Effect of Connector Design on Fracture Resistance in All-ceramic Fixed Partial Dentures for Mandibular Incisor Region

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Abstract

Yttrium tetragonal zirconia polycrystal frameworks were prepared for all-ceramic fixed partial dentures (FPDs) for the mandibular incisor region. The effects of the cross-sectional area and morphology of the connector on its strength were evaluated by fracture tests. Nine types of zirconia framework for a 3-unit FPD for a defect of 1 mandibular central incisor were prepared, each differing in cross-sectional area and morphology. Fracture tests were performed by loading until fracture using a universal testing machine. Fracture load was determined and fracture site examined. Significant differences were observed in fracture load according to the morphology and cross-sectional area of the connector (p < 0.05, p < 0.01, p < 0.001). Fracture load differed significantly among all groups according to cross-sectional area, and was also greater when the shape of the connector formed an isosceles triangle widest at the base and the connector had the same height and width. These values still far exceeded 311 N, however, which is the average occlusal force in the incisor region. The results of this study suggest that connector design affects fracture load.

Key words: Fixed partial dentures—Zirconia—All-ceramic FPDs—Fracture load—Incisor region

Introduction

Recently, ceramic materials have been used to markedly strengthen high-density sintered cores, and such cores have seen successful clinical application. Among the various types of ceramic available, 5% yttria-stabilized tetragonal zirconia polycrystal (Y-TZP), in particular, offers a number of favorable properties, including greater bending strength and fracture load than conventional ceramics for cores containing aluminum oxide. These properties now mean that they can be applied to fixed partial dentures (FPDs) in the molar region.

Being the narrowest part of an FPD, stress...
tends to be concentrated on the connector, placing considerable mechanical requirements on this area\(^{1,6,8,11}\). A cross-sectional area of 9.0 mm\(^2\) or greater has been reported to be desirable for the connector of a 3-unit Y-TZP framework in order to obtain sufficient strength\(^{16}\), and as such is recommended by many manufacturers of dental materials.

The mandibular incisors are frequently exposed, so the esthetic demands for prosthetics in this region are high, and treatment using all-ceramic crowns is very effective in this respect. However, in designing all-ceramic FPDs, it is often difficult to secure a cross-sectional area of 9.0 mm\(^2\) in the connector because of the low crown height. On the other hand, in the mandibular incisor region, the crown width is narrow, and less occlusal force is exerted there than on other parts. Therefore, FPDs applied to the incisor region do not need to be as fracture-resistant as those applied to the molar region. These observations suggest that determining the appropriate cross-sectional area of a connector is highly important in designing the framework of an FPD for the mandibular incisor region. Moreover, in designing FPDs with high esthetic qualities, it is important to make sure that the cross-sectional area of the connector is sufficient, as the height of the connector is known to exert a major effect on its fracture load if the cross-sectional area is the same value\(^{5,13}\).

In this study, therefore, we prepared Y-TZP frameworks for 3-unit FPDs for the mandibular incisor region with connectors of differing cross-sectional areas and morphologies and compared their fracture loads. We also evaluated the relationship between cross-sectional area and fracture load.

**Materials and Methods**

Assuming a 3-unit FPD for the lower anterior region with one missing central incisor, a master model of the abutments was designed as the standard preparation for the lower central and lateral incisors to simulate an actual framework (Fig. 1).

The abutments had a diameter of 5.0 mm and a height of 6.5 mm. The axial surface had a taper degree of 7°. The periphery was designed as a deep chamfer with a curvature radius of 1.0 mm. The distance between abutments was 11.0 mm. The abutment holder was made of stainless steel (SUS303). An impression was taken from the master model with vinyl silicone impression material (Fusion II wash type, GC, Tokyo, Japan) and a working cast prepared with high-strength dental stone (New Fujirock, GC). From this working cast, FPDs frameworks were prepared with the GN-1 system (GC). The frameworks were made of 5% Y-TZP (Aadva Zirconia, GC).

Connectors with the following 3 different cross-sectional areas were designed: 9.0 mm\(^2\), 7.0 mm\(^2\), and 5.0 mm\(^2\). Three different forms (Type I, II, and III) of cross-section were prepared for each of these 3 cross-sectional areas, as shown in Fig. 2. In Type I, the cross-section assumed the form of an isosceles triangle (height/width ratio of 2:1) to be fitted to the outside contour of the coping for the lower central incisor. The upper end of the triangle was matched to the upper end of the coping to position the connector.
The center of this triangle was defined as the point of intersection between two lines, one extended at 90° from halfway along the length of the triangle and the other extended at 90° halfway along the base of the triangle. This point of intersection was taken as the criterion point for application of the other two types. Type II had the same height and width as Type I, but its maximum width was designed to be located at the center. In this way, Type II was designed as an isosceles triangle with the base arranged on the labial side. Type III was designed, on the basis of Type I, as an isosceles triangle, with the height being three quarters of that of Type I and the area being equal to that of Type I. In application of Types II and III, the center point was matched against the criterion center point provided by Type I.

Each form of connector was designed to have no acute angle point so that operation was possible in the GN-I system.

At cross-sectional areas of 7.0 mm² and 5.0 mm², Type I, II, and III were designed to assume a similar figure around the criterion point. Five frameworks of each design were prepared (45 frames in total).

In the test model for the fracture test, the abutments were set in the abutment holders in the same way as in the master model. Silicone material with a thickness of 1.0 mm was placed between the abutment and the abutment holder to act as a pseudo-periodontium (Fusion II wash type). After the frameworks were fabricated, silicone impression material was used to check how well the copings fitted at the margin level (Fit Checker, GC). Zirconia frameworks cut with the GN-I system were sintered and checked for fit. They were then cemented to the mold with glass ionomer cement (GC Fuji I, GC). Each cemented framework was stored in distilled water at 37°C for 24 hrs. The static load bearing test was carried out with a universal testing machine (AG-I 20KN, SHIMADZU, Kyoto, Japan). Load was applied vertically to the middle of the pontic at a cross-head speed of 1.0 mm/min and fracture load determined. Teflon (polytetrafluoroethylene) disks with a thickness of 2.0 mm were interposed between the loading stamp and framework (Fig. 1). Each teflon disk was replaced with a new one every test.

A microscope system (MORITEX Co., Tokyo, Japan) was used to observe the fracture site in each connector, and fractured surfaces coated with Au-Pd were observed with a scanning electron microscope (SEM JSM-6340F, JEOL, Tokyo, Japan).

The obtained data on fracture load were subjected to a two-way ANOVA involving two factors: cross-sectional area and morphology. If no interaction was revealed by this method, a multiple comparison was performed (SAS Ver.9.1, SAS, North Carolina, USA). The Tukey method was employed for statistical evaluation of the results.

Results

Figure 3 shows the fracture loads of the FPDs. The mean fracture load was 2,416, 1,772, and 1,120 N in groups with cross-sectional areas of 9.0, 7.0, and 5.0 mm², respectively.

Examination of the fractured areas revealed that all crack lines ran from the gingival side to the incisal edge of the pontic in all samples.
The two-way ANOVA showed significant differences in fracture load according to the morphology and cross-sectional area \((p < 0.0001)\). No interaction was noted between them. The Tukey multiple comparison revealed that fracture load differed significantly according to cross-sectional area among all groups \((p < 0.001)\), with a marked differences \((p < 0.001)\) between Types I and II and between Types I and III and significant difference \((p < 0.05)\) between Types II and III (Table 1).

**Discussion**

In this study, fracture tests were performed on 3-unit FPDs for a defect of one mandibular central incisor. The FPDs were designed on the basis of the average morphology and size of abutment die and inter-abutment distance by simulating intraoral conditions. Lüthy *et al.* reported that the potential of the material appears more clearly in fixed than movable abutments and that there is no effect of cementation on the average load-bearing of FPDs\(^6\). Therefore, in the present study, fracture tests were performed under near-clinical conditions by placing a simulated periodontal ligament and using glass ionomer cement, a luting agent in wide clinical use. Also, the abutment teeth and pontic were arrayed linearly for morphological simplification of the samples. This array is often observed in

![Fig. 3 Load required for fracture by type of framework and cross-sectional area](image1)

*Fig. 3 Load required for fracture by type of framework and cross-sectional area*  
*p < 0.05*

![Fig. 4 Frequency of fracture origin](image2)

*Fig. 4 Frequency of fracture origin*

**Table 1** Mean fracture load of FPD frameworks

<table>
<thead>
<tr>
<th>Cross section area (mm(^2))</th>
<th>n</th>
<th>Average (N)</th>
<th>Tukey test</th>
</tr>
</thead>
<tbody>
<tr>
<td>9.0</td>
<td>15</td>
<td>2,765 ± 444</td>
<td>***</td>
</tr>
<tr>
<td>7.0</td>
<td>15</td>
<td>2,058 ± 360</td>
<td>***</td>
</tr>
<tr>
<td>5.0</td>
<td>15</td>
<td>1,230 ± 161</td>
<td></td>
</tr>
</tbody>
</table>

| Form | Type I     | 15 | 2,276 ± 840 | ***        |
|      | Type II    | 15 | 1,760 ± 587 | ***        |
|      | Type III   | 15 | 2,007 ± 656 | ***        |

*p < 0.05, **p < 0.01, ***p < 0.001*
actual clinical practice.

Usually, an FPD connector with a large cross-sectional area is advantageous in terms of strength. Clinically, however, a connector with a small cross-sectional area is esthetically more desirable and easier to clean. In the mandibular incisor region, in particular, the crown morphology is small, and it is occasionally difficult to secure a cross-sectional area of 9.0 mm$^2$, which is common in connectors. Therefore, we also performed fracture tests using frameworks with connectors of 7.0 and 5.0 mm$^2$ in cross-sectional area.

The fracture load differed significantly ($p<0.001$) among all groups according to cross-sectional area. The maximum occlusal force in the incisor region in the average Japanese has been reported to be 189.1–310.7 N$^3$. While the fracture load showed a significant reduction with a decrease in the cross-sectional area of the connector, it was well over 310 N, even when the cross-sectional area was only 5.0 mm$^2$. Therefore, zirconia frames are considered unlikely to fracture due to occlusal force. Also, FPDs with a connector cross-sectional area of less than 9.0 mm$^2$ are unlikely to cause clinical problems, as the inter-abutment distance is short in the mandibular incisor region.

According to the results of the fracture tests, the cross-sectional area and fracture load were nearly in proportion, and a fracture load sufficient for clinical application may be obtained with a cross-sectional area of 3.0 mm$^2$. However, frameworks with a connector cross-sectional area of 3.0 mm$^2$ resulted in many cracks occurring during cutting and sintering, ending in failure. Future improvements in all-ceramic materials and FPD preparation may allow a connector cross-sectional area of only 3.0 mm$^2$.

A triangular connector was considered appropriate for the mandibular incisor region, as this is a shape that efficiently provides sufficient height and width. First, in consideration of the morphology of the crowns, abutment teeth, and pontic, the connector was designed as an isosceles triangle that fitted the coping of the mandibular central incisor with a height/width ratio of 2:1 and its base on the gingival side (Type I). Also, assuming a ridge lap type and lateral pontics, two more types of triangle were evaluated: an isosceles triangle with the same cross-sectional area as Type I, but with the maximum width located in the center and the base on the lingual side (Type II); and, assuming that sufficient height could not be secured, a triangle with the same cross-sectional area, but the height reduced to 3/4, with the base on the gingival side (Type III).

With a 3-point bending test, the width and square of the height of a sample affect its fracture load$^5$. Also, fracture load tends to increase with increase in the thickness of the base, where tensile stress is concentrated. Based on these assumptions, we hypothesized that fracture load would be in the order of Type I > Type II > Type III. However, this was not supported by the results.

Fracture load was significantly higher in Type I than in Type II ($p<0.001$). Here, the fracture originated in the lower part of the connector, suggesting that the large width in this type prevented concentration of tensile stress at this site. In Type II, the lower part of the connector had only one corner, resulting in marked concentration of stress in this area. Type III, in which the connector is widest at the base, as in Type I, showed a significantly higher fracture load than Type II ($p<0.05$). This suggests that as fracture load was only proportionate to height with this type of morphology, this would be the best option in preventing concentration of stress.

On the other hand, Type I showed a significantly higher fracture load than Type III ($p<0.001$). Types I and III had the same cross-sectional area but differed in height and width. If the shape was similar, the height was shown to exert a greater effect than the width. These results suggest that FPDs should be designed with the maximum thickness at the base of the connector, where tensile stress is concentrated, while securing sufficient height.

As can be seen in the SEM image in Fig. 5, wrinkles extending radially from the point
indicated by the arrow show that the fracture originated in the lower part of the connector in all types. This was in agreement with clinical reports indicating that fracture originated on the gingival side (lower part) of the FPD connector.

Clinically, failure in FPDs with zirconia frames frequently results from fracture of the veneer porcelain and infrequently from fracture of the zirconia framework. The framework design is a major factor affecting fracture in veneer porcelain. Thus, a strong frame design is believed to reduce tensile stress on veneer porcelain. Further experiments using frameworks with sintered porcelain veneers are considered necessary.

As observed above, the cross-sectional area of the connector in 5% Y-TZP frameworks influenced fracture load. However, fracture load markedly exceeded 311 N, which is the average occlusal force in the incisor region. In connectors with the same cross-sectional area, the height and position of the maximum width affected fracture load.

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**Fig. 5** Fractured cross-section in Y-TZP FPDs (SEM VIEW)

a: Type I×50, b: Type I×100, c: Type II×50, d: Type II×100, e: Type III×50, f: Type III×100, †: origin of fracture