Effect of Connector Design on Fracture Resistance in Zirconia-based Fixed Partial Dentures for Upper Anterior Region

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Abstract

The purpose of this study was to investigate the influence of the cross-sectional form and area of the connector on fracture resistance in three-unit zirconia fixed partial denture (FPD) frameworks for the upper anterior region. Sixty FPD framework specimens were fabricated using the CAD/CAM system. The cross-sectional form (Type I, II, or III) and area (9.0, 7.0, 5.0, or 3.0 mm²) of the connectors differed. The specimens were fixed to a jig capable of applying a load axially to the abutment teeth at an angle of 135 degrees. Each specimen was subjected to fracture load measurements using a universal testing machine and cross-sectional microscopic examination. Fracture load fell significantly with a decrease in cross-sectional area (p<0.01). In terms of cross-sectional form, an isosceles triangle with a gingival base yielded the highest fracture load. These results suggest that the connector of a three-unit zirconia-based FPD framework for the upper anterior region should be triangular, have a gingival base, sufficient height in the loading direction, and a cross-sectional area of >5.0 mm².

Key words: Connector design—Zirconia—Fracture resistance—Fixed partial denture—Anterior teeth

Introduction

Recently, ceramic materials have been used to markedly strengthen high-density sintered cores. Moreover, the application of such cores in a clinical setting has been successful. One such ceramic, 5 mass% yttria-stabilized tetragonal zirconia polycrystal (Y-TZP), offers a number of favorable properties, such as greater bending strength and a higher fracture load than conventional ceramics with cores containing aluminum oxide. These properties make Y-TZP a popular choice in the fabrication of fixed partial dentures (FPDs) for the molar region. The superior bending strength and fracture resistance of Y-TZP, in particular, have been responsible for its clinical success, with its application to the fabrication of single crowns and FPDs, for example. The form of the
connector for an FPD can easily be altered using the CAD/CAM system. Connector form has been investigated employing destruction tests \(^1,15,16,21,32\) and finite-element analysis \(^17,36\). In none of these studies, however, did the loading conditions assume the anterior tooth region.

The upper incisors are frequently visible, so esthetic considerations here are of great importance, which is another reason for the popularity of all-ceramic crowns. This can cause some difficulty in designing an FPD for this region, as if the connector is too large, it might result in esthetic and periodontal problems. The larger the connector, however, the greater its strength \(^13,16\). Many manufacturers \(^16,21,32\) recommend a cross-sectional area of 7.0 to 9.0 mm\(^2\) for connectors in zirconia FPD frameworks for the upper anterior region designed using the CAD/CAM system. With a narrow FPD connector, mechanical stress is distributed vertically in the lower anterior region. In the upper anterior region, however, the stress will be distributed obliquely. The purpose of the present study was to investigate the influence of the cross-sectional form and area of the connector on fracture resistance in three-unit zirconia FPD frameworks for the upper anterior region.

**Materials and Methods**

A master model of two abutment teeth and a missing upper central incisor was fabricated with stainless steel (SUS303). The average all-ceramic crown size for central and lateral incisors in Japanese adults was used to determine the size of the abutment model (Fig. 1). The abutment of the central and lateral incisor was 6.0 and 5.2 mm, while height was 6.8 and 5.3 mm, respectively. The abutments had deep chamfer margins with a curvature radius of 1.0; the axial plane taper was 5 degrees, while the inter-abutment distance between the centers of the abutments was 16.1 mm.

An impression of the master model was taken with vinyl silicone impression material (Fusion II, GC, Tokyo, Japan) and a working cast fabricated with high-strength dental stone (New Fujirock, GC). The working cast was scanned using the GN-1 system (GN-1 system, GC) and the framework designed. As the material for the framework, Y-TZP (Aadva Zirconia, GC) was selected. Using CAD/CAM, the thickness of the coping was set at 0.5 mm and the cement space at 60 \(\mu\)m prior to fabrication of the framework. As shown in Fig. 2, three different forms of cross-section were used. In Type I, the cross-sectional form of the connector constituted an isosceles triangle with a gingival base and base width-to-long axis height ratio of 1:2; here, the long axis of the triangle was coincident with
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the axial direction of the coping. In Type II, the cross-sectional form of the connector constituted an isosceles triangle with a labial base and height of the long axis ratio of 2:1. In Type III, the cross-sectional form of the connector constituted an isosceles triangle with a palatal base and height of the long axis ratio of 2:1. All three types of connector had the same height and width. In order to ensure that the shape of the connector corresponded to that of the upper lateral incisor, the center of the triangle was set at the intersection between the point halfway between the top edge of the coping and the bottom edge of the pontic and the midpoint along the maximal bucco-palatal axis of the coping. Each cross-sectional connector type was designed to have no sharp edges to allow milling in GM-1000 (GM-1000, GC).

Five specimens each with a cross-sectional area of 9.0, 7.0, 5.0, or 3.0 mm² were produced for each type of connector, giving a total of 60 specimens.

Figure 3 shows the test model. Abutment rods with a hemispherical apex were fixed in an abutment holder with silicone acting as para-periodontal ligament. The portion of the abutment rod contained within the holder was 12 mm in length. The silicone para-periodontal ligament was 0.4 mm in thickness (Fusion II, GC). The fit of the framework was checked using silicone impression material (Fit Checker, GC). Each framework was fixed to the abutment with resin cement (ResiCem®, SHOFU, Kyoto, Japan) and stored in distilled water for 24 hr at 37°C.

The test model was fixed to a jig capable of applying load axially to the abutment teeth at an angle of 135 degrees (Fig. 4); fracture load was measured using a universal testing machine (AG-I 20KN, SHIMADZU, Kyoto, Japan) at a cross-head speed of 0.5 mm/min. The loading point on the pontic was located in the center of the mesiodistal axis at approximately 2.8 mm lingually from the top of the framework on the assumption that the lower anterior teeth would occlude toward this point. Load was applied with a steel ball of 5 mm in diameter. A piece of articulating paper (Articulating Paper, GC) and two layers of tin foil were inserted between the steel ball and specimen to act as a cushion; the cushioning materials were changed for each experiment.

The fracture mode of each specimen was confirmed after fracture load had been measured. The fractured surface was coated with Au-Pd and observed with a scanning electron microscope (SEM, JSM-6340 F, JEOL, Tokyo, Japan).

The measured fracture load values were
subjected to a two-way ANOVA, with the cross-sectional area and morphology of the connector as elements. The presence or absence of interaction between these elements was then determined based on the results. Multiple comparison (SAS, North Carolina, USA) was to be performed when no interaction was detected. The Tukey method was employed for evaluation.

### Results

Figure 5 shows mean fracture load with each type of framework. Fracture load ranged from 1,057 to 1,270 N on average with a 9.0 mm$^2$ cross-sectional connector area, 830 to 982 N with 7.0 mm$^2$, 558 to 618 N with 5.0 mm$^2$, and 324 to 373 N with 3.0 mm$^2$. Type I exhibited the highest fracture load values.

The two-way ANOVA revealed that cross-sectional area and type of connector both had a significant effect on fracture load (p<0.001), but no interaction between these parameters was observed (Fig. 6). No interaction was observed between the cross-sectional area and morphology of the connector. Therefore, a multiple comparison was performed using the Tukey method. The results revealed a significant difference in fracture load (p<0.001) among connectors depending on form (Table 1). Significant differences in

![Fig. 5 Mean fracture load and standard deviation of each FPD framework connector design](image)

![Fig. 6 Analysis of variance](image)

| Cross-sectional area (mm$^2$) | n  | Means ± SD (N) | Tukey analysis |
|-----------------------------|----|----------------|----------------|
| 9.0                        | 15 | 1,152 ± 164   | A              |
| 7.0                        | 15 | 904 ± 93      | B              |
| 5.0                        | 15 | 581 ± 50      | C              |
| 3.0                        | 15 | 345 ± 39      | D              |

Levels not connected by same letter are significantly different at p<0.01

| Cross-sectional form | n  | Means ± SD (N) | Tukey analysis |
|----------------------|----|----------------|----------------|
| Type I               | 20 | 811 ± 355     | a              |
| Type II              | 20 | 731 ± 331     | b              |
| Type III             | 20 | 694 ± 291     | b              |

Levels not connected by same letter are significantly different at p<0.05

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fracture load were observed between Type I and II, and Type I and III for different cross-sectional forms of connector (p<0.05). No significant difference was found between Type II and III.

Figure 7 shows examples of fractured specimens. Macroscopic examination of the fracture surface revealed mesio-distal linear fracture from the gingival surface of connector to the loading point on the pontic. Examination of Type I and II by SEM revealed twist hackle on the slope of the triangle opposite the loading point, and in the corner of the Type III triangle opposite the loading point (Fig. 8).

Discussion

The physical properties of zirconia frames have been investigated using various types of destruction tests\(^{1,15,16,21,32}\) and finite-element analysis\(^{17,36}\). Some studies using the former employed a thermal cycle to simulate aging\(^{1,15}\). The physical properties of zirconia change with frequency of firing\(^{33}\) or exposure to high temperatures after firing. In the present experiment, however, the influence of temperature was completely excluded in the destruction test. We believe that this may be of interest in comparing the present findings with those of studies employing high temperature treatment or thermal cycles.

Here, three-unit zirconia FPD frameworks for the upper anterior region, all assuming a missing central incisor, were subjected to fracture tests. The size of the abutments in the master model and inter-abutment distance were based on the average sizes seen in Japanese adults.

In constructing an experimental model, it is important to simulate the human intraoral environment as closely as possible. Vult von Steyn\(\text{e}r\)\ et al. and Fils\(\text{e}r\)\ et al. reported that fixed abutments exhibited higher fracture loads than natural abutment teeth in their test models\(^{35}\). Stress distribution measured in the presence of para-periodontal ligament was more similar to that in a human model than to that in the absence of para-periodontal ligament\(^{15}\). In the present study, therefore, the test model abutment rods were fixed in an abutment holder using silicone as para-periodontal ligament.

Clinically, failure of an all-ceramic FPD usually involves chipping of the porcelain-facing material\(^{20,22,28,29,34}\), not chipping of the framework. In addition to the firing strength required to fuse the porcelain and the frame, deformation of the frame is also a cause of this chipping of the porcelain-facing material. This suggests that both the design and strength of the framework may be involved in failure. Therefore, only fracture load of the framework was measured and evaluated in this experiment.

The abutment and pontic were aligned to avoid unknown variables, such as curvature of the dental arch, affecting the results. Many studies have reported that adhesive resin cement should be used with a zirconia coping\(^{11}\). In the present study, adhesive resin cement was utilized for the zirconia FPD frameworks in accordance with the manufacturer’s instructions.

Loading was applied axially to the abutment teeth at an angle of 135 degrees to simulate an actual anterior relationship\(^{23,30}\). Kagaya\ et al. reported that the mean overjet was 2.75 mm in Japanese adults\(^{8}\). Accordingly, the loading point was determined based on
an overjet of 3.0 or 1.0 mm in the thickness of facing porcelain in the present study. In addition, a piece of articulating paper and two layers of tin foil were inserted as a buffering material between the specimen and stainless steel loading ball to avoid fracture due to Hertzian cone crack, something that only rarely occurs in a clinical setting.

From the viewpoint of mechanical strength, the cross-sectional area of an FPD connector should be as wide as possible. Filser et al. reported that the cross-sectional area of a connector in a molar zirconia three-unit FPD should be >6.9 mm², as such connectors should be able to withstand a maximal occlusal force of >880 N, and hence, would be able to withstand forces generated by certain types of parafunctional behavior. The manufacturer of the GN-1 system recommends that the cross-sectional area of the connector used...
in the anterior region should be 9.0 mm$^2$. However, from an esthetic point of view\textsuperscript{13}, and taking into account periodontal considerations\textsuperscript{16,26}, the cross-sectional area should be as small as possible. Therefore, connectors with different cross-sectional areas (9.0, 7.0, 5.0, and 3.0 mm$^2$) were subjected to the fracture test in the present study. In earlier studies on the cross-sectional form of the connector, this group found that a cross-sectional area of 5.0 mm$^2$ was sufficient to withstand fracture\textsuperscript{2,16,21}. Indeed, until recently, it has been impossible to prevent cracking with a cross-sectional area of only 3.0 mm$^2$ when using the CAD/CAM system. The advent of the GN-1 system, however, has now made this possible.

Statistically significant differences ($p<0.001$) were observed in all groups depending on the cross-sectional area of the connector, with a reduction significantly decreasing fracture resistance. This result is consistent with those of Fischer et al.\textsuperscript{5}. Several reports have found that occlusal force measurements are dependent on the measurement methods and devices used\textsuperscript{22}. Takamizawa reported that the maximal occlusal force generated in the upper anterior area ranges from 148 to 290 N in the Japanese\textsuperscript{31}, which suggests that the fracture load of an FPD connector used in this region should be $>300$ N. In this study, while all the specimens in the 9.0, 7.0, and 5.0 mm$^2$ groups exhibited fracture loads of $>300$ N, several samples in the 3.0 mm$^2$ group displayed fracture loads of $<300$ N (283 to 396 N). This suggests that the cross-sectional area of the connector should be $>5.0$ mm$^2$.

The optimal cross-sectional form of an FPD connector for use in the upper anterior region is a triangle, as it allows sufficient height and width to be obtained (in terms of contact area) and the final prosthesis to be achieved. To investigate the influence of the position of the base of the triangle on fracture resistance, three variants of cross-sectional form were investigated: that of the Type I connector was an isosceles triangle with a gingival base and a base width-to-long axis height ratio of 1:2; that of Type II was an isosceles triangle with a labial base and a height of the long axis ratio of 2:1, assuming that ridge-lap or modified ridge-lap pontics were to be employed; that of Type III, which was similar to that of Type II, was an isosceles

| Connector Form of FPDs for Upper Incisor | Type I | Type II | Type III |
|----------------------------------------|--------|---------|----------|
| 9.0 mm$^2$                             | 3.55   | 3.31    | 3.31     |
| height of connector                    | 0.94   | 1.56    | 0.62     |
| contained extent of connector          | 0      | 0       | 0        |
| 7.0 mm$^2$                             | 3.12   | 2.92    | 2.92     |
| height of connector                    | 0      | 1.01    | 0        |
| contained extent of connector          | 0      | 0       | 0        |
| 5.0 mm$^2$                             | 2.61   | 2.46    | 2.46     |
| height of connector                    | 0      | 0       | 0        |
| contained extent of connector          | 0      | 0       | 0        |
| 3.0 mm$^2$                             | 2.03   | 1.91    | 1.91     |
| height of connector                    | 0      | 0       | 0        |
| contained extent of connector          | 0      | 0       | 0        |
triangle with a palatal base and height of long axis ratio of 2:1. All three types of triangle were found to correspond to the borders of the upper lateral incisor coping.

Table 2 shows the relationship between the height of the connector in the loading direction and the area of the connector against the loading axis. Statistically significant differences in fracture load were detected between the Type I and II connectors, and Type I and III connectors (p<0.05). These findings suggest that, with a 135-degree angle of loading, the height of the connector in Type I would be effective in resisting fracture. This result is consistent with the fact that the sample height is as influential as the square of the value in the three-point bending test. No statistically significant difference was observed between the Type II and III connectors (p>0.05), suggesting that the position of the triangle base does not affect fracture resistance. In Type II and III connectors with a 9.0 or 7.0 mm² cross-sectional area, the contained extent of loading axis involvement differed, despite the height of the connectors being the same in the loading direction. No statistically significant difference was observed in fracture load between the Type II and III connectors, which indicates that the area of the connector acting against the loading axis does not have a marked influence on fracture resistance, which contradicts the original assumption of this study. Furthermore, a broader dispersion of fracture load was observed in the Type II connectors with a cross-sectional area of 9.0 mm². This may have been due to the location of the furnace during the firing process or a defect in the milling instruments used. These results suggest that the cross-sectional connector of an FPD framework for use in the upper anterior region should have a triangular cross-sectional form, a gingival base, and as large a base width-to-long axis height ratio in the loading direction as possible.

Kelly et al. reported that fractures started on the gingival side of the connector where stress concentration occurred in a three-unit FPD in the molar area. In the present study, SEM revealed twist hackle on the slope of the triangle opposite the loading point in the Type I and II connectors, and in the corner opposite the loading point in the Type III connectors. We believe that these cracks might have originated in the twist hackle. The arrows in Fig. 8 show where the twist hackle and arrest lines were mainly observed. Twist hackle indicates the direction of crack propagation, which suggests that the fracture observed here started in the region marked with an arrow.

The fracture load of the FPD frameworks was measured. Further study of FPD frameworks with porcelain facing is required. There are many unknown variables that can affect fracture resistance. The present study aimed to exclude these factors and focus on the effects of the cross-sectional form of the connector. Many studies have reported that the curvature of the dental arch and curvature radius of the connector also influence fracture resistance, and that the strength of an FPD framework decreases in a dynamically loaded or wet environment. Therefore, further study is needed to elucidate the influence of connector form employing conditions as close as possible to that of the oral environment.

Conclusion

Within the limitations of the present study, the following conclusions were drawn: the cross-sectional area of the connector of an FPD framework for use in the upper anterior region should be >5.0 mm²; moreover, such connectors should be triangular, and have a gingival base and sufficient height in the loading direction.

References

1) Bahat Z, Mohamood DJH, Vult von Steyern P (2009) Fracture strength of three-unit fixed partial denture cores (Y-TZP) with different
connector dimension and design. Swed Dent J 33:149–159.
2) Dornhofer R, Arnetzl GV, Koller M, Arnetzl G (2007) Comparison of the static loading capacity of all-ceramic bridge frameworks in posterior teeth using three hardcore materials. Int J Comput Dent 10:315–328.
3) Drummond JL (1989) In vitro aging yttria-stabilized zirconia. J Am Ceram Soc 72:675–676.
4) Filser F, Kocher P, Weibel F, Lüthy H, Schärer P, Gauckler LJ (2001) Reliability and strength of all-ceramic dental restorations fabricated by direct ceramic machining (DCM). Int J Comput Dent 4:89–106.
5) Fischer H, Weber M, Marx R (2003) Lifetime prediction of all-ceramic bridges by computational methods. J Dent Res 82:238–242.
6) Geis-Gerstorfer J, Fäbler P (1999) Untersuchungen zum Ermüdung verhalten der dental keramiken zirkondioxid-TZP und In-Ceram. Dtsch Zahnarztl Z 54:692–694. (in German)
7) ISO 6872:2008, Dentistry—Ceramic materials.
8) Kagaya K, Minami I, Nakamura T, Sato M, Ueno T, Igarashi Y (2009) Three-dimensional analysis of occlusal curvature in healthy Japanese young adults. J Oral Rehabil 36:257–263.
9) Kelly JR (1999) Clinically relevant approach to failure testing of all-ceramic restorations. J Prosthet Dent 81:652–661.
10) Kelly JR, Tesk JA, Sorensen JA (1995) Failure of all-ceramic fixed partial dentures in vitro and in vivo: analysis and modeling. J Dent Res 74:1253–1258.
11) Kern M, Wegner SM (1998) Bonding to zirconia ceramic: adhesion methods and their durability. Dent Mater 14:64–71.
12) Kiliasidis S, Kjellberg H, Wennenberg B, Engstöm C (1993) The relationship between maximal bite force, bite force endurance, and facial morphology during growth. Acta Odontol Scand 51:323–331.
13) Kokich VO Jr, Kiyak HA, Shapiro PA (1999) Comparing the perception of dentists and lay people to altered dental esthetics. J Esthet Dent 11:311–324.
14) Lohbauer U, Ameberger G, Quinn GD, Scherrer SS (2010) Fractographic analysis of a dental zirconia framework: A case study on design issues. J Mech Behav Biomed Mater 3:623–629.
15) Mahmood DJH, Linderoth EH, Vult von Steyern P, Wennenberg A (2013) Fracture strength of all-ceramic (Y-TZP) three- and four-unit fixed dental prostheses with different connector design and production history. Swed Dent J 37:179–187.
16) Murase T, Nomoto S, Sato T, Shinya A, Koshihara T, Yasuda H (2014) Effect of connector design on fracture resistance in all-ceramic fixed partial dentures for mandibular incisor region. Bull Tokyo Dent Coll 55:149–155.
17) Nomoto S, Matsunaga S, Sato T, Yotsuya M, Abe S (2015) Basic finite element analysis of para-periodontal ligament in all-ceramic zirconia fixed partial denture. Bull Tokyo Dent Coll 56:215–222.
18) Oh W, Anusavice KJ (2002) Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures. J Prosthet Dent 87:536–542.
19) Oh W, Götzten N, Anusavice KJ (2002) Influence of connector design on fracture probability of ceramic fixed-partial dentures. J Dent Res 81:623–627.
20) Okabayashi S, Nomoto S, Sato T, Miho O (2013) Influence of proximal supportive design of zirconia framework on fracture load of veneering porcelain. Dent Mater J 32:372–377.
21) Onodera K, Sato T, Nomoto S, Miho O, Yotsuya M (2011) Effect of connector design on fracture resistance of zirconia all-ceramic fixed partial dentures. Bull Tokyo Dent Coll 52:61–67.
22) Pelaez J, Cogolludo PG, Serrano B, Lozano JL, Suarez MJ (2012) A prospective evaluation of zirconia posterior fixed dental prostheses: three-year clinical results. J Prosthet Dent 107:375–379.
23) Pereira JR, de Ornelas F, Conti PC, do Valle AL (2006) Effect of a crown ferrule on the fracture resistance of endodontically treated teeth restored with prefabricated posts. J Prosthet Dent 95:50–54.
24) Plengsombut K, Brewer JD, Monaco EA Jr, Davis EL (2009) Effect of two connector designs on the fracture resistance of all-ceramic core materials for fixed dental prostheses. J Prosthet Dent 101:166–173.
25) Polack MA (2006) Restoration of maxillary incisors with a zirconia all-ceramic system: a case report. Quintessence Int 37:375–380.
26) Pröbstler L (1993) Survival rate of In-Ceram restorations. Int J Prosthodont 6:259–263.
27) Raigrodski AR, Hillstead MB, Meng GK, Chung KH (2012) Survival and complications of zirconia-based fixed dental prostheses: A systematic review. J Prosthet Dent 107:170–177.
28) Sailer I, Fehér A, Filser F, Gauckler LJ, Lüthy H, Hämerle CH (2007) Five-year clinical results of zirconia frameworks for posterior fixed partial dentures. Int J Prosthodont 20:383–388.
29) Schmitter M, Müssotter K, Rammelsberg P, Stober T, Ohlmann B, Gabbert O (2009)
Clinical performance of extended zirconia frameworks for fixed dental prostheses: two-year results. J Oral Rehabil 36:610–615.

30) Strub JR, Gerds T (2003) Fracture strength and failure mode of five different single-tooth implant-abutment combinations. Int J Prosthodont 16:167–171.

31) Takamizawa T (1965) Studies on the co-relative and individual biting forces of normal permanent teeth. Nihon Hotetsu Shika Gakkai Zasshi 9:217–236, (in Japanese)

32) Takuma Y, Nomoto S, Sato T, Sugihara N (2013) Effect of framework design on fracture resistance of zirconia 4-unit all-ceramic fixed partial dentures. Bull Tokyo Dent Coll 54: 149–156.

33) Tang X, Nakamura T, Usami H, Wakabayashi K, Yatani H (2012) Effect of multiple firings on the mechanical properties and microstructure of veneering ceramics for zirconia frameworks. J Dent 40:372–380.

34) Vult von Steyern P, Carlson P, Nilner K (2005) All-ceramic fixed partial dentures designed according to the DC-Zircon technique. A 2-year clinical study. J Oral Rehabil 32:180–187.

35) Vult von Steyern P, Kokubo Y, Nilner K (2005) Use of abutment-teeth vs. dental implants to support all-ceramic fixed partial dentures. An in-vitro study on fracture strength. Swed Dent J 29:53–60.

36) Zheng Z, Lin J, Shinya A, Matinlinna JP, Botelho MG, Shinya A (2012) Finite element analysis to compare stress distribution of gold alloy, lithium-disilicate reinforced glass ceramic and zirconia based fixed partial denture. J Investig Clin Dent 3:291–297.

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