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ORIGINAL ARTICLE

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

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Abstract

Objective. This study evaluated the influence of framework material and vertical misfit on stress created in an implant-supported partial prosthesis under load application. ***Materials and methods.*** 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 15 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. ***Results.*** 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. ***Conclusion.*** 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.

Key Words: *biomechanics, dental implant, dental prosthesis, implant-supported, finite element analysis, osseointegration*

Introduction

Dental implant prostheses usually consist of a framework with a veneering material [1,2]. Initially, gold alloy was the material most often used for framework fabrication, but, 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, esthetics 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

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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] and in over-denture 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 implant-supported partial prosthesis presented a considerable increase of stresses in the peri-implant bone tissues with the misfit amplification; however, the increase of the misfit did not influence the values of the stresses in the peri-implant bone in over-denture 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 × 10-mm length) was modeled using specific 3-D modeling software (SolidWorks 2010, SolidWorks Corp., Concord, MA) 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, PA). The models were divided into fifteen 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 I.

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.

Table I. 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]

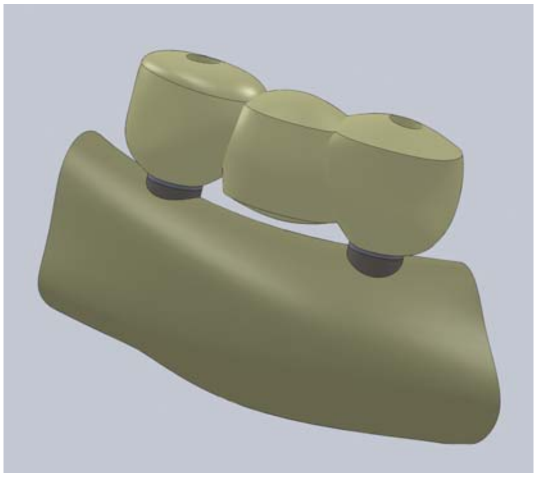

Figure 1. Configuration of the investigated models.

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 II. MPS values in the frameworks with 10 μm of vertical misfit are presented in Figure 2.

Table II. Maximum principal stress (MPa) in the prosthesis framework.

	Misfit		
Material	10 μm	50 μm	100 μm
Au	297.72	637.32	1479.10
Ag-Pd	309.13	702.83	1646.00
Ti	318.92	754.38	1776.60
Co-Cr	366.82	1155.30	2766.20
Zr	386.91	1318.20	3110.90

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

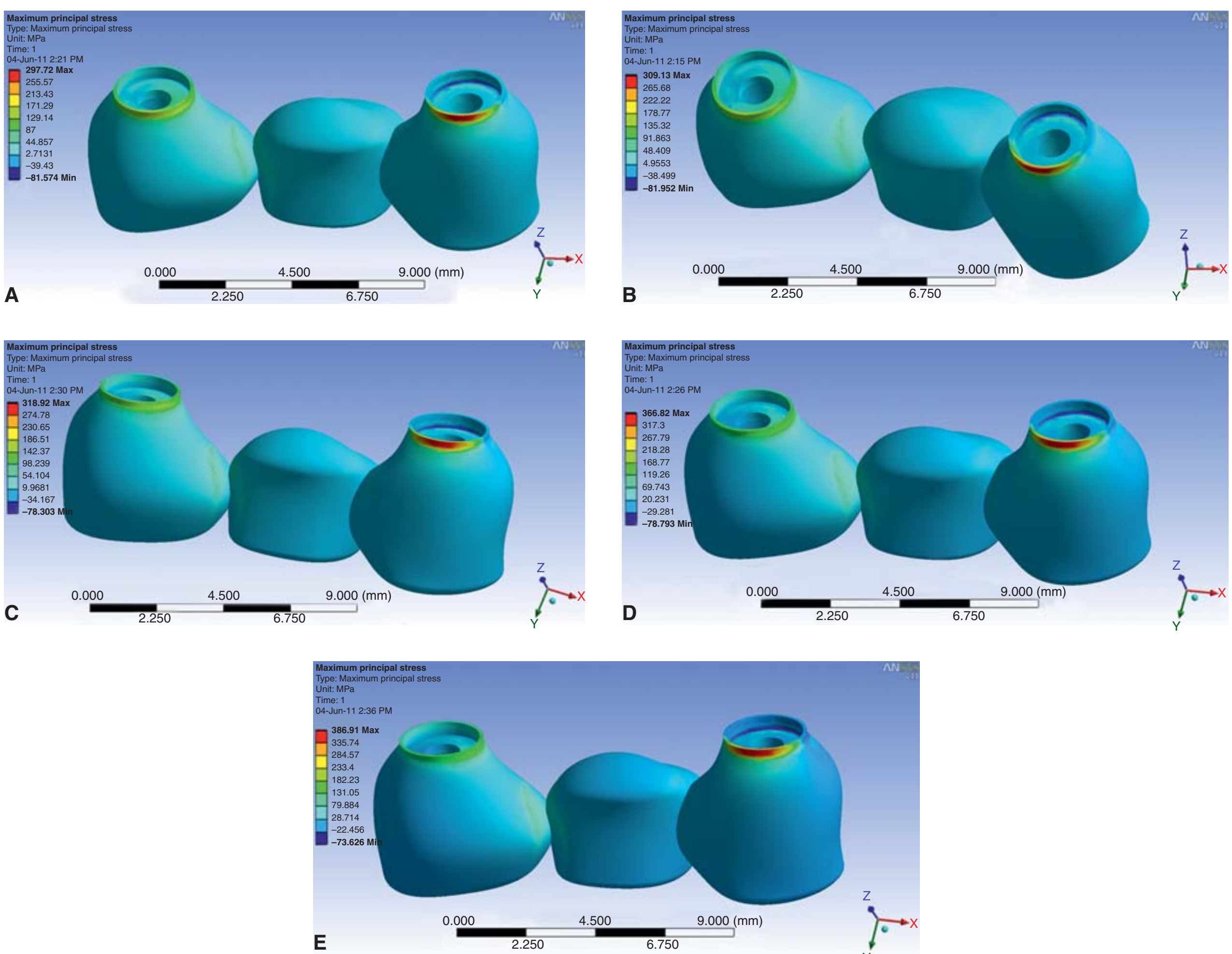

Figure 2. Maximum principal stress distribution in the frameworks with 10 μm of vertical misfit: (A) gold type IV alloy, (B) silver-palladium alloy, (C) commercial pure titanium, (D) cobalt-chromium alloy and (E) Zirconia.

Table III. Maximum principal stress (MPa) in the porcelain veneer.

	Misfit		
Material	10 μm	50 μm	100 μm
Au	189.93	579.88	1080.50
Ag-Pd	179.72	553.19	1072.00
Ti	166.28	534.22	1056.40
Co-Cr	124.64	419.57	1045.20
Zr	120.04	406.83	1030.70

frameworks' interface. All the stress values for the porcelain veneer are listed in Table III.

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 IV.

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 V. The MPS in all levels of vertical misfit with a type IV gold alloy framework is presented in Figure 3.

Table IV. von Mises stress (MPa) in the screw.

	Misfit		
Material	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

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 on 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 not more vulnerable to fractures. A decrease in the stress value of the retaining screw was observed with stiffer

Table V. Maximum principal stress (MPa) in peri-implant bone.

	Misfit		
Material	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

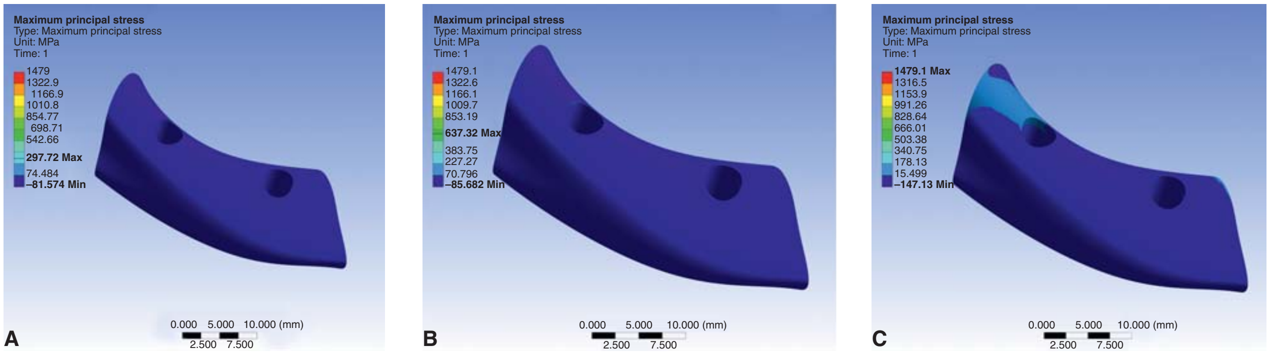

Figure 3. Maximum principal stress distribution in bone tissue with type IV gold alloy framework in the levels of (A) 10 μm, (B) 50 μm and (C) 100 μm of vertical misfit.

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 to 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 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 linguo-buccal 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 to 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 stresses 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 under 10 μm can be considered as clinically acceptable. However, a later study suggested that a misfit under 150 μm was considered acceptable [41] and in another study the mean misfits of 111 μm and 91 μm for the 1- and 5-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 might not 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.

Conclusion

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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