Influence of the Supporting Structure on Stress Distribution in All-Ceramic FPDs

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Purpose: The aim of this study was to investigate the influence of the design and material composition of the supporting structure of a zirconia four-unit fixed partial denture (FPD) on stress distribution during in vitro loading. **Materials and Methods:** A three-dimensional finite element model of an all-ceramic FPD ranging from the maxillary left first premolar to second molar was constructed. The supporting structures were modeled in four versions. In version 1, the socket and rigidly fixed abutment teeth were made of a nickel-chromium (Ni-Cr) alloy. Version 2 was similar to version 1 but abutment teeth were embedded resiliently. Version 3 replaced the Ni-Cr alloy with polyurethane as the material for the socket and abutment teeth. Version 4 was designed according to the in vivo situation with a simulated periodontal ligament, the socket consisting of spongiosa, and abutment teeth composed of dentin. An occlusal force of 1,630 N was distributed over the marginal ridges of the pontics. **Results:** The highest tensile stresses were located within the framework underneath the connector between the second premolar and first molar and ranged between 289 and 633 MPa, according to the model version. The resilient support of abutment teeth resulted in considerably higher maximum tensile stresses. **Conclusions:** The choice of material for abutment teeth and the socket, as well as the type of tooth support, significantly influence stresses generated in FPDs during in vitro load tests. To achieve realistic results, FPDs should be supported by resiliently embedded abutment teeth made of a moderately rigid material (eg, polyurethane). In clinical practice, risk of failure is likely to rise with an increasing resilience of the abutment teeth if occlusal contacts are directed over the pontic/connector region rather than being spread over the retainers. *Int J Prosthodont 2010;23:63–68.*

In recent years, high-strength ceramic materials have been substantially improved for use in dental restorations, and clinical follow-up studies with relatively short observation periods show promising results for treatment with all-ceramic fixed partial dentures (FPDs), even for posterior teeth.[1] Besides in vivo follow-up studies, further test methods are known for investigating longevity, fracture strength, and stresses in dental restorations. In most studies, the in vivo situation is simulated in vitro, but no standard in vitro test procedure has been developed for FPDs until now. A review of the literature reveals that cast bases used for supporting FPDs, for instance, are differently designed to such an extent that comparability of the results seems to be questionable. Besides natural teeth,[2,3] cast teeth made of alloys[4,5] and polymers[6] are used as abutments for FPDs. Furthermore, cast teeth are either fixed[7,8] or resiliently embedded[9–13] in their respective sockets. For simulating periodontal resilience, roots are generally covered with materials ranging from polyether material[9] to gum resin.[10,11] Kohorst et al[12,13] investigated the load-bearing capacity of four-unit all-ceramic FPDs

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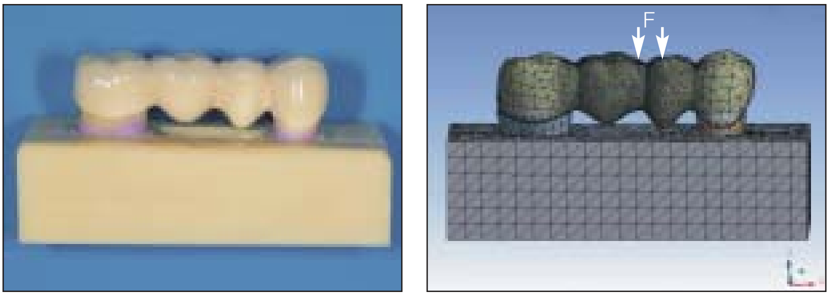

Fig 1 *(above)* **(left)** In-vitro cast and **(right)** virtual model (meshed) of the FPD with load application (F) in the direction of the arrows.

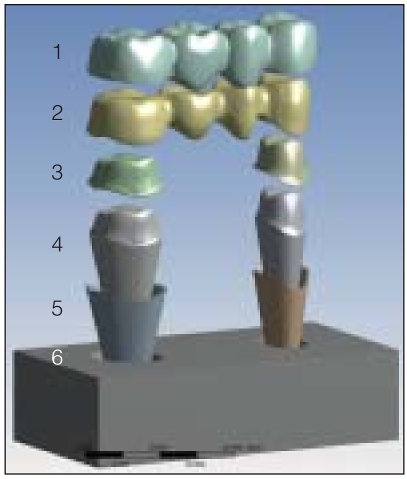

Fig 2 *(right)* Expanded view of the FPD model: **(1)** veneering layer, **(2)** zirconia framework, **(3)** cement layer, **(4)** model teeth, **(5)** resilient interface layer, and **(6)** socket.

with respect to the type of zirconia framework and cyclic fatigue in water. In both studies, the authors used polyurethane (PUR) cast teeth covered with a thin latex layer in the root area to simulate the periodontal ligament (PDL). These teeth were then embedded in a PUR socket to simulate bone support.

The second established method to analyze FPDs is by finite element (FE) analysis. Initially, axisymmetric FE models, which analyze the stress distribution in molar crowns for different types of preparations, were created.[14,15] No root or just a small portion of the root was modeled, presumably due to a lack of knowledge about periodontal resilience. In later years, two-dimensional FE models with a simulated root and PDL were created.[16–18] With increasing technologic advancements, more complex three-dimensional (3D) models have been developed, some considering roots and periodontal resilience,[19] and some considering neither.[20–23]

The aim of this study was to calculate stresses in an all-ceramic four-unit FPD undergoing in vitro load testing by way of FE analysis and to reveal the influence of different parameters on the resulting stress distribution. Two factors were to be considered: the choice of material for the socket and abutment teeth supporting the FPD in the experiment, and the method of embedding the abutment teeth in the socket (ie, rigidly or resiliently). Finally, the respective stress results would be compared with those obtained using an FE model approximating the in vivo situation.

Materials and Methods

A four-unit all-ceramic FPD spanning from the maxillary left first premolar to second molar, as used in previous experimental studies for determining load-bearing capacity,[12,13] served as a master for constructing the virtual model. The FPD was scanned optically (ATOS II SO, GOM) both before and after veneering and the surface coordinates were determined by means of triangulation. Prior to scanning, a thin antireflex layer of a special suspension with a grain size of approximately 1 µm was applied with an airbrush system. Afterwards, the resulting polygon meshes were transferred to 3D volume models by reverse engineering using a software program (PointMaster, Knotenpunkt). Then, the framework was virtually subtracted from the veneered FPD to obtain a separate model of the veneering layer (Cercon ceram S, Degudent). Connector cross sections of the frame were shaped elliptically, with areas of (mesial to distal) 12.5 mm^2, 15.6 mm^2, and 11.6 mm^2, respectively. The span between both abutment teeth amounted to 14.5 mm. Model teeth with layers of luting cement (approximately 100-µm thick) and, if applicable, a 300-µm-thick layer in the root area for simulating resilience of either the PDL or latex layer, were created (DesignModeler, ANSYS) and virtually embedded in a PUR block, according to the aforementioned experimental study (Fig 1).[12,13] The expanded view in Fig 2 shows the constituents of the virtual model.

A 3D FE model comprising 110,380 hexahedral and tetrahedral elements was created by a software-integrated meshing tool (DesignSpace, ANSYS), with an element size of 0.5 mm preselected for the framework and veneering layer in the connector area. All but one material constant was taken from the literature; Poisson ratio and Young modulus of the reinforced PUR (AlphaDie Top, Schütz Dental) were determined in a preliminary tensile test. By assigning appropriate elastic properties to the various model parts, four model versions were constructed. In version 1, the socket and rigidly fixed abutment teeth were made of a nickel-chromium (Ni-Cr) alloy. Version 2 was similar to version 1 but the layer in the root area was assumed to be resilient and consisting of latex. Version 3 was made in correspondence to the in vitro study with the socket and resiliently embedded abutment

Table 1 Material Constants in the FE Model*

Material	Young modulus (MPa)	Poisson ratio	Source
Zirconia	210,000	0.27	Munz and Fett[24]
Veneering ceramic	70,000	0.20	Manufacturer s data
PUR	3,525	0.33	Dittmer et al[25]
Glass-ionomer cement	15,900 (median)	0.33 (median)	Denisova et al[26]
Resilient lining	92	0.49	Calculated†
Ni-Cr alloy	200,000	0.30	Marxkors and Meiners[27] Suansuwan and Swain[28]
Dentin	18,300	0.30	Anusavice et al[29] Goel et al[30] Versluis et al[31]
Spongiosa	1,370	0.30	Meijer et al[32]
PDL	69	0.45	Farah et al[33]

PUR = polyurethane; PDL = periodontal ligament.
*Linear elasticity and isotropy assumed.
†Geometric mean of Young moduli for elastomer and PUR deemed to simulate the hyperelasticity of latex.

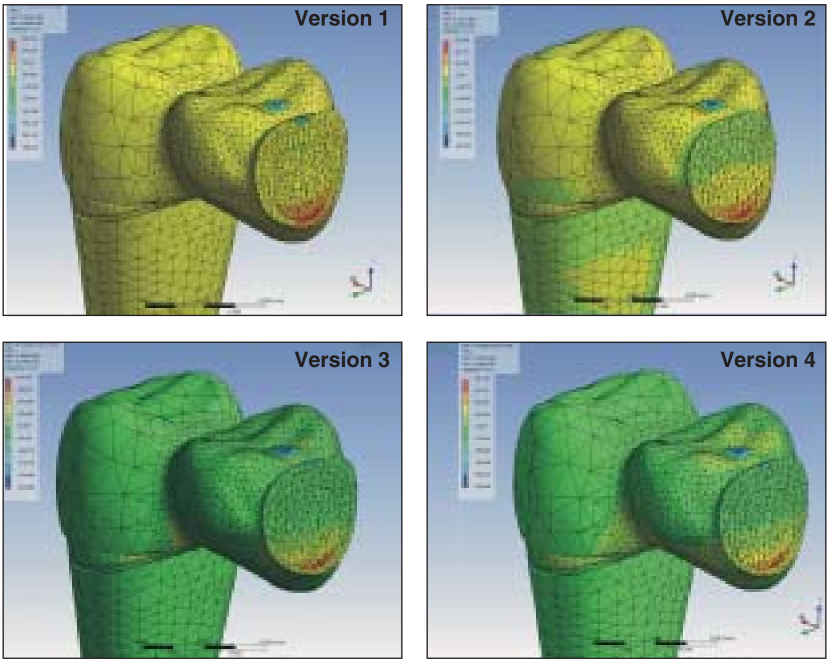

Fig 3 Overview of stress distribution on surfaces of the framework, resilient layer, and cross-section through the connector area between the maxillary left second premolar and first molar in versions 1 through 4. Note, the PUR block is left out for clarity.

teeth made of PUR. Version 4 approximated the in vivo situation with the socket consisting of spongiosa, abutment teeth composed of dentin, and a simulated PDL.

All material constants are listed in Table 1. The resilient lining of the tooth roots in the PUR socket for simulating the PDL behaves nonlinearly and gets more stiff the greater the load applied (hyperelasticity). Because of the software-given restriction to linear elastic calculation, hyperelasticity could only be approximated by assigning an effective modulus of elasticity, which was chosen as the geometric mean of the moduli of elastomer and PUR. By applying an occlusal force of 1,630 N evenly over two circular areas on the occlusal surface near the connector between the maxillary left second premolar and first molar, loading conditions were tailored according to the aforementioned in vitro study (Fig 1).[12,13,34] In the experiment, the force of 1,630 N led to failure of 63.2% of FPDs (Weibull analysis). In all simulated cases, the PUR socket was virtually fixed at the bottom to prevent any displacement. The connections between the different parts of the model were defined as being bonded. Formerly, in a series with FE models of increasing mesh density described elsewhere,[25] convergence of stress results had been verified, and the model delivering acceptable accuracy at the least computational effort possible was selected for use in this and other studies.

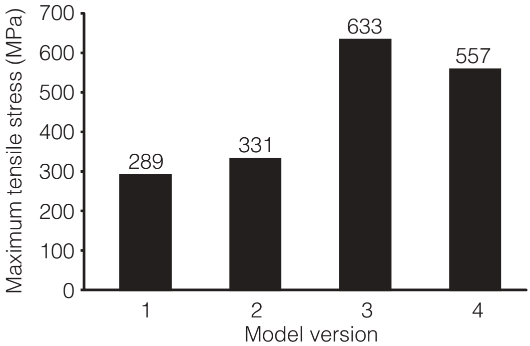

Fig 4 Highest maximum principal stress for the different model versions.

Results

In all model versions (1 through 4), the maximum tensile stresses within the entire structure developed close to the surface of the framework at the gingival embrasure of the connector between the premolars and molars (Fig 3). Generally, the peak tensile stress was higher when FPDs were supported more resiliently, either by introduction of a simulated PDL or by choice of a less stiff material for the abutment teeth and socket (Fig 4). Maximum tensile stress in the connector area between the second premolar and first molar rose from 289 MPa in version 1, to 331 MPa in version 2, up to 633 MPa in version 3, and decreased to 557 MPa in version 4.

In all four versions, maximum displacements were observed in the pontic region near the point of force application, mainly in the direction of force application with a slight shift towards the buccal side. Displacement increased from version 1 to 3 and amounted to 22 µm in version 1, 36 µm in version 2, and 115 µm in version 3, which simulated the in vitro model of Kohorst et al.[12,13] Simulating the in vivo PDL in version 4 revealed a maximum displacement of 104 µm in the pontic region and at the same time, an axial intrusion of the abutment teeth of approximately 110 µm.

As is evident from the cross-sectional view of the stress distribution in the connector between the second premolar and first molar, the stress gradient within the framework ran almost vertically from the basal to the occlusal side (Fig 3). Maximum principal stresses ranged from 633 MPa (tension, basal side) to –669 MPa (compresssion, occlusal side). The veneering layer unconsidered, maximum compressive stresses appeared in versions 1 and 2 at the distocervical dentin area of the premolar, while in versions 3 and 4 they appeared close to the point of force application in the connector area between both pontics (Fig 3).

Discussion

Formerly reported FE analyses dealing with FPDs have all identified the connector area as the part with the highest stress concentrations within the structure.[17,22,23,35,36] These findings compare well with the results of the present study. In all four versions, the location of maximum tensile stress was in the framework close to its surface at the gingival embrasure of the connector between the premolars and molars (Fig 3). In versions 1 and 2, maximum compressive stresses were found in the cervical dentin area near the residual ridge. This is in agreement with the results of Oruc et al[37] and Yang et al,[17] who also observed a stress concentration in this region. Additionally, the axial intrusion of the abutment teeth of 110 µm (version 4) corresponds quite well with physiologic tooth mobility, whose upper limit is in the order of 150 µm.[38]

As shown in Fig 4, an increase in maximum tensile stresses was observed when the FPD support was assumed to be more resilient. The use of a resilient elastomer layer in comparison to a rigid abutment tooth induced 14.5% higher tensile stresses (versions 1 and 2). Risk of failure rises with higher stresses, and tensile stresses in particular affect the crack growth in ceramics positively. As a result of these findings, it can be concluded that FPDs will also fail earlier in vivo if support is more resilient. Yang et al[17] investigated the need for modeling supporting structures and simulated bone loss, which resulted in higher movement of abutment teeth. They concluded that a loss of bone increases stress generated in the structures. That mobility and resilience of abutment teeth significantly influence stresses is in agreement with the findings of the current study (Fig 4). Supporting these results, Molin et al[39] showed, by 3D FE analysis of a three-unit FPD, that a simulated PDL induces 40% higher stress values than a nonligament model. Rees[40] found much higher variations in stresses when varying the supporting structures and pointed out the need to model both the PDL and alveolar bone when undertaking FE analysis of teeth. However, it has to be mentioned that these two cited studies simulated the in vivo and not the in vitro situation.

Different in vitro studies have also investigated the influence of abutment tooth resilience on the load-bearing capacity of FPDs. Major differences in materials used to simulate the resilient layer are obvious. In two studies, a polyether layer was used,[41,42] while another two[43,44] used a material based on silicon. All four studies concluded that load-bearing capacity is considerably lower when the support is more resilient. This clearly underlines the results of this study. Only Kern et al[10] found that a more resilient support does not affect the load-bearing capacity of FPDs.

A further aspect evaluated in the current work was the influence of abutment tooth material on stress distribution in FPDs. As seen in Fig 4, the use of bone, dentin, and PDL material constants in version 4 led to 68.3% higher stress than the simulation with abutment teeth made of Ni-Cr alloy and an elastomer layer (version 2). Hence, abutment tooth material seems to have a great influence on stresses in loaded FPDs, and so it is of crucial importance to consider it when experimentally simulating the in vivo situation. There are some disadvantages to using natural abutment teeth regarding restrictions in reproducibility and comparability between various specimens. To avoid these disadvantages, Kohorst et al[12,13] generated a model with artificial abutment teeth made of reinforced PUR, which has an elastic modulus lower than dentin but similar to bone. This experimental situation served as the guideline for version 3 of the FE model presented in this study. As shown in Fig 4, maximum tensile stress developed in version 3 is only 14% higher in comparison to version 4, which simulates the in vivo situation. Therefore, the in vitro situation is acceptably approximated by version 3. In contrast, version 2, using a high modulus material for the abutment teeth and socket, deviates further from the in vivo situation with respect to stresses in the framework. This leads to the suggestion that a moderately rigid material (PUR) be used for model teeth and the socket in in vitro load tests of FPDs. This additionally provides a 10% safety margin by creating higher stresses in the FPD, resulting in a correspondingly lower load-bearing capacity in the experiment compared with the in vivo situation.

Another important factor affecting stress distribution and hence, load-bearing capacity, is the method of loading the FPD.[25,45] While evenly distributed occlusal forces are the ideal case, the more or less concentrated loads applied at the neighboring rims of the pontics in the in vitro experiments by Kohorst et al[12,13] as well as in the FE analysis of this study represent a worst case scenario, with higher stresses in the FPD than under usual clinical loading conditions.

FE analysis is an approximate method for simulating the behavior of structures under a load. Naturally, the model could only consist of a finite number of elements. The assumption of linearly elastic behavior in the case of latex was an additional approximation, which was at least partly compensated for by the introduction of an effective elastic modulus. Finally, residual stresses generated by thermal loading during veneering have not been considered. These stresses are in the order of 80 to 110 MPa[46] and would have to be superposed to the ones created by mechanical loading. Nevertheless, comparison of the stress results obtained for the different model versions under equivalent conditions is possible, and gives an appraisal of the influence of the supporting structures on load-bearing capacity.

Conclusions

Based on the findings of this study, it can be concluded that abutment tooth and socket material, as well as the type of tooth support, have a significant influence on stresses generated in FPDs during in vitro load tests. To produce stresses in the FPD as close as possible to those of the in vivo load case, it is particularly important to use resiliently embedded abutment teeth made of a moderately rigid material (eg, reinforced PUR) to support specimens. Transferring these results to the in vivo situation means that risk of failure of an FPD is likely to rise with increasing resilience of the abutment teeth, especially if occlusal contacts are directed over the pontic/connector region rather than being spread over the retainers.

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