

Influence of Connector Width on the Stress Distribution of Posterior Bridges under Loading

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Abstract:

Objective: In all ceramic fixed partial dentures the connector area is a common fracture location. The survival time of three-unit fixed partial dentures may be improved by altering the connector design in regions of maximum tension. The purpose of this study was to determine the effect of buccolingual increase of the connector width on the stress distribution in posterior fixed partial dentures made of IPS Empress 2. To simulate the anatomical condition, we used three-dimensional finite element analysis to generate.

Materials and Methods: Three models of three-unit bridges replacing the first molar were prepared. The buccolingual connector width varied from 3.0 to 5.0 mm. Bridges were vertically loaded with 600 N at one point on the central fossa of the pontic, at 12 points along the cusp-fossa contact (50 N each), or at eight points along the cusp-marginal ridge contact (75 N each). Alternatively, a load of 225 N was applied at a 45° angle from the lingual side.

Results: Stress concentrations were observed within or near the connectors. The von Mises stress decreased by increasing connector width, regardless of whether the loading was applied vertically or at an angle.

Conclusion: Within the limitations of this study, we conclude that increasing the connector width decreases the failure probability when a vertical or angled load is applied.

Key Words: Ceramics; Finite Element Analysis; Denture, Partial, Fixed

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INTRODUCTION

Clinical stress distribution in ceramic dental restorations may be quite complex. Several factors are associated with crack initiation and propagation, including the shape of the restoration, micro structural nonhomogeneities, the size and distribution of surface flaws, residual stresses and stress gradients induced by polishing and/or thermal processing, the environment in contact with the restoration, ceramic/cement interfacial features, thickness and thickness variation of the restoration, different

elastic modules of the restoration components and the magnitude and orientation of the applied load [1]. Failure may be a mixed pattern of compressive, tensile, and/or shear failures [2]. However, brittle materials (e.g., dental ceramics) usually demonstrate the least strength when exposed to tensile stress. Thus, tensile stress is useful for assessing the service time of ceramic materials, especially when flaws are present. The long-term mechanical behavior of the restoration is also critical, since ceramics exhibit a time-dependent strength decrease due

to subcritical crack growth.

Of the structural factors, the connector areas are the most influential in failure. Failure rate is relatively high in three unit all-ceramic bridges around the sharp connector area [3,4]. The FPD shape is not uniform clinically, but is a complex combination of multiple convexities and concavities that depend on the geometry and alignment of the teeth [5,6]. In all ceramic resin-bonded FPD the occlusogingival height of the interdental connector must be as large as possible (minimum 4.0 mm) [7]. Furthermore, the connector area is usually narrowly constricted for biological or esthetic reasons, which typically considers stresses relative to the average stress levels in other areas of the prosthesis. The minimal recommended connector cross section area is 12-16 mm² [2,4,8]. Previously, it had been hypothesized that fracture initiation sites in dental ceramics could be controlled by changing the ceramic thickness. However, it is now believed that ceramic thickness plays a secondary role in fracture initiation, and that critical flaws in regions of reduced thickness are generally more important [1]. Several studies have analyzed the stress distributions in FPDs. One such study investigated distal cantilevered FPDs with differing cantilever morphologies made of two different restorative materials, where the width of the curved connector between the cantilever and primary abutment restoration was 2.25 mm. The average von Mises stress values revealed a higher stress at the occlusal embrasure of the connector between the pontic and second premolar abutment compared to the cervical embrasure [9]. In another study, the occlusal and gingival embrasures of connectors were reported to be the areas of highest stress [10].

The purpose of the present study was to determine the influence of buccolingual increase of the connector width in posterior FPDs made of Empress 2 core ceramics on stress distribution. The finite element method (FEM) was used to determine the optimal stress distribution in the ceramic bridges that would reduce the risk of connector fracture.

MATERIALS AND METHODS

Using patient computed tomography scans, geometries of the second premolar, the first and second molars were determined. To make an initial model of the teeth for data pre-processing, we used the Mimics™ software (Ver. 8, Materialise, Leuven, Belgium) to convert two-dimensional scan reformatted views to a three-dimensional (3-D) model. The model is an STL model of objects to be studied, and must therefore be clean and free of inaccuracies. The STL file was imported to another software (Magics™ Ver 9, Materialise). Because Magics has a limited ability to design novel parts, a primary surface mesh model was converted to the Solid Works 2004 format for redesigning and assembling. After modeling, the STL file was used for analysis with FEM software (ANSYS™ Ver. 9) (Fig 1).

Using Solid Works, the second premolar and first molar teeth were prepared with a convergence angle of 12°. The finishing lines were prepared in shoulder shape with rounded internal angles (Fig 2). The prepared teeth exhibited smooth and regular surfaces joined by rounded angles and were used to make an FPD. Three different FPDs with various connector sizes were prepared with the connector shape and location being similar to the actual clinical shape and position. The occlusogingival and mesiodistal dimensions of the con-

Table 1. Material Properties.

Material Type	Poisson's Ratio	Young's Modulus (MPa)
Dentin	0.15	1.4×10 ⁵
IPS Empress	0.3	1.02×10 ⁴

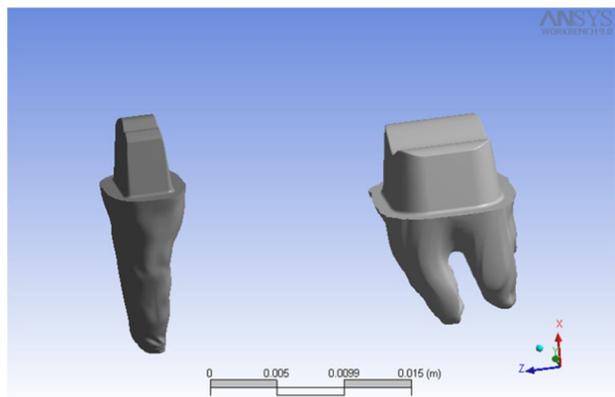


Fig 1. Bridge modeling.

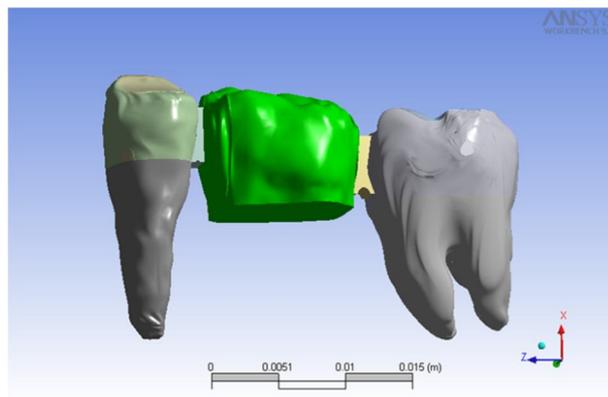


Fig 2. Preparation of teeth with a 12 taper.

connector were 4.0 mm and 0.3 mm, respectively. The connector widths used were 3.0 mm (model A), 4.0 mm (model B), and 5.0 mm (model C). The arc of the connector on the gingival side of the premolar region has a smaller radius, but in the molar region the radius of curvature was 0.45 mm at the gingival embrasure [11].

The material properties used are shown in Table 1 and are based on a 3D FEM analysis of all-ceramic posterior crowns [12]. The pontic were assumed to be subjected to 225 N of load acting at an angle of 45°, with the tooth axis on the side. A load of 600 N was vertically applied in one of three ways: at one point on the central fossa (case 1), at eight points along the cusp-marginal ridge contact area (75 N, case 2) or at 12 points along the cusp-fossa contact area of the pontic (50 N, case 3).

RESULTS

Inspection of the results revealed critical zones with particular stress behaviors. Using Ansys Ver. 9, we computed the von Mises stresses during 600 N of vertical loading or 225 N of loading at a 45° angle on the lingual surface of

the pontic (Fig 3, 4). The von Mises stresses could not be distinguished into specific contributions of tensile, compressive, and shear stresses. We therefore, investigated the mechanical behavior of E2 bridges as functions of the direction and magnitude of the maximum von Mises stresses generated by lateral loading with 225 N. These results are tabulated in Table 2.

The stress levels descended by increasing the connector width, from model A (3.0 mm) to C (5.0 mm). The maximum stresses in all three models were found on the connector between the premolar and the pontic (Table 2). Analysis of the von Mises stresses for all models with a vertical loading of 600 N revealed a higher stress on the connector area between the premolar and pontic. The highest stresses were located on the gingival embrasure of this connector. The stress gradient decreased from model A to C in all three cases.

In contrast, when 600 N was distributed over 8 points along the cusp-marginal ridge (case 2, 75 N each) or over 12 points along the cusp-fossa (case 3, 50 N each), the maximum von Mises stresses occurred in the gingival aspect

Table 2. Von Mises stresses or equivalent (45°) stress under 225 N.

Model	Maximum	Location of Maximum
Equivalent stress model A	1.002×10^8	Connector between premolar and pontic
Equivalent stress model B	0.667×10^8	Connector between second molar and pontic
Equivalent stress model C	0.450×10^8	Connector between premolar and pontic

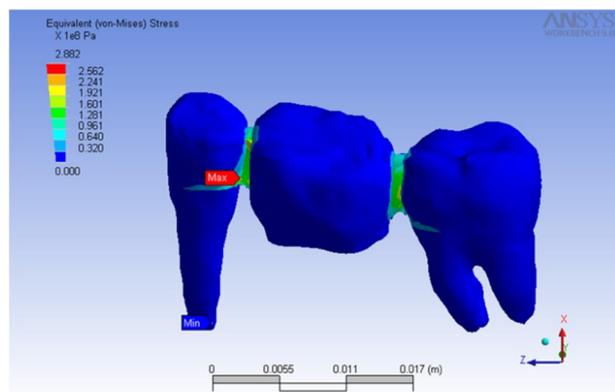


Fig 3. Equivalent (von Mises) stress under a vertical load of 600 N, Model A (case 1). Color range is used to demonstrate the von Mises stress from lowest (blue) to highest (red) concentration.

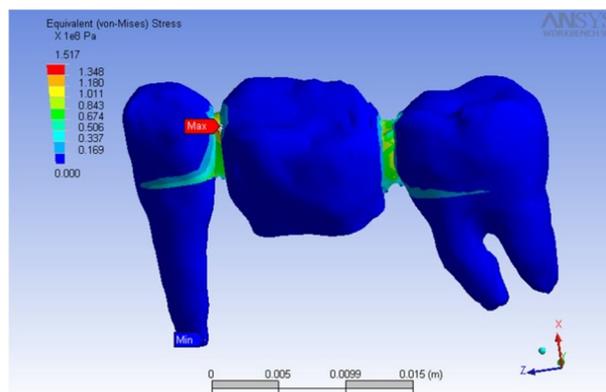


Fig 4. Equivalent (von Mises) stress under an angled load (45° lingually) of 225 N, Model A. Color range is used to demonstrate the von Mises stress from lowest (blue) to highest (red) concentration.

of the connector between the second premolar and the pontic. In models A and B at 600 N, the lowest von Mises stresses were seen under the conditions of case 2. In model C, the lowest von Mises stress was observed in both cases where multiple points were used (cases 2 and 3) (Tables 3). Location of maximum von Mises stresses under 600 N of vertical loading over one or more points was connector between premolar and pontic.

DISCUSSION

Here we examined the effect of increased connector width on the stress distribution in posterior FPDs. The points of greatest stress were found within or near the connectors. Furthermore, the von Mises stresses increased with increasing connector width, both when the loading was applied vertically and when it was applied at an angle. Ceramic bridges exhibit better esthetics and excellent biocompatibility compared to other bridge materials, but they have limited loading capability. Wide span bridges may have increased deformation, due

to the higher forces normal to the posterior regions of the oral cavity. The limited loading capability may also influence the limited strength and low fracture toughness of ceramic materials. However, measuring the strength of a ceramic restoration in the oral environment or through laboratory testing is difficult. The stress distribution induced by an experimental load is complex, and it is difficult to induce a purely compressive, tensile, or shear stress in a test environment.

Another approach for strength determinations is FEM, which has been used extensively in industrial applications, as well as in dental applications to study fracture mechanics, implants and dental restorative devices, and various dental materials. The FEM is a mathematic analysis that can predict strength values; however, the results may differ from those determined in oral or laboratory environments [13]. In the present study, we utilized an FEM where the underlying structure was dentin, and influences due to the pulp, periodontal ligament, or the adhesive or cement layer were not

Table 3. Maximum von Mises stress under 600 N of vertical loading over one or more points.

Model	Case 1 (Pa)	Case 2 (Pa)	Case 3 (Pa)
Equivalent stress model A	2.1×10^8	2.1×10^8	1.7×10^8
Equivalent stress model B	1.4×10^8	1.3×10^8	1.22×10^8
Equivalent stress model C	1.04×10^8	0.6×10^8	0.9×10^8

considered [14]. The materials were assumed to be homogeneous, isotropic and linearly elastic. We used data originally acquired from the computed tomography scans of a live object. Models of the proposed teeth were created from data cropped from based DICOM data. The bridge configuration was modeled based on the geometry of the natural teeth.

Changes in prosthesis component contours, particularly abrupt ones, may affect the stress distribution in a ceramic prosthesis. Local stresses can significantly increase the overall stress at highly curved regions, such as at surface notches or other abrupt shape changes. This effect may be much more significant in a brittle material, such as ceramic, that contains many small flaws or cracks of various sizes and orientations. These factors may also be more critical in posterior FPDs. Posterior areas experience higher loads, and the connector height may be limited by the short clinical molar crowns [11]. Cracks are initiated adjacent to load points and propagate along the plane of maximum tensile stress to the gingival side of connectors [15]. The failed surface displayed some pores and incomplete crystallized or densified areas [8].

The reported results of any FEM are the normal and shearing stress values of the structure upon loading. However, interpreted alone, these values do not provide a comprehensive solution to the problem of failure. Especially in 3D analysis, where estimation based on a single parameter is virtually impossible, various interpretation methods should be used. One alternative approach is to use failure criteria, which combine the effects of each of the normal and shearing stresses into a single, physically meaningful parameter, providing more sound analyses. The present study was conducted using the 3D von Mises criteria. The primary normal stresses act on the principal planes, on which the shearing stresses are zero. The structural configuration and loading may result in all positive, all negative, or a combi-

nation of positive and negative principal stresses; however, the von Mises stress is always positive. Von Mises criteria, which result in a tensile-type normal stress, were chosen because the brittle tooth material primarily fails due to tensile-type normal stresses [16]. A negative stress value would represent compression and a positive value would represent tensile stress [13].

The height of the mastication force significantly influenced our results. The average chewing force in literature varies between 11 and 150 N, whereas force peaks are 200 N in the anterior, 350 N in the posterior and 1000 N with braxism. Adding a 30% safety loading buffer results in requirements of 300 N for anterior application and 500-580 N total for an average person [17,18]. Mastication chewing forces are also reported to be equivalent to 37% of a maximum bite force of 600 N [18]. Based on this previous study, we choose 600 N for the maximum bite force and 225 N for the chewing force.

Under a vertical load of 600 N applied at one point, the von Mises stress gradient decreased from model A to C. In case 3 (loading at the cusp-fossa contact) and case 2 (loading at the cusp-marginal ridge contact), the von Mises stress decreased from model A to B and C. Under an applied load of 225 N at 45°, the von Mises gradient was similar to that achieved under 600 N at one point. We modeled the teeth as natural teeth, with the same concavity and convexity as in the mouth. Loading was done at one point, either vertically or at 45° to the lingual surface. The stress distribution was not equal under the two scenarios, since the connector parts were not in a defined geometric shape.

Lüthy et al [8] showed that when mastication forces of 500 N are applied to bridges with a connector cross-sectional area of 7.3 mm², four-unit frameworks made of IPS Empress 2 display a 100% failure probability. The flexural strength of E2 is 215 MPa (SD=40) [10]. In

our study, the connector areas used were 12 mm², 16 mm², and 20 mm². The minimal recommended connector cross-sectional area is 12-16 mm², indicating that increasing the connector height from 3.0 to 4.0 mm dramatically reduces the stress levels within the connectors [6,19]. Oblique stress application amplifies the resolved stresses and alters stress distributions within structures, allowing bridges to apply stress at points of natural convexity and concavity. In the present study, the stress distribution was not equal throughout the structure, and the stress gradient decreased from model A to B to C. The maximum von Mises stress under a 600 N load was found in model A (3.0 mm connector width) in all three cases. No differences were observed between cases 1 and 2.

The 3D model enabled us to investigate the stress distribution within and along any section of the bridge or the abutment teeth. We modeled the connectors between the premolar, pontic, and second molar as separate sections; although in reality they act as a bounded section to the premolar crown core. Thus, we observed some concentration of stress at the edge of the connector where it bound to the core. In the laboratory, the connectors are made in one piece with crowns and pontic, with no stress concentration at this point. We therefore ignored the high stress concentration in these very small places.

All six faces of the connectors and cores were inspected and the maximum stress was found to be in the gingival aspect of the connector between the premolar and pontic. An *in vitro* study revealed an oblique orientation of the fracture path extending from the gingival embrasure to the occlusal contact area [11]. Because the failure mechanism is influenced by the contact area and load during function, the size of the ball used for loading is an important consideration in an *in vitro* study. A large-diameter steel ball was used in this study to develop as clinically relevant of contacts as

possible. Three uniformly distributed contacts were established on the inclines of the triangular ridges around the central fossa of the pontic to prevent possible failure by localized impact [8], which is not common in clinical situations. Few specimens exhibited localized fractures at the contact area with the steel ball. This may partly explain the rare occurrence of a cone crack in the oral environment and it may relate to the large contact area achieved with the large-diameter steel ball. In this study, we used the cusp-fossa and cusp-marginal ridge relationships to distribute the force and avoid stress concentration at one point.

Oh and Anusavice [11] suggested that the fracture probability may be significantly reduced by using a connector with a curvature radius of approximately 0.9 mm. To reduce the stress concentration and to maintain a constant connector height of 4.0 mm without altering the curvature at the gingival embrasure, they suggested a gingival embrasure curvature radius of 0.45 mm. This propagates the cracks from the gingival embrasure toward the occlusal loading on the pontic. The fracture origin was most commonly at the center of the gingival embrasure in the buccolingual dimension and was shifted slightly toward the abutment crown in the mesiodistal direction. We likewise modeled the gingival embrasure between the pontic and second molar with a curvature radius of 0.45 mm. However, we could not use the same curvature in the gingival embrasure of the connector between the pontic and second premolar, because of the 0.3 mm connector thickness and the morphology of the natural-like premolar. In our study, high stresses were observed in the gingival embrasure between the second premolar and pontic and toward the gingival margin of the premolar abutment, consistent with Oh and Anusavice [11].

Eraslan et al [9] applied loads on the abutment teeth and cantilever of distal cantilevered FPDs designed with varying cantilever mor-

phologies and restorative materials. They found that the stress was concentrated in the cervical region of the second premolar and the connector between the second premolar and the cantilever in the gingival and occlusal embrasures. However, they had modeled the periodontal ligament, which we did not do. In contrast, we applied a 600 N load on the pontic. If we had instead divided the 600 N load on the abutments and pontics as occurs in reality, the stress distribution would have been very desirable and the stress concentrations would have been lower than we observed.

CONCLUSION

In conclusion, within the limitation of this study, we found that increasing the connector width decreases the failure probability when a vertical or angled load is applied. This understanding of the role of the connector width may aid in the design of dental restorations with reduced failure probabilities.

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