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RESEARCH AND EDUCATION

Flexural strength of small connector designs of zirconia-based partial fixed dental prostheses

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ABSTRACT

Statement of problem. Partial fixed dental prostheses with a small connector size are required for optimal esthetics and limited interarch space; however, final strength is endangered.

Purpose. The purpose of this in vitro study was to evaluate the effect of different connector designs on the flexural strength of simulated 3-unit partial fixed dental prostheses made of yttria-stabilized tetragonal zirconia polycrystalline using computer-aided design and computer-aided manufacturing technology.

Material and methods. To simulate a 3-unit partial fixed dental prosthesis, 20 rectangular bar-shaped specimens were fabricated with dimensions of 4 ±0.05 mm (H)×4 ±0.05 mm (W)×30 0.5 mm (L). Each bar specimen had 2 constricted parts on both sides, representing the connector and defining a central pontic of 10 ±0.10 mm in length. The specimens were divided into 4 groups according to the connector diameter and design, as follows: SR: 2 mm (H)×3 mm (W) round 0.6 mm radius of curvature; SS: 2 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature; CR: 3 mm (H)×3 mm (W) round 0.6 mm radius of curvature; and CS: 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature. An additional 5 specimens were fabricated with no constriction and served as the control group. The specimens were subjected to a 3-point flexural strength test in a universal testing machine with a crosshead speed of 0.5 mm/min until failure. Scanning electron microscopic and photomicrograph images were used to examine the fracture surfaces. Two-way ANOVA and the Tukey-Kramer post hoc test were used to analyze the data (α=.05).

Results. The mean flexural strength for SR 2 mm (H)×3 mm (W) round 0.6 mm radius of curvature (583.6 ±49.7 MPa) was significantly higher than that of SS, which was 2 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature (502.8 ±23.3 MPa). Similarly, the mean flexural strength for CR was 3 mm (H)×3 mm (W) round 0.6 mm radius of curvature (682.9 ±36.8), which was significantly higher than that of CC, 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature (486.7 ±35.6 MPa).

Conclusions. The flexural strength of the yttria-stabilized tetragonal zirconia polycrystalline-based ceramics was affected by the connector dimension and design. The 2-round connector design was more able to withstand occlusal forces than the sharp design. The 3-connector design with a minimum cross section of 2×3 mm is recommended for anterior fixed dental prostheses, provided it has a round curvature. (J Prosthet Dent 2016;115:224-229)

Dental patients increasingly demand esthetic restorations; however, the inherent weakness of ceramic restorations is considered a significant limitation.[1] The introduction of yttria-stabilized tetragonal zirconia polycrystalline (Y-TZP)-based restoration is considered a breakthrough in the dental field because of the excellent tissue tolerance, esthetics, and transformation toughening property, which give a high flexure strength (up to 1200 MPa) and fracture toughness (6-9 MPa·m$^{1/2}$) compared with other ceramic systems.[2-5]

However, in spite of the high mechanical properties of Y-TZP-based restorations, clinical failures still occur, usually at the connector area of fixed dental prostheses (FDPs).[6] Clinical recommendations for Y-TZP-based FDP connectors varied from 2 to 4 mm in height and width.[7-12] In

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Clinical Implications

The fabrication of partial fixed dental prostheses with a minimum connector size requires the use of a round connector design to improve flexural strength.

a survey of the literature, Larsson et al[7] compared the fracture strength of 4 unit Y-TZP-based FDP with different connector size (2.0, 2.5, 3.0, 3.5, and 4.0 mm) and concluded that the minimum recommended diameter is 4 mm. Schmitter et al[11] stated that the use of Y-TZP-based FDPs with a 9 mm^2 connector was clinically recommended. Oh and Anusavice[12] concluded that the fracture strength of ceramic restorations is influenced by the radius of the gingival curvature of the pontic, where the failure load increased with the increase of the radius at the gingival embrasure. Similarly, Plengsombut et al[13] stated that fracture usually occurs in the gingival surface of the connector and propagates toward the pontic. Kamposiora et al,[14] in their 2-dimensional finite element stress analysis method, also showed that increasing the connector height dramatically reduces the stress levels within the connectors. Y-TZP-based restorations can be fabricated by copy milling or computer-aided design and computer-aided manufacturing (CAD/CAM) technology. Wimmer et al[15] found that FDPs fabricated from CAD/CAM resin withstand higher load values than those conventionally fabricated.

In the fabrication of FDPs, properly designed connectors allow the separation of the units, thus permitting the development of natural-appearing labial embrasures. The smaller connector size can allow technicians to achieve this goal. Although this affects overall strength,[7] determining the minimum dimensions for the connector that can be used clinically is useful. A metal ceramic restoration allows the use of a small connector size, so the esthetic goal can be achieved from this perspective.[16] However, with ceramic restorations, different studies have evaluated connector size and shape and addressed the minimum size needed to fabricate a clinically acceptable ceramic restoration. Results have proven controversial.[17-24]

The purpose of this in vitro study was to evaluate the effect of different connector designs on the flexural strength of simulated 3-unit zirconia FDPs using CAD/CAM technology. The null hypothesis was that the flexural strength of Y-TZP-based FDPs would not be affected by altering the connector size or geometry.

MATERIAL AND METHODS

A power analysis of the flexural strength data was designed to determine an adequate sample size by Epi Info v6 (US Centers for Disease Control and Prevention) to calculate the sample size, guided by the power of the test, 80%, and the accepted margin of error, 5%. The alpha level was .05, β .20 for the power of 80%. The effect size calculation was based on the study of Plengsombut et al,[13] which found a statistically significant difference equal to 386.2 N between Zir-CAD (IPS e.max Zir CAD) and Press (IPS e.max Press). The predicted number was 5 specimens in each group.

To simulate a 3-unit FDP, 20 rectangular bar-shaped specimens, of the same design used by Plengsombut et al,[13] with dimensions of 4 ±0.05 mm (H)×4 ±0.05 mm (W)×30 ±0.5 mm (L), were made. Each bar specimen had 2 constricted parts on both sides representing the connector and defining a central pontic of 10 ±0.10 mm in length (Fig. 1). An additional 5 specimens fabricated with no constriction served as the control group.

The specimens were divided in 4 groups according to the connector diameter and design, as follows: small round connector (SR): 2 mm (H)×3 mm (W) round 0.6 mm radius of curvature; small sharp connector (SS): 2 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature; conventional round connector (CR): 3 mm (H)×3 mm (W) round 0.6 mm radius of curvature; and conventional sharp connector (CS): 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature.

To fabricate the specimens, partially sintered zirconia blocks (41×84 mm) (Procera zirconia; Nobel Biocare) were attached to a 5-axis water-cooled CAD/CAM milling machine (CAD/CAM machine 3011; Nobel Biocare). The system software created a 3-dimensional model of a bar-shaped specimen (4×4×30 mm), and the machine was activated to mill the required design. Each milling block produced 5 specimens. After initial milling, the 4 connector designs previously described were machined at both ends of each bar at a 10 mm distance. The design of each connector was added to the software module and the milling tool number 12 and 13 (Diamond disc cutter 110; Nobel Biocare), then changed to meet the design of the required connectors. After milling, the specimens were detached from the mounting frame and polished on a metallographic rotating device (M3000; Buehler Ltd) with ascending grit silicon carbide paper (600, 800, 1000) to remove milling trace lines. The specimens were then sintered according to the manufacturer's instructions (1500°C for 6 hours) in a furnace (Procera Lava; Nobel Biocare). The dimensions of the produced specimens were measured and verified with an electronic caliper with an accuracy of 0.01 mm.

All sintered specimens were then subjected to 3-point flexure strength test using a universal testing machine (model LRX-plus; Lloyd Instruments Ltd). The specimens were vertically loaded by means of a steel ball with a diameter of 3 mm at a crosshead speed of 0.5 mm per minute placed in the center of the pontic. The span of the

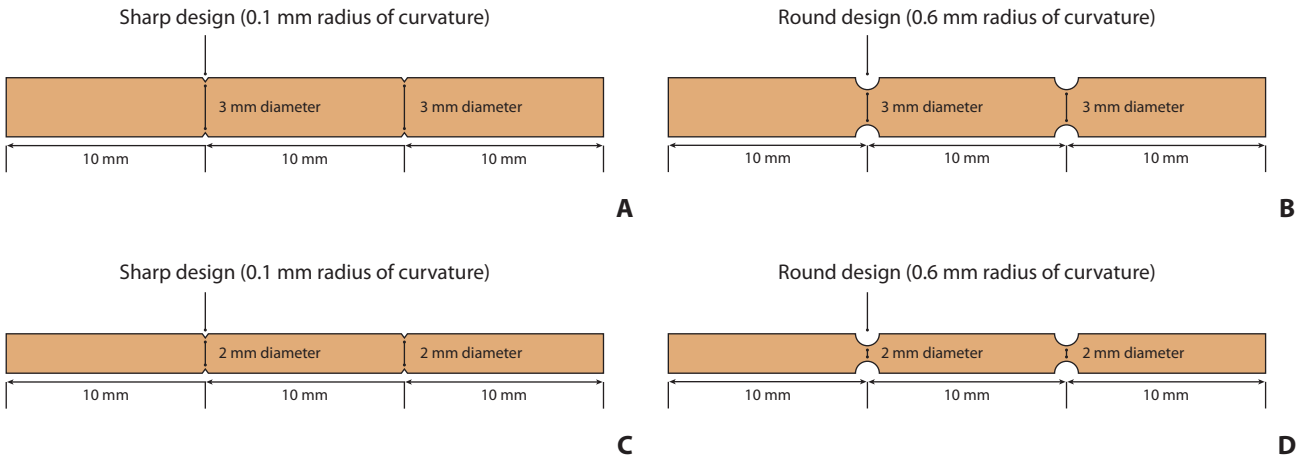

Figure 1. Diagrams showing shape and dimension of specimens.

Table 1. Mean ±SD values of flexural strength values (N)

Connector Diameter	Connector Design	Mean	SD	Rank	*P*
3 mm	Round	682.9	36.8	B	.04*
	Sharp	502.8	23.3	D	
2 mm	Round	583.6	49.7	C	
	Sharp	486.7	35.6	E	
Control		891.8	57.9	A	

SD, standard deviation.
*Significant at $P \leq .05$. Different letters were statistically significant.

Table 2. Two-way ANOVA results for effect of different variables on flexural strength

Source of Variation	SS	*df*	MS	F	Power	*P*
Connector diameter	16 643.9	1	16 643.9	9.3	1.000	<.001*
Connector design	95 962.4	1	95 962.4	53.4	1.000	<.001*
Connector diameter×connector design	8655.5	1	8655.5	4.8	0.990	<.001*
Error	644.2	20	32.2			

ANOVA, analysis of variance; SS, sum of squares; *df*, degree of freedom (n-1); MS, mean squares.
*Significant at $P \leq .05$.

supporting rollers was 25 mm, with the connectors equally positioned around the center of the loading point. The peak load F (N) recorded at the fracture point and the flexure strength ó (MPa) of the specimens were calculated using the following formula[25]:

$$ó=(F\times\text{span length})/(\text{radius}^3\times\pi),$$

where ó is the flexural strength, F is the maximum load at fracture (N), L is the length of the specimen between the 2 supports (mm), and R is the radius of the connector (π=3.14).

Two specimens from each group were selected for examination of the fracture pontic-connector interface under a scanning electron microscope (SEM; JSM-636OLV; JOEL Ltd). These specimens were steam cleaned for 10 seconds before SEM imaging. The surfaces of the zirconia bars were sputter-coated with gold-palladium alloy (Balzers-SCD Sputter Coate; Fürstentum) under high vacuum. Photomicrographs at ×200 magnification were made from different regions to evaluate the fractured surfaces. Data were presented as mean and standard deviation (SD) values. A regression model using 2-way analysis of variance (ANOVA) was used to test the significance of the effect of the connector diameter, connector design, and their interactions on the flexural bond strength. The Tukey-Kramer post hoc test was used for pairwise comparison between the mean values when the ANOVA test was significant (α=.05). Statistical analysis was performed with software (SPSS v20 for Windows; IBM Corp) (Tables 1, 2).

RESULTS

The results of this study showed that the highest mean flexural strength was recorded in the control group, 891.8 ±57.9MPa, while the lowest mean flexural strength was recorded in SS, 2 mm (H)×3 mm (W) sharp 0.1 mm radius of the curvature group, 502.8 ±23.3MPa. SR 2 mm (H) ×3 mm (W) round 0.6 mm radius of curvature was significantly larger than CS 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature (583.6 ±49.7MPa, 502.8 ±23.3MPa) (Table 1). The 2-way ANOVA results revealed that the connector diameter, design, and the interaction between the 2 variables had a statistically significant effect (P<.05) on the mean flexural strength (Table 2).

The results also showed that the SR 2 mm (H)×3 mm (W) round 0.6 mm radius of curvature was significantly higher than that of the SS 2 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature. Similarly, the CR 3 mm (H)×3 mm (W) round 0.6 mm radius of curvature was significantly higher than that of the CS 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature (Table 2). SEM observation of the fracture pattern from a cross-sectional view

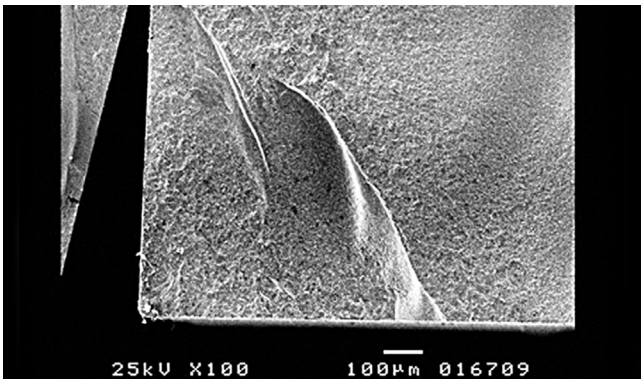

Figure 2. SEM demonstrating crack origin at connector site (original magnification, ×100).

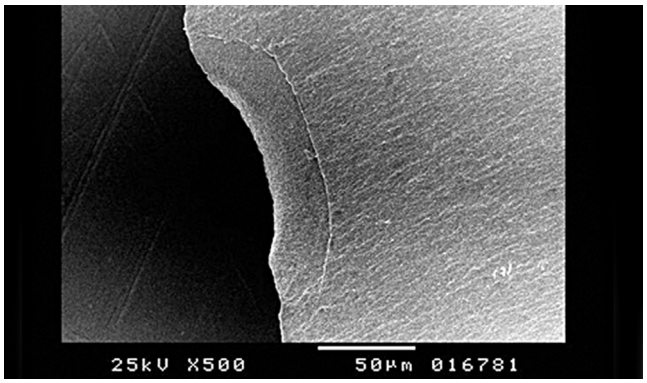

Figure 3. SEM demonstrating critical crack site at connector region. Secondary crack line and boundary are visible (white arrow) (original magnification, ×500).

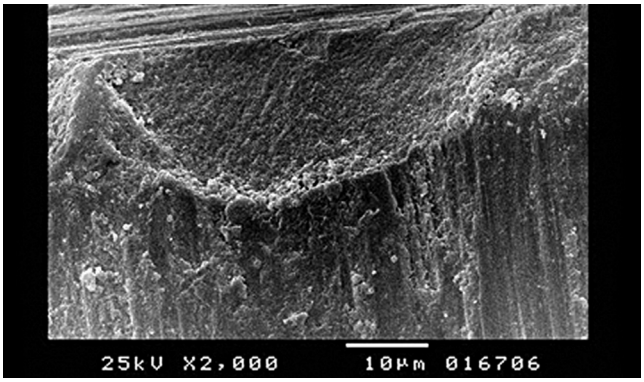

Figure 4. SEM demonstrating high magnification of fracture site at narrow connector design and machining trace lines on surface (white arrow) (original magnification, ×2000).

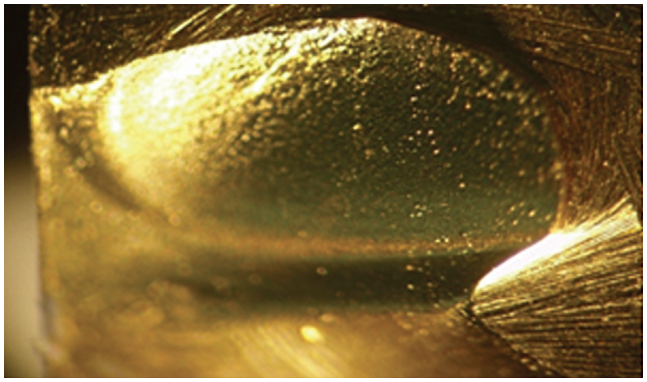

Figure 5. Photomicrograph demonstrating origin of critical crack at tensile surface of 3 mm round connector. Origin of fracture lies in smooth mirror area surrounded by black arrows (original magnification, ×200).

revealed that the fracture initiated from the gingival side of the connector toward the pontic (Figs. 2-4). In addition, photomicrographs showed oblique fracture patterns in all test specimens (Figs. 5, 6).

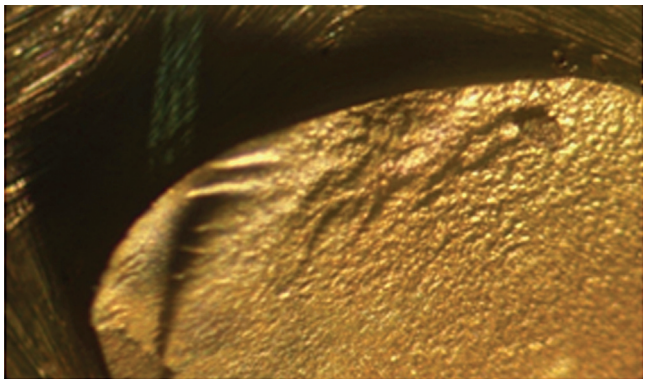

Figure 6. Photomicrograph demonstrating origin of critical crack at tensile surface of 2 mm round connector. Origin of fracture lies in smooth mirror area surrounded by black arrows (original magnification, ×200).

DISCUSSION

The null hypothesis of this study was rejected because the results showed that altering the connector size and geometry affects the flexural strength of Y-TZP-based FDP.

The specimens used had the same design as previously used by Plengsombut et al.[13] Although the use of an anatomic FDP shape would be more clinically relevant, a standardized dimension was needed to calculate the flexural strength. For the purpose of this standardization, the use of a bar-shaped design was selected, incorporating 2 connector designs (round and sharp radius).

The Y-TZP-based framework was used without adding a porcelain veneer because the veneer layer could not be standardized. This was another reason to exclude the effect of veneering technique on the flexural strength whether pressed, layered, or CAD/CAM.

Using a Y-TZP-based FDP with 2 mm (H)×3 mm (W) connector size and a round or sharp radius of

curvature is considered a new approach because of the importance of examining such a small connector size when the manufacturer recommends the minimum to be 3×3 mm.

In this study, a special equation was used to calculate the flexural strength values of the tested groups according to ASTM C1684-13e126, as the specimen design used in this study was straight with 2 constrictions, which represent the connectors.

The results of this study showed that the FDP with the largest dimensions produced higher flexural strength values than those obtained with smaller ones. These results agree with other studies, which found that fractures of zirconia in ceramic FDPs increased with smaller-sized connectors. This was in agreement with the findings of Studart et al,[10] who compared 3-, 4-, and 5-unit zirconia FDPs, suggesting that the minimal connecting surface should not be less than 2.7 mm^2, 4.0 mm^2, and 4.9 mm^2, respectively. Moreover, the obtained results are in agreement with a previous finite element study, which found that when the connector height increased from 3 to 4 mm, stress would decrease by 50%, thus increasing the flexural strength. Other studies reported that the connector's cross-sectional diameter and shape are crucial factors in the long-term success of zirconia FDPs and should be determined according to material properties, anatomic limitations, and esthetic expectations.[22,23]

When occlusal force is applied directly to the long axis of a ceramic FDP connector, compressive stresses develop on the occlusal aspect, while tensile stress develop on the gingival aspect; such stresses contribute to the propagation of microcracks located at the gingival surface, leading to fracture. Increasing the dimensions of the connector may decrease this effect.

This study also showed that an FDP with a connector of 2 mm (H)×3 mm (W) round 0.6 mm radius of curvature was significantly larger than that a connector of 3 mm (H)×3 mm (W) sharp 0.1 mm radius of curvature. These findings have important clinical effects, as they can allow dental laboratories to choose such a combination when fabricating FDPs in areas with limited space, thus increasing esthetics and function. This was in agreement with a study by Tsumita et al,[20] which concluded that the shape of the framework of an FDP will affect the stress distribution on the definitive restoration and may lead to failure.

In the present study regarding connector geometry, the fracture pattern of all specimens with the sharp connector design was less angulated toward the pontic compared to that of the round connector design. This finding may be explained by the different stress levels within rounded and sharp connector designs. The obtained results were in agreement with other preliminary studies, which found that smoother and less angled connectors showed lower stress levels,[2,5,21] possibly explaining the different fracture directions in each connector design in this study.

The mean adult occlusal force in the posterior area was reported to be 400 to 800 N, 300 N in the premolar area, and 200 N in the anterior area.[26] As the mean flexural strength of the smallest connector in this study with a round configuration was 583.6 ±49.7 MPa, it can be used in the anterior area.

The tested specimens showed brittle fracture pattern without any signs of plastic deformation, and the fracture surface was smooth and glossy. Fracture started in an area where the concentration stress was high. The propagation action of the crack increased as the stress increased until it reached a level at which the crack continued to propagate without any additional stress (catastrophic failure).[12]

In vitro studies cannot yet be correlated with the clinical situation, but such studies allow tested groups to be compared. This study used bar-shaped specimens with no ceramic veneer. Although the addition of such a veneer might influence the result of the study, it was excluded to standardize the variable of the framework design because many factors can be intercorrelated with ceramic veneer, including the ceramic used, fabrication techniques, and shear bond strength. Further studies are needed to evaluate the effect of ceramic veneer, together with follow-up clinical studies.

CONCLUSIONS

Within the limitations of this in vitro study, the following conclusions were drawn:

1. The flexural strength of the Y-TZP-based ceramics is affected by the connector dimension and design.
2. The round connector design may withstand occlusal forces better than the sharp design.
3. Connectors with a minimum cross section of 2×3 mm^2 are recommended for anterior FDPs, provided they have a round curvature.

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