos apresentam grandes limitações voltadas para a impossibilidade de avaliar a presença do desajuste quando existe carga oclusal (Natali *et al.,* 2006) bem como a impossibilidade de avaliar as consequências em estruturas protéticas, fatores importantes no sucesso do tratamento e apontados como diretamente afetados pela presença do desajuste (Spazzin *et al.*, 2011, Assunção *et al.*, 2010, Kunavisarut *et al.*, 2002).

Alguns estudos clínicos têm apontado certo nível de tolerância do tecido ósseo frente à ausência de adaptação passiva de próteses implantossuportadas. Em estudos prévios, desajustes de até 150 μm foram considerados como aceitáveis (Jemt,1991), desajustes médios de 111 μm e 91 μm foram encontrados para grupos de acompanhamento longitudinal de 1 ano e 5 anos, respectivamente, o qual não mostrou correlação entre desajuste e alterações no nível ósseo marginal (Jemt & Book, 1996). Entretanto, esses estudos foram realizados com pacientes edêntulos reabilitados com próteses totais suportadas por cinco a sete implantes. Assim, estes mesmos níveis de tolerância de desajuste podem não ser aceitáveis pela prótese parcial suportada por um número mínimo de implantes, visto que alguns fatores tem sido apontados como responsáveis por influenciar as tensões transmitidas aos tecidos de suporte, dentre eles o número de implantes e o tipo de prótese (total, parcial ou unitária) (Brunski & Hoshaw, 1994; Koriot & Johann, 1999).

O método considerado mais seguro para avaliar a resposta biomecânica é a avaliação clínica. Entretanto, o estudo do comportamento biomecânico de estruturas *in vivo* fica inviabilizado por aspectos éticos e/ou metodológicos (Abreu *et al.*, 2010). O desenvolvimento de modelos tridimensionais (3-D) específicos por elementos finitos é ferramenta alternativa para investigar forças que ocorrem no osso de forma semelhante ao que acontece *in vivo* sem danificar estruturas, oferecendo informações precisas e confiáveis a respeito da biomecânica envolvida em diversas situações clínicas (Bergendal & Palmqvist, 1995; Taddei *et al.*, 2006). Essa metodologia possibilita prever e quantificar as tensões induzidas no sistema prótese/implante e tecidos de suporte e determinar a capacidade de cada estrutura em suportar determinadas cargas dentro de dada situação clínica. Dessa forma, baseado nos resultados obtidos por meio dessa metodologia, o profissional estará melhor preparado para interpretar as situações clínicas bem como sugerir estudos clínicos para desvendar certas situações específicas (Geng *et al.*, 2001).

Estudos prévios utilizaram a metodologia de Elementos Finitos para avaliar a influência do desajuste vertical em próteses parciais fixas implantossuportadas (Winter *et al.,* 2010; Kunavisarut *et al.,* 2002) e barras para retenção de *overdentures* suportadas por dois implantes (Abreu *et al.,* 2010; Spazzin *et al.,* 2011) quanto às tensões transmitidas ao tecido ósseo peri-implantar; entetanto, resultados controversos foram observados uma vez que o aumento do desajuste causou um aumento nas tensões em tecido ósseo nos estudos com próteses parciais fixas implantossuportadas e não influenciaram os valores de tensão no tecido ósseo peri-implantar em sistemas para retenção de *overdentures*. Diferentes materiais de infraestrutura em coroas unitárias (Sevimay *et al*., 2005), próteses parciais fixas (Erkman *et al*., 2011; Meriç *et al*., 2011) e prótese totais (Sertgöz *et al*., 1997) foram avaliadas em relação às tensões transmitidas ao tecido ósseo peri-implantar e estruturas protéticas; entretanto, a presença do desajuste vertical, uma possibilidade clínica, não foi considerada.

O propósito neste estudo foi utilizar o método por elementos finitos para avaliar a influência do material da infrestrutura frente à diferentes níveis de desajuste vertical nas tensões criadas nas estruturas protéticas (infraestrutura e porcelana de cobertura), parafuso de fixação e tecido ósseo peri-implantar de prótese parcial fixa implantossuportada na condição de assentamento protético e sob carga oclusal.

O presente trabalho é apresentado no formato alternativo de dissertação de acordo com as normas estabelecidas pela deliberação 002/06 da Comissão Central de Pós-Graduação da Universidade Estadual de Campinas. O capítulo 1 foi submetido à revista *Journal of Prosthodontics: Implant, Esthetic and Reconstructive Dentistry* e o capítulo 2 está formatado nas normas da revista *Journal of Oral Rehabilitation*.

CAPÍTULO 1

Stress distribution in fixed-partial prosthesis and peri-implant bone tissue by different framework materials and vertical misfit levels – 3-D finite element analysis

Running title: Prosthetic framework and misfit on stress concentration

Abstract

Purpose: The aim of this study was to evaluate the influence of framework material and vertical misfits on the stresses created in an implant-supported partial prosthesis during the settlement of the prosthesis.

Material and Methods: A 3-D finite element model was defined starting with clinical data taken from a common situation. A posterior part of a severely reabsorbed jaw with two osseointegrated implants at the second premolar and second molar was modeled using specific modeling software (SolidWorks 2010). Finite element models were obtained by importing the solid model into mechanical simulation software (ANSYS Workbench 11). The models were divided into groups according to the prosthesis framework material (type IV gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy, or zirconia) and vertical misfit level (10 µm, 50 µm, and 100 µm) created at one implant-prosthesis interface. The gap of the vertical misfit was set to be closed and the stress values were measured in the framework, porcelain veneer, retention screw, and bone tissue.

Results: Stiffer materials led to higher stress concentration in the framework and increased the stress values in the retention screw, while in the same circumstances, the porcelain veneer showed lower stress values, and no relevant difference in stress in the peri-implant bone tissue was observed. A considerable increase in stress concentration was observed in all the structures evaluated within the misfit amplification.

Conclusion: The framework material influenced the stress concentration in the prosthetic structures and retention screw, what was not observed in bone tissue. All the structures were considerably influenced by the increase in the misfit levels.

Key-words: dental implant; dental prosthesis, implant-supported; biomechanics.

Introduction

With the advent of osseointegrated dental implants, significant improvements in prosthetic treatment in either partially and completely edentulous patients have been observed.[1] A more rigid connection between the osseointegrated implant and the peri-implant bone tissue is observed in comparison with the resilience of the periodontal ligament of the natural dentition.[2] Thus, a passive fit at the implant-prosthesis interface has been suggested to be crucial for the long-term success of osseointegration[3] and to prevent future complications.[4] Dimensional changes can occur during the clinical and laboratory procedures of prosthesis fabrication as a result of inappropriate clinical practice or manufacturing defects.[2,5]

Many complications can be caused by misfit in prosthetic frameworks. These complications may include biologic effects such bone deformation and remodeling, microdamage, continual resorption, or even loss of osseointegration.[2,6] Mechanical complications include porcelain fracture, screw loosening or fracture, and framework fracture.[2,4,7,8] Some publications associate these complications and the misfit of the prosthesis.[9-11] However, the exact relationship between prosthesis misfit and implant complications is still poorly understood.[12]

During the clinical and technical steps of prosthesis fabrication, some distortions can occur, harming the achievement of passive fit. These distortions may be related to the impression procedure,[13] master cast fabrication,[14] wax pattern fabrication,[15] casting,[16,17] porcelain firing,[13] and tolerance of the different implant components.[2,15,18] In addition, a biologic tolerance has been suggested regarding the presence of misfit;[10,19,20] however, there is difficulty in determining these states due the limitations of these studies and the ethical principles involved with in *in vivo* studies.[21]

The influence of materials type used in framework fabrication has been suggested to be very important for biomechanical reasons. When loads are applied on the superstructure, stresses are created within them and transferred to the bone-implant interface, implant, and prosthetic components.[22] They could influence the survival of the restoration and affect the bone stress distribution around implants.[23,24] Initially, the gold alloy was the most frequently used material for framework fabrication, but due to its high cost, alternative alloys were introduced in dentistry, including cobalt-chromium, silver-palladium, and titanium alloys.[21] More recently, zirconia frameworks were proposed as an esthetic alternative for the metallic implant framework due to their chemical durability, aesthetics, biocompatibility, unsupportive plaque accumulation potential, and superior mechanical properties,[25] solving some problems that were observed in metal alloys, such as corrosion and esthetic limitations.[1,26] These facts led to an exponential increase in zirconia application as a framework material for dental prosthesis.[2,27]

The stresses on prosthetic structures and bone tissue are not observed only when occlusal loads are applied. Stresses are created also when ill-fitting prostheses are installed,[28,29] and the values of these generated stresses are influenced by the different stiffness of the framework material.[23,24] Previous finite element analyses (FEA) evaluated the influence of the increase in vertical misfit in implant-supported fixed prostheses[12,30] and overdenture-retaining bars supported by two implants.[21,31] However, controversial results were observed where the misfit amplification caused a considerable increase in stresses in peri-implant bone tissue in implant-supported fixed prostheses and did not influence the stress values in peri-implant bone in overdenture retaining systems. Different framework materials for single crowns[32] fixed-partial prostheses[26] and full arch prostheses[33] were evaluated with respect to the stresses transferred to peri-implant tissue and prosthetic structures; however, the presence of vertical misfit, a clinical possibility, was not considered. The aim of this study was to evaluate the influence of the framework material and different levels of vertical misfit on the stresses created in a partial implant-supported prosthesis (framework and porcelain veneer), retention screw, and peri-implant bone during the settlement of the prosthesis.

Materials and Methods

The three-dimensional model was defined starting with clinical data taken from a common situation. A posterior part of a severely reabsorbed jaw with two osseointegrated titanium implants (External Hexagonal, 4.0-mm diameter x 10-mm length) at the right second pre-molar and second molar with a distance of 16.1mm between them and a fixed-partial denture were modeled using specific 3-D modeling software (SolidWorks 2010, SolidWorks Corp., Concord, Massachusetts, USA). The implant thread was removed because, after convergence tests, they were found to be irrelevant to the analysis and caused a relevant reduction in elements.

Finite element models were obtained by importing the solid model into mechanical simulation software (ANSYS Workbench 11, Ansys Inc., Canonsburg, Pennsylvania, USA). The models were divided into groups according to the framework material – type IV gold alloy (Au), silver-palladium alloy (Ag-Pd), commercially pure titanium (Ti), cobalt-chromium alloy (Co-Cr), or zirconia (Zr) – and misfit level (10 µm, 50 µm, and 100 µm) created at the second pre-molar implant-prosthesis interface. All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. The elastic properties used were taken from the literature (Table 1).

Model stability was ensured to obtain a reliable model that was regarded as relevant with respect to engineering and clinical aspects.[21] The total number of elements and nods generated in the FE models were, respectively, 736.750 and 1178.870 for 10 µm, 742.289 and 1187.188 for 50 µm, and 725.737 and 1160.223 for 100 µm of vertical misfit. The shape of the element was tetrahedral with 10 nodes. The investigated models showed the configurations presented in Figure 1. The 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 base of the mandible was set to be the fixed support, the gap of the vertical misfit was set to be closed, and data for the maximum principal stresses (framework, porcelain veneer, and bone tissue) and von Mises stresses (retention screw) were produced numerically, color-coded, and compared among the models.

Results

Framework

A relevant increase in the maximum principal stress (MPS) values in the frameworks was observed when stiffer materials were evaluated. The increase in the stress values was also proportional to the misfit levels. The higher stress concentrations occurred in the metallic strap of the abutment of the molar, more specifically in the mesial region, where it comes into contact with the implant platform. All the stress values are represented in Table 2. The MPS values in the frameworks with 100 µm of vertical misfit are presented in Figure 2.

Porcelain Veneer

There was a relevant decrease in the MPS values in the porcelain veneer when stiffer frameworks were analyzed (Co-Cr and Zr). However, the use of less rigid materials (Au, Ag-Pd, and Ti) did not lead to relevant differences in the stress distribution. A significant increase in the stress values was observed when the misfit was amplified. As in the frameworks, the maximum stress values were observed in the metallic strap of the abutment of the molar in the mesial region close to the framework interface. All the stress values for the porcelain veneer materials are listed in Table 3.

Retention screw

The von Mises stress values occurred in the molar retention screw, and an increase in the values was observed in accordance with the increase of the stiffness of the frameworks. An increase in the stress values in the screw was also observed with the misfit amplification. The stresses were observed in the long axis of the screws. The stress values for the screws in the different situations analyzed are presented in Table 4.

Bone stress

The framework material was shown to be irrelevant in influencing the MPS in the bone tissue. An increase in the stress concentration could be observed with the misfit amplification. The cortical bone at the implant-bone interface showed higher stress values. The stress values for the different situations are presented in Table 5. The MPS at all levels of vertical misfit in the cobalt-chromium alloy framework is presented in Figure 3.

Discussion

Studies with the greatest potential to provide evidence are conducted through clinical evaluation. However, *in vivo* biomechanical measurement is limited by ethical and/or methodological aspects. Thus, finite element analysis (FEA) has been used extensively as a tool of functional assessment in implant research. This methodology consists of a mathematical model that is built based on the prosthesis, implant, and alveolar process geometries, the boundary conditions, and the material properties (Young's modulus and Poisson's ratio). The implant system performance is measured in specific values and by a gradient of stress/strain distribution in all structures of the model.[36-38]

The model generation and material properties in the present study were adapted to some simplifications and assumptions. Although all the structures were assumed to be isotropic, homogeneous, and linear elastic, it is known that these conditions do not occur in live tissues, such as the cortical bone, which is transversely isotropic and inhomogeneous.[32] The level of osseointegration considered was 100%, which also has been demonstrated to be incompatible with real conditions; however, studies have found that the analysis of non-linear frictional contacts and complete osseointegration of the bone-implant interface led to similar results.[39,40] The screw and implant thread were removed, as they were found to be irrelevant to the analysis after convergence tests and provided a relevant reduction in elements.[32]

In the current study, an increase in stress in the retention screw and in the framework was observed when stiffer materials were used for the prosthesis frameworks. These findings are in agreement with a previous study, which suggested that stiffer framework materials cause higher stress concentrations due to their lesser deformation.[21] However, according to the present study, it is possible to infer that this increase in the stress values does not comprise a clinical problem because the difference in the stresses is lower than the stiffness of the metallic structures. Regarding the retention screws, it has been suggested that the lesser deformation of stiffer frameworks that occurs during the closure of the misfit is responsible for transmitting greater stress to it. The lower stress values in porcelain veneer were observed when stiffer frameworks (Co-Cr and Zr) were evaluated, which is in agreement with a previous study.[33] This can be explained by the fact that, during prosthesis settlement, a less rigid material tends to suffer greater deformation, increasing the transference of stresses to the veneering material. In this way, materials with similar values of elastic modulus (Au, Ag-Pd, and Ti) did not present relevant differences in the stress distribution, probably because they have similar deformation capability. The higher stress concentration in the framework and porcelain veneer occurred in the cervical of the molar crown, more specifically in the mesial region that comes into contact with the implant platform, probably due to the rotational tendency of the prosthesis during the closure of the misfit, which cause stress concentration between these structures. The retention screw of the molar presented high stress values, which were observed in the long axis and probably caused by tensile forces.

A considerable increase in stress was observed in the framework, porcelain veneer, and retention screw proportional to the misfit increase, which is in agreement with previous studies.[12,21,31,34] It has been suggested that these structures are more sensitive to a lack of passive fit and are directly responsible for clinical failures such as loosening or fracture of the abutment or prosthetic screw and fracture of the framework or veneers[5,31,41] due to asymmetric contact among the various components of the system.[42-44] The relation of the vertical misfit and screw loosening has been established by previous studies.[7,45]

In the current study, the different stiffness of the framework materials did not demonstrate a relevant effect on stress values at the bone tissue surrounding implants, corroborating previous studies.[21,32-34] It is also been postulated that the viscoelasticity of bone compensates for any differential rigidity among the prosthetic materials.[34,46] Higher values of MPS were observed in the cortical bone, which can be explained because of its higher elastic modulus compared to the cancellous bone.[22,47]

The changes in vertical misfit showed a considerable influence on the stress values in the peri-implant bone tissue; this fact was also observed in other FEA reports.[12,30] Previous studies were performed using primates[19] and rabbits[20,48-50] aiming to evaluate the consequences of different levels of vertical misfit on the peri-implant bone tissue. However, these tests presented limitations, such as the impossibility of evaluating the influence of vertical misfit in the face of occlusal loads[29] and the evaluation of the consequences in the prosthetic structures, which are important factors in the success of the treatment. Other clinical studies have suggested the existence of a certain level of tolerance of bone tissue to a lack of passive fit in implant-supported prostheses. In a previous study, a misfit until 150 µm was considered acceptable,[51] and in another study, the mean misfit was 111 µm and 91 µm for 1-year and 5-year follow-up groups, respectively, which did not present a correlation with marginal bone level changes.[52] These studies were performed in edentulous patients rehabilitated with full arch prostheses supported by five to seven implants. However, the number of implants and the nature of the prosthesis (full, partial, or single) seem to be important factors in the stress distribution of implant-borne prostheses.[24] The misfit tolerance observed in full-mouth rehabilitations seems to be unacceptable for partial prosthesis supported by a minimal number of implants.

Based on these considerations, complementary studies evaluating the influence of occlusal load in the stress distribution of implant-supported partial prostheses are necessary to verify the behavior of ill-fitting prostheses under chewing conditions.

Conclusion

Considering the conditions evaluated in this FEA study, it can be concluded that:

- Stiffer materials promote higher stress concentrations in the framework, which increase proportionally to the stiffness of the materials.
- Stiffer frameworks increase the stress values in the retention screw, while in the same circumstances, the porcelain veneer shows lower stress values.
- The stiffness of the materials does not cause a relevant difference in the stresses in peri-implant bone tissue.
- A considerable increase in stress concentration was observed in all structures (framework, porcelain veneer, retention screw, and peri-implant bone) when the misfit was increased.

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Table 1 – Materials properties adopted in the study.

Material	Young's modulus (GPa)	Poisson's ratio
Cortical bone[21]	13.7	0.30
Cancellous bone[21]	1.37	0.30
Titanium (implant)[21]	110	0.33
Titanium (screw)[21]	110	0.28
Procera All-Ceran Zirconia[34]	269	0.25
Cobalt-chromium[21]	218	0.33
Commercially pure titanium[21]	110	0.28
Silver-palladium alloy[21]	95	0.33
Type IV gold alloy[21]	80	0.33
Vita VMK 68 (Porcelain veneer)[35]	70	0.19

Table 2 – Maximum Principal Stress (MPa) in the prosthesis framework.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	134.97	791.47	1,649.10
Ag-Pd	152.94	878.52	1,841.00
Ti	167.25	943.59	1,983.50
Co-Cr	274.64	1,457.00	3,093.80
Zr	312.37	1,642.70	3,458.50

Table 3 – Maximum Principal Stress (MPa) in the porcelain veneer.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	84.68	613.09	1,376.00
Ag-Pd	83.17	607.30	1,368.30
Ti	82.16	606.80	1,361.00
Co-Cr	74.27	564.26	1,243.10
Zr	71.78	546.44	1,211.60

Table 4 – von Mises Stress (MPa) in the screw.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	7.10	35.75	71.80
Ag-Pd	7.38	37.14	74.59
Ti	7.67	38.65	77.58
Co-Cr	9.18	45.97	92.17
Zr	9.56	47.85	95.90

Table 5 – Maximum Principal Stress (MPa) in peri-implant bone.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	11.49	57.25	113.90
Ag-Pd	11.93	59.43	118.26
Ti	12.11	60.36	120.12
Co-Cr	12.19	60.71	120.78
Zr	11.67	58.13	115.64

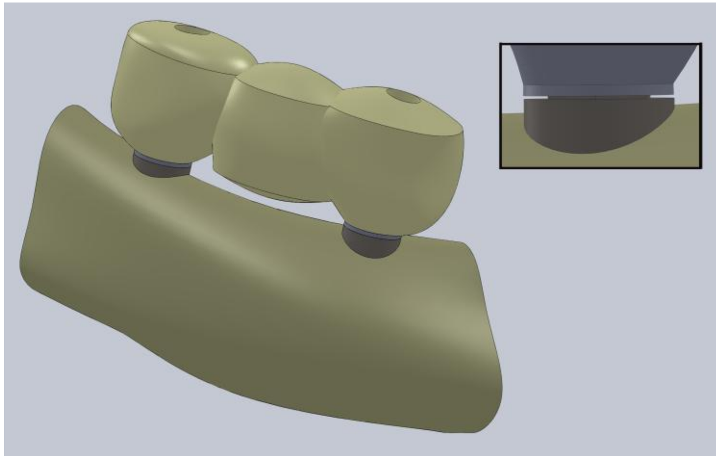
Figure 1. Configuration of the investigated models.

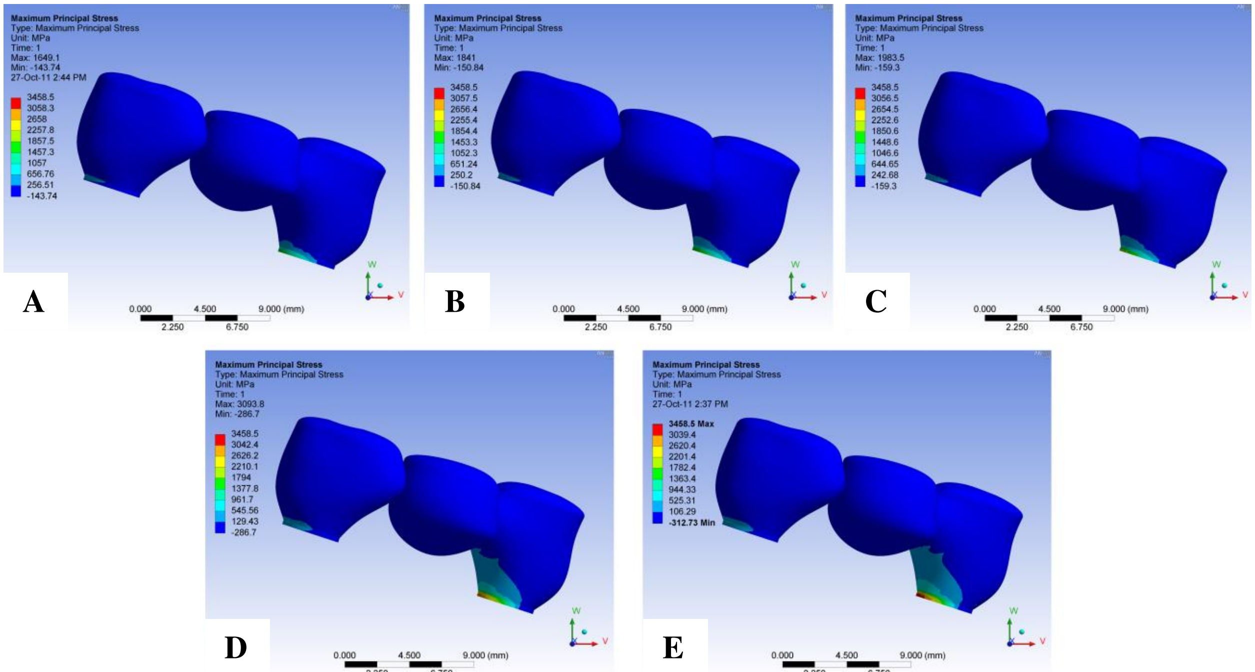

Figure 2. Maximum Principal Stress distribution in the frameworks with 100 µm of vertical misfit: (A) gold type IV alloy, (B) silver-palladium alloy, (C) commercial pure titanium, (D) cobalt-chromium alloy and (E) Zirconia.

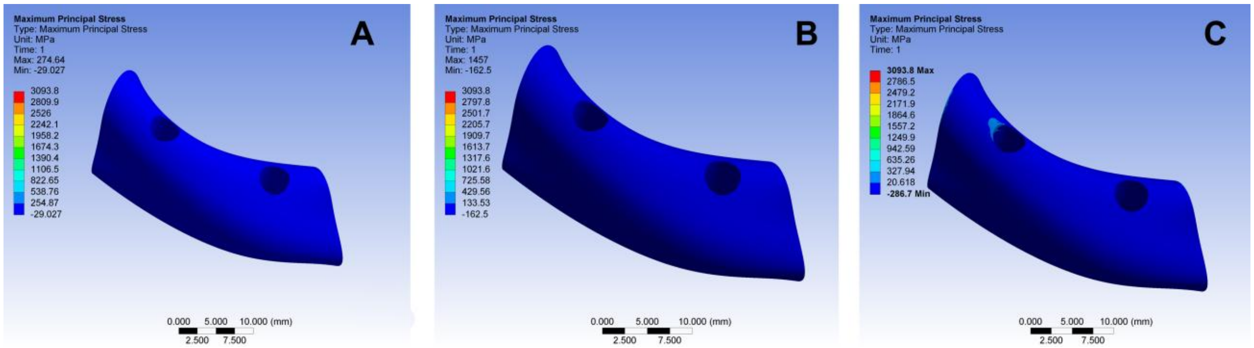

Figure 3. Maximum Principal Stress distribution in bone tissue with cobalt-chromium alloy framework in the levels of (A) 10 µm, (B) 50 µm and (C) 100 µm of vertical misfit.

CAPÍTULO 2

Effect of framework material and vertical misfit on stress distribution in implant-supported partial prosthesis under load application: 3-D finite element analysis

Abstract

This study evaluated the influence of framework material and vertical misfit on stress created in an implant-supported partial prosthesis under load application. The posterior part of a severely reabsorbed jaw with a fixed partial prosthesis above two osseointegrated titanium implants at the place of the second premolar and second molar was modeled using SolidWorks 2010 software. Finite element models were obtained by importing the solid model into an ANSYS Workbench 11 simulation. The models were divided into groups according to their prosthetic framework material (type IV gold alloy, silver-palladium alloy, commercially pure titanium, cobalt-chromium alloy, or zirconia) and vertical misfit level (10 µm, 50 µm, and 100 µm). After settlement of the prosthesis with the closure of the misfit, simultaneous loads of 110 N vertical and 15 N horizontal were applied on the occlusal and lingual faces of each tooth, respectively. The data was evaluated using Maximum Principal Stress (framework, porcelain veneer, and bone tissue) and a von Mises Stress (retention screw) provided by the software. As a result, stiffer frameworks presented higher stress concentrations; however, these frameworks led to lower stresses in the porcelain veneer, the retention screw (faced to 10 µm and 50 µm of the misfit), and the peri-implant bone tissues. The increase in the vertical misfit resulted in stress values increasing in all of the prosthetic structures and peri-implant bone tissues. The framework material and vertical misfit level presented a relevant influence on the stresses for all of the structures evaluated.

Keywords: dental implant, osseointegration, dental prosthesis, implant-supported, biomechanics, finite element analysis.

Introduction

A dental implant prosthesis usually consists of a framework with a veneering material (1, 2). Initially, gold alloy was the material most often used for framework fabrication, however, due to its high cost, alternative alloys were introduced in dentistry, among them cobalt-chromium, silver-palladium, and titanium alloys (3). More recently, the metal-free technology was implemented because of its chemical durability, aesthetics, and biocompatibility (1), which solve some of the problems observed in metal alloys, such as corrosion and esthetic limitations (2, 4).

It has been suggested that the material used for framework fabrication is very important for obtaining clinical success since it influences the biomechanics and propagating stresses during functioning, which could be transferred to the bone-implant interface, implant, prosthetic structures, and support components (5).

Furthermore, the longevity and success of a treatment depend on a passive fit at the implant-prosthesis interface (6). During the treatment and prosthesis fabrication, distortions can occur in all dimensions (x, y, and z) (7–9), caused by factors such as impression procedure, master cast fabrication (regarding technique and material), framework fabrication (waxing, casting, or machining), and final prosthesis fabrication (addition of veneering material) (10).

Many complications could be caused by a misfit in the prosthetic framework. These complications may include mechanical failures, such as fractures in veneering material, framework, fixation screws, and abutment screws, as well as loosening of the screws. Biological complications were also observed, such as gingival inflammation, pain, fistula, and peri-implant bone loss (9, 11); therefore, no longitudinal study has shown an implant failure attributed specifically to a framework misfit (6).

Previous studies were performed using primates (12) and rabbits (13–16) and aimed at evaluating the consequences of different levels of vertical misfit on the peri-implant bone tissues. However, these tests presented a considerable limitation: the impossibility of evaluating the influence of a vertical misfit during an occlusal load (17). Previous finite element analysis (FEA) evaluated the influence of the vertical misfit in an implant-supported partial prosthesis (18, 19) with overdenture retaining bars supported by two implants (3, 20) on the stresses transferred at the peri-implant bone tissues. Controversial results were observed in these studies, in that a considerable increase of stresses was observed in the peri-implant bone tissues with the misfit amplification in the implant-supported partial prosthesis; however, the increase of the misfit did not influence the values of the stresses in the peri-implant bone in overdenture retaining systems. Different framework materials were also evaluated on the stresses transferred to the prosthetic structures and peri-implant bone tissues in single crowns (21), fixed-partial prosthesis (2), and full-arch prosthesis (22); however, the presence of the vertical misfit, a clinical possibility, was not considered.

This study aimed at evaluating, through FEA, the influence of the framework material and different levels of vertical misfit on stress created in the prosthetic structures (framework and porcelain veneer), retention screw, and peri-implant bone tissues in an implant-supported partial prosthesis under loading conditions.

Materials and Methods

The posterior part of a severely reabsorbed jaw with a fixed partial prosthesis above two osseointegrated titanium implants (external hexagonal; 4.0-mm diameter x 10-mm length) was modeled using specific 3-D modeling software (SolidWorks 2010, SolidWorks Corp., Concord, Massachusetts, U.S.A.) starting from clinical data taken from a common situation. The implants were positioned at the right second pre-molar and second molar with 16.1 mm of distance between their centers. The implant threads were removed because, after convergence tests, they were found to be irrelevant to the analysis and caused a relevant reduction in the elements.

Finite element models were obtained by importing the solid model into mechanical simulation software (ANSYS Workbench 11, Ansys Inc., Canonsburg, Pennsylvania, U.S.A.). The models were divided into groups according to the prosthetic framework's material—type IV gold alloy (Au), silver-palladium alloy (Ag-Pd), commercially pure titanium (Ti), cobalt-chromium alloy (Co-Cr), or zirconia (Zr)—and the vertical misfit level (10 µm, 50 µm, and 100 µm) created at the second premolar implant/prosthesis interface. All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. The elastic properties used were taken from the literature (3, 23, 24) and are presented in Table 1.

Model stability was ensured to obtain a reliable model that was regarded as relevant in its engineering and clinical aspects (3). The total number of elements and nodes generated in the FE models were 736.750 and 1178.870 for 10 µm, 742.289 and 1187.188 for 50 µm, and 725.737

and 1160.223 for 100 μm of vertical misfit. The shape of the element was tetrahedral with 10 nodes. The investigated models produced the configurations in Figure 1. The stability of the model was checked, with particular attention paid to the refinement of the mesh resulting from the convergence tests at the bone/implant interface.

The base of the mandible was set as the fixed support, the settlement of the prosthesis with the closure of the vertical misfit was induced and loads were applied. Each tooth was loaded with simultaneous 110 N vertical and 15 N horizontal forces at the occlusal and lingual faces, respectively, with the aim of creating a resultant oblique load as has been previous reported (17). Data for the Maximum Principal Stresses (MPS; framework, porcelain veneer, and bone tissues) and von Mises stresses (retention screw) were produced numerically, color-coded, and compared among the models.

Results

Framework

An increase in MPS values in the framework was verified according to the stiffness of the evaluated materials. The misfit levels also caused relevant increases of the stress concentrations in the frameworks, which were potentially observed in stiffer materials. The higher stress concentrations occurred in the cervicolingual region that contacts the implant platform. All the stress values are presented in Table 2. MPS values in the frameworks with 10 μm of vertical misfit are presented in Figure 2.

Porcelain Veneer

There was a decrease of the MPS values in the porcelain veneer when stiffer frameworks were utilized. Amplification of the misfit induced relevant increases in the stress values. The maximum values of the stresses were observed at the cervicolingual region of the crowns, which is close to the frameworks' interface. All the stress values for the porcelain veneer are listed in Table 3.

Retention screw

The von Misses stress values occurred in the molar screw and decreased when stiffer frameworks (Co-Cr and Zr) were evaluated in the misfits of 10 µm and 50 µm. However, less stiff materials (Au, Ag-Pd, and Ti) did not present relevant differences in their stresses. The stiffness of the material did not cause a significant difference in the von Misses stress values when 100 µm of vertical misfit was evaluated. The increase of misfit levels promoted an increase of the stress values. Higher stresses were concentrated in the neck of the screw. The stress values for the screws in the different situations analyzed are presented in Table 4.

Bone stress

There was a relevant decrease in the MPS when materials with a higher stiffness were evaluated (Co-Cr and Zr). However, lower stiffness materials (Au, Ag-Pd, and Ti) did not present relevant differences among them. An increase of the stress concentration could be observed when the misfit levels were increased. The cortical bone in contact with the implant presented the higher values of stress concentration. The stress values for the different situations are presented in Table 5. The MPS in all levels of vertical misfit with a type IV gold alloy framework is presented in Figure 3.

Discussion

The FEA was utilized in this study and has been demonstrated and published as a suitable tool for implant research. This method consists of a mathematical model built based in prosthesis, implant, and alveolar process geometries and then boundary conditions and the material properties (Young's modulus and Poisson's ratio) are set according to each material. The performance of the implant system is measured in specific values and by a gradient of stress/strain distribution in all structures of the model, which could not be observed with different methods due to ethical and methodological limitations (25–28). However, this test does not completely replace a clinical or experimental study.

In this study, some simplifications and assumptions in the material properties and model generation were realized. The structures were assumed to be isotropic, homogeneous, and linear elastic. However, these conditions are not realistic for some materials and living tissues, such as cortical bone that is known to be transversely isotropic and inhomogeneous (21). Although the implants have been considered 100% osseointegrated, previous studies demonstrated that this does not match the real conditions (2). Other studies have shown that the results based on complete osseointegration and non-linear frictional contacts among bone implants are very similar (21, 29, 30). The screw and implant thread were removed because, after convergence tests, they were found to be irrelevant to the analysis and they provided a relevant reduction in elements.

In the present study, when stiffer materials were evaluated, a greater stress concentration in the framework was observed. These findings agree with previous studies (2, 21, 22) that attribute these outcomes to the fact that these materials are stiffer, more resistant to deformation, and concentrate high stresses values. However, according to the current study, this increase in the stress values does not constitute a problem, since the stresses increase proportionally according to the stiffness of the framework. Thus, although stiffer materials have high values of stresses, they are less vulnerable to fractures. A decrease in the stress value of the retaining screw was observed with stiffer framework materials (Co-Cr and Zr) faced to 10 µm and 50 µm of the vertical misfit. This data agrees with others' studies in which the authors suggest that the high resistance of the framework reduces the risk of mechanical overloading for the retaining screws (22, 23). However, materials with similar stiffness (Au, Ag-Pd, and Ti) did not demonstrate any relevant effect on stress values, probably due their closer elastic modulus. The present study also suggests that the stiffness of the frameworks have no relevant influence on stress values in the retention screw after a certain level of vertical misfit (100 µm).

Regarding the stresses in the porcelain veneer, lower values were observed when stiffer frameworks were evaluated, and these results that are in agreement with a previous report (22). This can be explained by the fact that less rigid material tends to suffer more deformation, increasing the transference of stress to veneering materials. That the higher stress concentration at the framework and porcelain veneer occurred in the cervicolingual region close to the implant platform and in the neck of the screw could be due to the horizontal force applied in a linguobuccal direction.

The data of the present study also shows the effects of vertical misfit on the framework, the porcelain veneer, and the retention screw. Previous reports showed a considerable increase of stresses in prosthetic frameworks and retention screws associated with vertical misfit increases (3, 19, 20, 23), and these findings are also verified by the current study. It has been suggested that these frameworks are sensitive to the lack of a passive fit due an asymmetrical contact among the various components of the system (31–33), which may be directly responsible for clinical failures such as loosening or fracturing of abutment or prosthetic screws, and fracturing of the framework or veneers (20, 34, 35). The effect of vertical misfit on screw loosening was evaluated by previous studies that found statistical correlation between the factors (36, 37).

According to some authors, the stiffness of the framework of an implant-supported prosthesis did not have any effect on stress values at the peri-implant bone tissue (3, 21–23), and these results were corroborated by the current study since materials with a similar stiffness were evaluated (Au, Ag-Pd, and Ti). A follow-up study on metal ceramic implant-supported prostheses postulated that the viscoelasticity of the bone compensates for any differential rigidity among resin, metal, and porcelain (23, 38), which was also suggested by this study regarding less rigid materials without a great stiffness discrepancy. However, there was a tendency of a decrease in the values of stressors in the peri-implant bone tissues when stiffer materials (Co-Cr and Zr) were utilized. It is possible to assume that due to the materials' capability to resist bending and to support more stress concentration leads to a lower transmission of stress to the peri-implant bone tissues. The MPS in the cortical bone was higher than that in the cancellous bone, which can be explained because of the latter's higher elastic modulus (23, 39).

The outcomes of this study demonstrated that the increase in the vertical misfit has a considerable influence on the stress levels in the peri-implant bone tissues, which was also observed by previous FEA reports (18, 19). However, clinical studies have attributed a certain level of tolerance of the bone tissue to the lack of a passive fit of the implant-supported prosthesis. Initially, Branemark (40) established that a misfit until 10µm can be considered as clinically acceptable. However, a later study suggested that a misfit until 150µm was considered acceptable (41), and in another study the mean misfits of 111µm and 91µm for the one- and five-year follow-up groups, respectively, did not show correlations with marginal bone level changes (42). Likewise, these studies were performed in edentulous patients rehabilitated with a full-arch prosthesis, supported by five to seven implants.

Previous reports pointed out that several factors influence the stresses on dental implants, such as the number of implants and the type of the prosthesis (full, partial, or single) (43, 44) and suggested that the misfits presented by these studies cannot be acceptable for a partial prosthesis supported by a minimal number of implants. Based on these considerations, clinical observations are necessary to evaluate the misfit's influence on an implant-supported partial prosthesis.

Considering the conditions evaluated by this FEA study, it can be concluded that (1) stiffer frameworks promote higher stress concentrations and the stresses increase proportionally to their stiffness; (2) stiffer frameworks promote lower stresses in the porcelain veneer, peri-implant bone tissue, and retention screw, yet the framework material seems to be irrelevant on the stress in the retention screw after an advanced level of the vertical misfit, and (3) the increase of the vertical misfit results in an increase of stress values in the prosthetic structures, retention screw, and peri-implant bone tissues.

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Table 1 – Materials properties adopted in the study.

Material	Young's modulus (GPa)	Poisson's ratio	Reference
Cortical bone	13.7	0.30	3
Cancellous bone	1.37	0.30	3
Titanium (implant)	110	0.33	3
Titanium (screw)	110	0.28	3
Procera All-Ceran Zirconia	269	0.25	23
Cobalt-chromium alloy	218	0.33	3
Commercially pure titanium	110	0.28	3
Silver-palladium alloy	95	0.33	3
Type IV gold alloy	80	0.33	3
Vita VMK 68 (Porcelain veneer)	70	0.19	24

Table 2 – Maximum Principal Stress (MPa) in the prosthesis framework.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	297.72	637.32	1,479.10
Ag-Pd	309.13	702.83	1,646.00
Ti	318.92	754.38	1,776.60
Co-Cr	366.82	1,155.30	2,766.20
Zr	386.91	1,318.20	3,110.90

Table 3 – Maximum Principal Stress (MPa) in the porcelain veneer.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	189.93	579.88	1,080.50
Ag-Pd	179.72	553.19	1,072.00
Ti	166.28	534.22	1,056.40
Co-Cr	124.64	419.57	1,045.20
Zr	120.04	406.83	1,030.70

Table 4 – von Mises stress (MPa) in the screw.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	92.45	105.17	130.52
Ag-Pd	90.64	105.11	131.70
Ti	89.29	105.45	133.44
Co-Cr	80.26	101.72	132.75
Zr	77.24	100.61	133.15

Table 5 – Maximum Principal Stress (MPa) in peri-implant bone.

Material	Misfit		
	10 µm	50 µm	100 µm
Au	64.46	102.75	159.54
Ag-Pd	63.20	101.90	158.81
Ti	62.32	101.05	156.44
Co-Cr	56.78	95.25	152.46
Zr	55.33	91.77	146.24

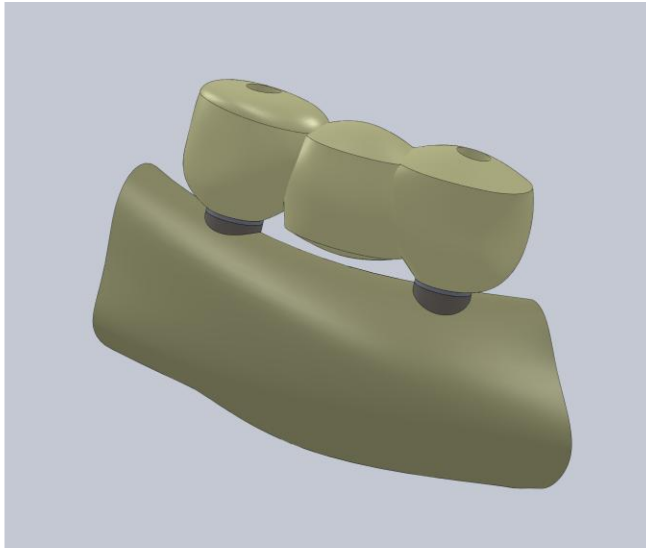

Figure 1. Configuration of the investigated models.

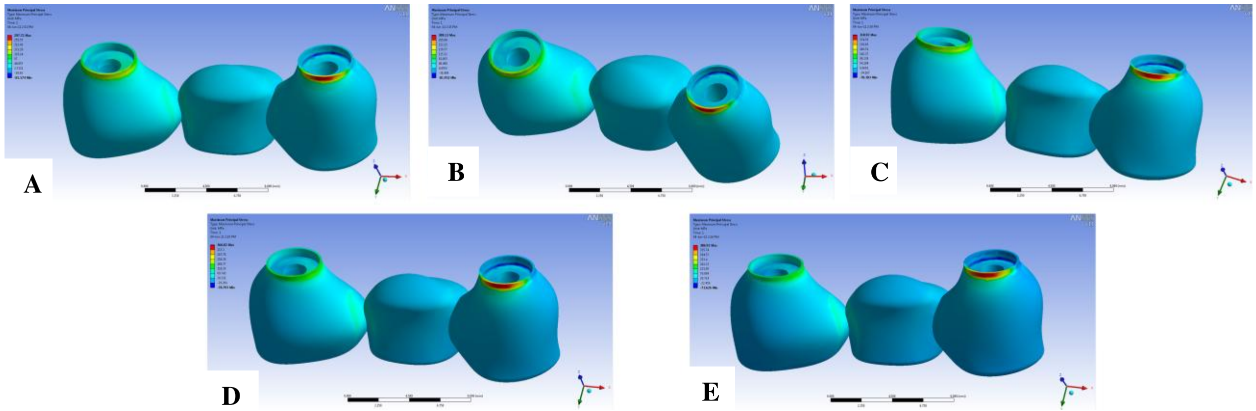

Figure 2. Maximum Principal Stress distribution in the frameworks with 10 µm of vertical misfit: (A) gold type IV alloy, (B) silver-palladium alloy, (C) commercial pure titanium, (D) cobalt-chromium alloy and (E) Zirconia.

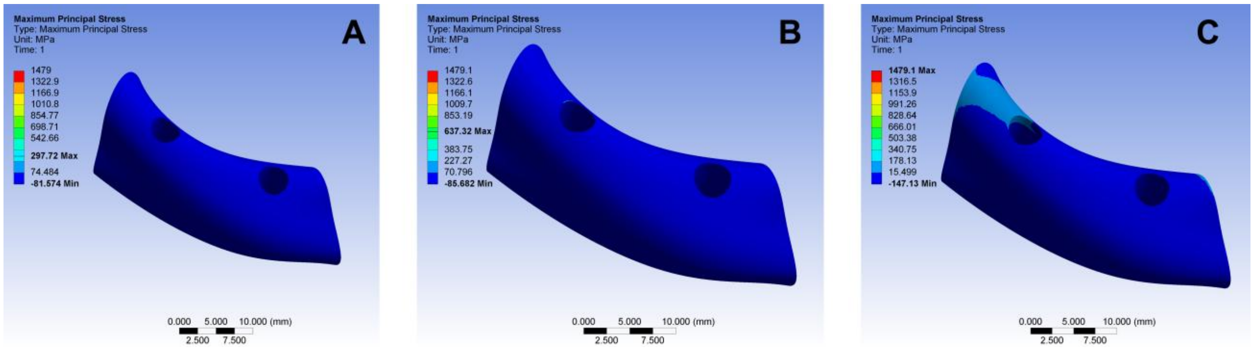

Figure 3. Maximum Principal Stress distribution in bone tissue with type IV gold alloy framework in the levels of (A) 10 µm, (B) 50 µm and (C) 100 µm of vertical misfit.

CONCLUSÃO GERAL

Dentro das condições avaliadas neste estudo, pode-se concluir que:

- Infraestruturas mais rígidas apresentam maior concentração de tensões internas; entretanto, causam menor concentração de tensão na porcelana de recobrimento.
- As infraestruturas mais rígidas causam menor tensão no parafuso de retenção e tecido ósseo peri-implantar quando carga oclusal é aplicada. Na condição estática, as infraestruturas com materiais mais rígidos aumentam a tensão no parafuso de retenção e não tiveram influência relevante quanto à tensão no tecido ósseo peri-implantar.
- O aumento do desajuste vertical promove considerável aumento da concentração de tensões em todas as estruturas (infraestrutura, porcelana de cobertura, parafuso de retenção e tecido ósseo peri-implantar).

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