Check for updates

https://doi.org/10.4047/jap.2023.15.4.171

Effect of connector configuration on the fracture load in conventional and translucent zirconia three-unit fixed dental prostheses

Chonticha Subsomboon, Somchai Urapepon*
Department of Prosthodontics, Faculty of Dentistry, Mahidol University, Bangkok, Thailand

ORCID
Chonticha Subsomboon
https://orcid.org/0000-0002-0440-6011
Somchai Urapepon
https://orcid.org/0000-0002-4275-480X

PURPOSE. The purpose of this study was to determine the effect of the connector configuration on the fracture load in conventional and translucent zirconia of three-unit fixed dental prostheses (FDPs). **MATERIALS AND METHODS.** Six different three-unit FDPs were prepared (n = 6) from three types of zirconia (3Y-TZP (Katana ML®), 4Y-TZP (Katana STML®), and 5Y-TZP (Katana UTML®)) in combination with two connector configurations (4×2.25, 3×3 mm). The Co-Cr master models were scanned, and the FDPs were designed and fabricated using CAD-CAM. The FDPs were cemented on the metal model and then loaded with a UTM at a crosshead speed of 1 mm/min until failure. Two-way ANOVA and Tukey's test were used for statistical analysis ($\alpha = .05$). **RESULTS.** Fracture loads of 3Y-TZP (2740.6 ± 469.2 and 2718.7 ± 339.0 N for size 4×2.25 mm and 3×3 mm, respectively) were significantly higher than those of 4Y-TZP (1868.3 ± 281.6 and 1663.6 ± 372.7 N, respectively) and 5Y-TZP (1588.0 ± 255.0 and 1559.1 ± 110.0 N, respectively) ($P < .05$). No significant difference was found between fracture loads of 4Y-TZP and 5Y-TZP ($P > .05$). The connector configuration within 9 mm^2 was found to have no effect on the fracture loads on all three types of zirconia ($P > .05$). **CONCLUSION.** Fracture loads of three-unit FDPs were affected by the type of zirconia. The fracture loads of conventional zirconia were higher than those of translucent zirconia. However, it was not affected by the connector configuration when the connector had a cross-sectional area of 9 mm^2. [J Adv Prosthodont 2023;15:171-8]

KEYWORDS
Fracture load; Fixed dental prosthesis; Connector configuration; Conventional zirconia; Translucent zirconia

Corresponding author
Somchai Urapepon
Department of Prosthodontics, Faculty of Dentistry, Mahidol University, 6 Yothi Street, Rachathevi, 10400, Bangkok, Thailand
Tel +66 2 200 7817 18
E-mail S.urapepon@gmail.com

Received April 3, 2023 / **Last Revision** June 19, 2023 / **Accepted** August 10, 2023

 https://doi.org/10.4047/jap.2023.15.4.171

INTRODUCTION

Because of the strength of the metal framework and the aesthetic benefit of veneering porcelain, porcelain fused to metal (PFM) restorations have been widely utilized and have proven to be durable restorations for fixed dental prostheses (FDPs) for a long time. However, there are certain disadvantages, such as a conspicuous metallic color, an allergic reaction in some individuals, and porcelain chipping.[1] Partially stabilized zirconia, which can be created using a computer-aided design/computer-aided manufacturing (CAD-CAM) system, has recently been used in restorative dentistry due to the advancement of sophisticated dental ceramics. Because of their high fracture toughness, yttria-stabilized tetragonal zirconia polycrystals (Y-TZP) have been shown to be structural ceramics. However, Y-TZP has a high opacity that might cause aesthetic issues in dental restorations affecting the natural dental appearance. To enhance its translucency, transparent zirconia has been created as a monolithic material to replace bilayer zirconia, not only to eliminate veneer chipping, fracturing, or delamination but also to reduce the opacity and white coloring of the original zirconia.[2,3] The hardness of Y-TZP ceramics allows it to be used in the molar area, which is a unique property. However, more translucency of this monolithic zirconia will reduce its strength with regard to bridge failure, especially in the posterior region. Bridge failures occurred most often around the connector area between retainers and a pontic.[4,5] In the connector region, there is an increased risk of failure when the radius of curvature is increased.[6] Finite element (FE) modeling has shown that the highest stress concentration lies in the region of the connectors.[7] These considerations may be even more important in posterior FDPs. For the connector of a 3-unit Y-TZP framework, a cross-sectional area of 9.0 mm^2 or higher has been established as the desired size to achieve appropriate strength[6,8-10] and was recommended by several dental material manufacturers. A higher connection height was recommended to be effective to resist the occlusal load, rather than the connection width. However, in Asian people, the connection height is often limited by short clinical molar crowns, and larger loads are created. Due to the structure of the abutment tooth, securing the correct cross-sectional area and shape in terms of the height of the connector of an FDP in the molar region can be challenging. Therefore, we are interested in the size of the connector when the height is reduced to accommodate the teeth of Asian patients. The size of the connector 4 × 2.25 mm (9 mm^2) is the minimum preferable connector, compared with decreasing the height of the connector to 3 × 3 mm to obtain the same connector area (9 mm^2). In this study, we prepared Y-TZP frameworks for 3-unit FDPs for the mandibular posterior region with different connector configurations and compared their fracture loads. We also investigated the relationship between zirconia variation and fracture loading.

MATERIALS AND METHODS

In this experimental laboratory study, a mandibular typodont model was used (model ANA-4; Frasaco, Tettnang, Germany). The left second premolar and second molar were prepared for abutment of a three-unit bridge with a deep chamfer finish line (0.8 - 1 mm). The occluso-gingival abutment height was 5 mm. The gingiva of the residual ridge was reduced 3 mm in height to prepare the space for bridge bending during the fracture test. The prepared model was duplicated and cast with Co-Cr alloy (Vitallium; Dentsply Sirona, York, PA, USA) for use as a master model (Fig. 1).

The master models were scanned with the 3Shape Dental Scanner (Trios 3; 3Shape, Copenhagen, Denmark). Three-unit FDPs with two different connector

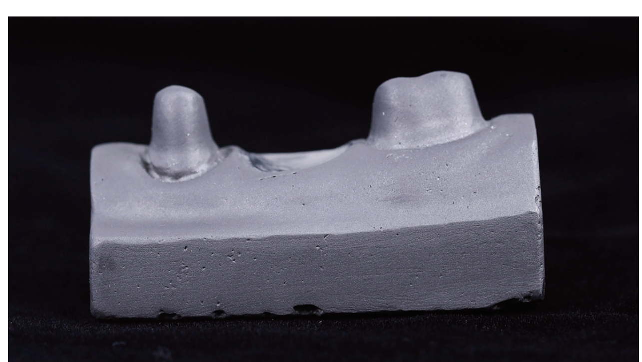

Fig. 1. Master model of abutment for three-unit fixed dental prosthesis.

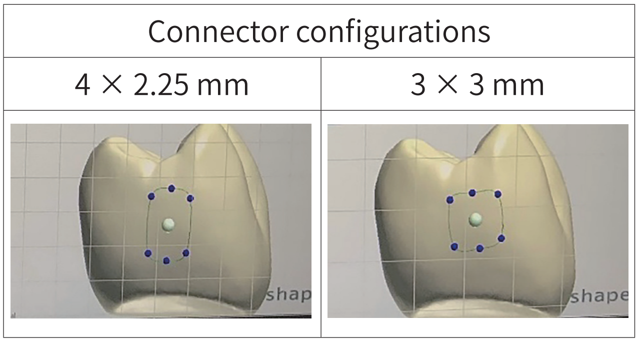

Fig. 2. Different types of connector shapes.

configurations (4 × 2.25, 3 × 3 mm) (Fig. 2) and a hygienic pontic were then designed using CAD software (3Shape Dental System; 3 shape). The data of the designed FDPs were sent to an InLab MC-X5 machine (Dentsply Sirona, York, PA, USA) to mill three-unit FDPs with three different translucent zirconia materials (Katana ML (3Y-TZP), Katana STML (4Y-TZP), Katana UTML (5Y-TZP); Kuraray Noritake, Chiyoda-ku, Japan). Then, final sintering and glazing were done with a VITA ZYRCOMAT 6000MS machine (VITA Zahnfabrik, Bad Sackingen, Germany).

A total of 36 experimental three-unit FDPs were milled with different connector configurations and materials and can be divided into six groups as follows in Table 1.

All 3-unit FDPs were cemented on the metal die using U200 resin cement (3M ESPE, Seefeld, Germany). The static load test was carried out using a universal testing machine (Model Instron 5566; Intron Ltd., Buckinghamshire, England) to assess the fracture load of the 3-unit FDPs. The vertical load was transferred through a 5 mm diameter steel ball at the center of the pontic at a cross-head speed of 1 mm/min until the fracture of the specimen occurred (Fig. 3). The fracture loads were collected and then statistically analyzed using two-way ANOVA and Tukey's test in SPSS v26.0 (IBM Corp., Armonk, NY, USA). The fracture characteristics of the fracture specimen were observed. Then, fractography was analyzed by scanning electron microscopy (Model JSM-6610LV; JEOL Ltd., Tokyo, Japan).

Table 1. Overview of the different groups of specimens

Group	Name	Connector configurations (mm)	Type of zirconia
I	Katana ML®	4 × 2.25	3Y-TZP
II	Katana ML®	3 × 3	3Y-TZP
III	Katana STML®	4 × 2.25	4Y-TZP
IV	Katana STML®	3 × 3	4Y-TZP
V	Katana UTML®	4 × 2.25	5Y-TZP
VI	Katana UTML®	3 × 3	5Y-TZP

RESULTS

Fracture loads are presented in Table 2. The mean fracture loads of 3Y-TZP were 2740.6 ± 469.2 N for a size of 4 × 2.25 mm, and 2718.7 ± 339.0 N for size 3 × 3 mm. The mean fracture loads of 4Y-TZP were 1868.3 ± 281.6 for size 4 × 2.25 mm, and 1663.6 ± 372.7 N for size 3 × 3 mm. The mean fracture loads of 5Y-TZP were 1588.0 ± 255.0 for size 4 × 2.25 mm, and 1559.1 ± 110.0 N for size 3 × 3 mm.

Two-way ANOVA (Table 3) showed that the fracture load was significantly different among the different zirconia types. Moreover, the fracture load was not significantly different by connector configurations 4 × 2.25 and 3 × 3 for the three groups (3Y-TZP, 4Y-TZP, and 5Y-TZP) ($P > .05$), and there was no interaction between the two factors. Tukey's post hoc test showed that the fracture loads in 4Y-TZP ($P < .05$) and 5Y-TZP

Fig. 3. Position of the 5 mm steel ball on the fixed dental prostheses.

https://doi.org/10.4047/jap.2023.15.4.171

Table 2. The fracture load of zirconia three-unit FDPs (N ± SD)

Zirconia / Connector configurations	Mean fracture loads (N) ± SD 4 × 2.25 mm	3 × 3 mm
Katana ML (3Y-TZP)	2740.6 ± 469.2 ^{A,a}	2718.7 ± 339.0 ^{A,a}
Katana STML (4Y-TZP)	1868.3 ± 281.6 ^{B,b}	1663.6 ± 372.7 ^{B,b}
Katana UTML (5Y-TZP)	1588.0 ± 255.0 ^{B,c}	1559.1 ± 110.0 ^{B,c}

Within the same connector size (vertical column), the values marked by the same superscript uppercase letter are not significantly different (P > .05).
Within the same zirconia type (horizontal row), the values marked by the same superscript lowercase letter are not significantly different (P > .05).

Table 3. Results of Two-way ANOVA

	Type III SS	df	MS	F	Sig
Zirconia	9208559.42	2	4604299.71	43.82	< .001
Connector size	65266.63	1	65266.63	0.62	.44
Zirconia × connector size	643680.17	2	32184.09	0.31	.74
Error	3151921.71	30	105064.06		
Total	159828043.00	36			

were significantly lower than those in 3Y-TZP. No significant difference was found between 4Y-TZP and 5Y-TZP (P > .05) (Table 3).

All zirconia bridges revealed catastrophic bulk fractures. The fracture pattern was found to be the same for all groups. Most of the fractures were associated with fractures that initiated from occlusal surfaces of the pontic and extended to the bottom of the connector at mesial and distal portions of connector. The fractures always occurred in the base of the connector, spreading to the margin of the crown, which was less thick. Crown fractures in the second molar were mostly found on the lingual side (72.2%), and some were found in the second premolar on the buccal side (41.7%) (Fig. 4). The margin of molar crown thickness at buccal was 0.9 mm and at lingual was 0.8 mm. The margin of premolar crown thickness at buccal was 0.8 mm and at lingual was 0.9 mm. The fracture was observed in a plane perpendicular to the occlusal surface of the pontic. The fracture surface examina-

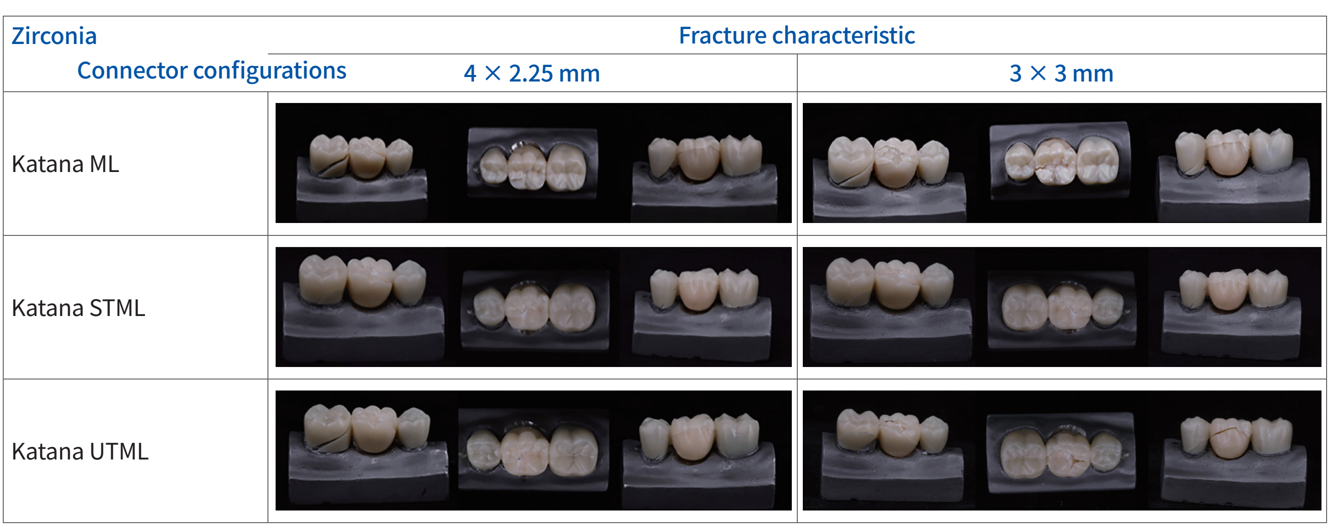

Fig. 4. Typical fracture in each group (Each group showed lingual, occlusal, and buccal view, respectively).

tions show the general direction of crack propagation as evidenced by the origin point, and hackles can be mapped on the fractured parts. The fracture started from the occlusal surface of the pontic and propagated toward the bottom of the connector (Fig. 5, Fig. 6, Fig. 7).

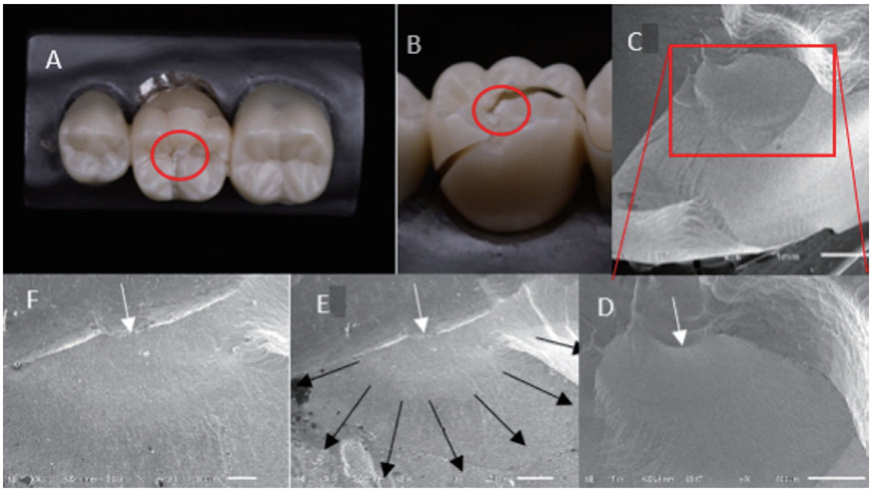

Fig. 5. Fracture direction mapping at occlusal compress site. (A) Side view of fracture, (B, C) Occlusal view of the bulk fractured specimen, SEM images: (D) Overview of the fractured part at low magnification (× 16), (E, F) Detailed view at high magnification (× 50, × 100): black arrow represents the direction of crack propagation.

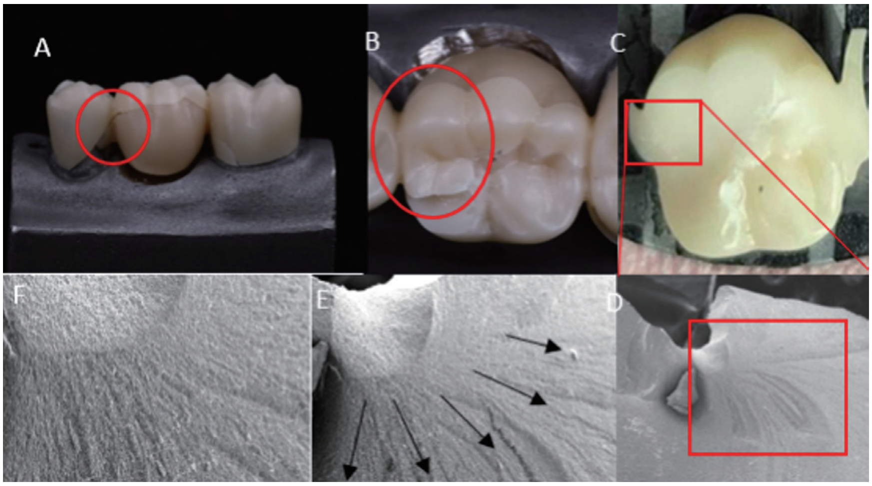

Fig. 6. Fracture direction mapping at mesial connector site. (A) Side view of fracture, (B, C) Occlusal view of the bulk fractured specimen, SEM images: (D) Overview of the fractured part at low magnification (× 16), (E, F) Detailed view at high magnification (× 50, × 100): black arrow represents the direction of crack propagation.

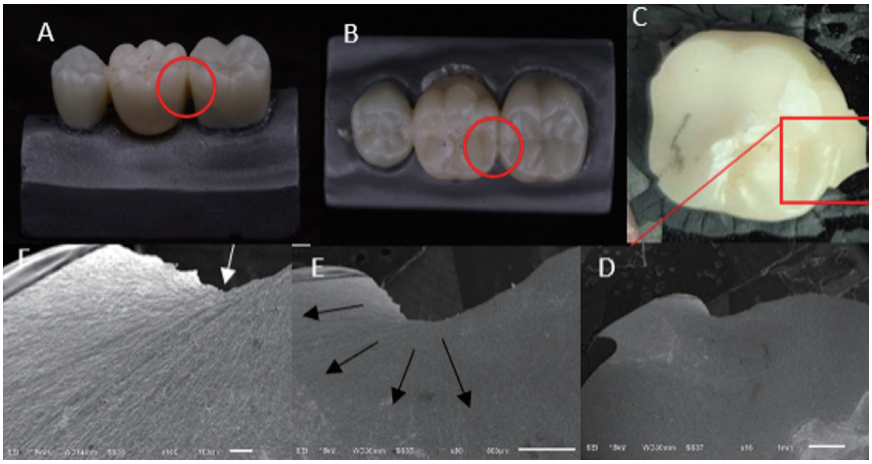

Fig. 7. Fracture direction mapping at distal connector site. (A) Side view of fracture, (B, C) Occlusal view of the bulk fractured specimen, SEM images: (D) Overview of the fractured part at low magnification (× 16), (E, F) Detailed view at high magnification (× 50, × 100): black arrow represents the direction of crack propagation.

https://doi.org/10.4047/jap.2023.15.4.171

DISCUSSION

This *in vitro* study was designed to mimic the anatomic configuration of teeth in order to more accurately model anatomic stressors and more consistently evaluate risk factors for the fracture of three-unit FDPs.[4] A bar design that follows the mathematical beam theory does not reflect any clinical significance. Generally, a test design with immovable supports usually results in a higher load bearing capacity, overestimating the capability of the material.[8] In addition, model Co-Cr was used as the supporting structure due to its high modulus of elasticity. Therefore, it may lead to a higher value of fracture load than the clinical estimates.

In this study, there was a significant difference between conventional (3Y-TZP) and translucent zirconia (4Y-TZP, 5Y-TZP) in fracture load. Generally, conventional zirconia contains 0.5 to 1.0% wt Al_2O_3 and 3 - 6% Y_2O_3. Translucent zirconia with reduced Al_2O_3 content and increased Y_2O_3 content improved the optical properties, whereas these factors also reduced the mechanical properties.[11,12] This phenomenon can be explained by Kolakarnprasert *et al*.,[13] who reported that 3Y-TZP, 4Y-TZP, and 5Y-TZP have average grain sizes of 0.6 ± 0.03, 2.8 ± 0.2, and 4.1± 0.9 µm, respectively. These findings are in line with the yttria content and cubic proportion since cubic grains are noticeably larger than tetragonal grains. The zirconia transformation toughening mechanism can be impacted by grain sizes that are too small (in the 200 nm range). The majority of the grain size is larger than 1 µm, and the simultaneous t → m transformation upon cooling from the sintering temperature can reduce the strength of 3Y-TZP.[14] Therefore, a grain size too large or too small will affect the strength of the zirconia. Additionally, larger grain sizes also result in a decrease in the grain boundary, which results in reduced light scattering.[15] Thus, the high translucency of large grain sizes relies on light scattering.

However, in this study, translucent zirconia (4Y-TZP, 5Y-TZP) was not found to have any difference in fracture loads. However, 4Y-TZP had 8.5 wt% Y_2O_3 content and no detectible Al_2O_3, and 5Y-TZP had 9.5 wt% Y_2O_3 content and < 0.01 wt% Al_2O_3.[13] In fact, increasing the translucency results in a decrease in strength. Therefore, three-unit fixed dental prostheses are more easily fractured when translucency is increased. The advantages and disadvantages of each material should be considered before the final decision is made for appropriate clinical performance to achieve the best success rate and longevity of the restorations.

For strength, the connector used in the posterior bridge should have a cross-sectional area of more than 9 mm^2.[7,10] The fracture load decreased as the cross-sectional area of the connector decreased. Several studies have suggested that the connector dimension should be larger vertically than horizontally.[6,7,16-18] However, clinically, it is already difficult to achieve a connector that is larger in the vertical direction than in the horizontal direction unless the space for hygiene is compromised. Additionally, this study was challenged by reducing the vertical height because it is difficult to design a connector to be vertical rather than horizontal due to Asian people's typically having short posterior teeth. In the present study, a connector cross-sectional area of 9 mm^2 can be used for three-unit fixed dental prostheses. Therefore, in this study, we adopted a cross-sectional area of 9 mm^2 with connector configuration sizes of 4 × 2.25 mm and 3 × 3 mm. The mean fracture load ranged from 2740.6 - 1559.1 N. The maximum occlusal force was 250 - 400 N in the posterior teeth,[19] and other authors[20,21] suggested that parafunctional occlusal force was assumed to be 500 - 880 N. Therefore, the fracture load in this study was over 880 N in all groups with a cross-sectional area of 9 mm^2. There was no significant difference among connector configurations. In the present study, a connector cross-sectional area of 9 mm^2 can be used for three-unit fixed dental prostheses in posterior teeth.

The connector area was the most influential factor in fractures.[6,7,10,18,22] The fracture rate was relatively high in three-unit fixed dental prostheses around the connector area. The location of the fracture indicates that fractures are initiated by tension at the cervical margin or gingival embrasure of the connector. According to a previous study, a gingival embrasure was the site of connection fracture initiation. They concluded that using a larger radius of curvature at the gingival embrasure decreased the magnitude of tensile stress and improved the fracture resistance of the FDPs.[6] However, in the present study, it was fractured at the pontic and crown rather than the connector area. Although no fracture was discovered at the connector if the connector was strong enough, this study discovered a destructive fracture on the occlusal side as a result of the contact between the round ball and the pontic. Additionally, it was found that second molar crown fractures were predominantly found on the lingual side, while second premolar crown fractures were mostly found on the buccal side. The margin is seen as the weak component, which is perhaps why the restorations' margin failed by 25.7%. As a result of its thin restoration thickness, this region is prone to breaking.[23] Undoubtedly, a sufficient thickness of the zirconia material in all of these areas must be provided to ensure an adequate functional life of the restoration.

This study's limitations are that it only performs height-reduced connection configuration testing. Moreover, the absence of the periodontal ligament and the metal alloy abutment, which differed from that of natural teeth, did not accurately reflect the clinical environment. For FDPs in the posterior region, a higher flexural strength is required to counteract the greater occlusal force. The use of zirconia materials for 3-unit FDPs in the posterior region appears promising considering the results of this study. However, repetitive dynamic functional loading or aquatic environmental conditions were not mimicked in the present study. In addition, all loads in this study were applied axially to the pontic center. Variations in the shape of failure in 3-unit zirconia FDPs may be caused by the clinical contact surfaces on the pontic and adjacent abutments. As a result, there may be some discrepancy between the results and the values measured intraorally. Although it is difficult to compare the results obtained with the data from clinical trials, it is possible that a connector with a cross-sectional area of 9 mm^2 is practical. Therefore, future studies should concentrate on connector configurations that are wider rather than higher and should be performed in an oral environment.

CONCLUSION

The fracture loads of three-unit fixed dental prostheses were significantly affected by the type of zirconia. The fracture loads of conventional zirconia were significantly higher than those of translucent zirconia. However, it was not affected by the connector configuration when the connector had a cross-sectional area of 9 mm^2.

ACKNOWLEDGMENTS

The author would like to thank the Department of Prosthodontics, Faculty of Dentistry, Mahidol University for financial support and the Research Service section for searching and providing equipment and materials in the laboratory and supporting and developing equipment and materials.

REFERENCES

1. Joiner A. Tooth colour: a review of the literature. J Dent 2004;32 Suppl 1:3-12.
2. Dangra Z, Gandhewar M. The use of newer high translucency zirconia in aesthetic zone. Case Rep Dent 2014;2014:432714.
3. Tuncel İ, Turp I, Üşümez A. Evaluation of translucency of monolithic zirconia and framework zirconia materials. J Adv Prosthodont 2016;8:181-6.
4. Kelly JR, Tesk JA, Sorensen JA. Failure of all-ceramic fixed partial dentures in vitro and in vivo: analysis and modeling. J Dent Res 1995;74:1253-8.
5. Dittmer MP, Kohorst P, Borchers L, Stiesch-Scholz M. Finite element analysis of a four-unit all-ceramic fixed partial denture. Acta Biomater 2009;5:1349-55.
6. Oh WS, Anusavice KJ. Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures. J Prosthet Dent 2002;87:536-42.
7. Pöppel ML, Rosentritt M, Sturm R, Beuer F, Hey J, Schmid A, Schmidt F. Fracture load and fracture patterns of monolithic three-unit anterior fixed dental prostheses after in vitro artificial aging-a comparison between color-gradient and strength-gradient multilayer zirconia materials with varying yttria content. J Clin Med 2022;11:4982.
8. Larsson C, Holm L, Lövgren N, Kokubo Y, Vult von Steyern P. Fracture strength of four-unit Y-TZP FPD cores designed with varying connector diameter. An in-vitro study. J Oral Rehabil 2007;34:702-9.
9. Filser F, Kocher P, Weibel F, Lüthy H, Schärer P, Gauck-

https://doi.org/10.4047/jap.2023.15.4.171

ler LJ. Reliability and strength of all-ceramic dental restorations fabricated by direct ceramic machining (DCM). Int J Comput Dent 2001;4:89-106.

10. Onodera K, Sato T, Nomoto S, Miho O, Yotsuya M. Effect of connector design on fracture resistance of zirconia all-ceramic fixed partial dentures. Bull Tokyo Dent Coll 2011;52:61-7.
11. Zhang F, Inokoshi M, Batuk M, Hadermann J, Naert I, Van Meerbeek B, Vleugels J. Strength, toughness and aging stability of highly-translucent Y-TZP ceramics for dental restorations. Dent Mater 2016;32:e327-37.
12. Tabatabaian F. Color aspect of monolithic zirconia restorations: a review of the literature. J Prosthodont 2019;28:276-87.
13. Kolakarnprasert N, Kaizer MR, Kim DK, Zhang Y. New multi-layered zirconias: Composition, microstructure and translucency. Dent Mater 2019;35:797-806.
14. Denry I, Kelly JR. State of the art of zirconia for dental applications. Dent Mater 2008;24:299-307.
15. Zhang Y. Making yttria-stabilized tetragonal zirconia translucent. Dent Mater 2014;30:1195-203.
16. Takuma Y, Nomoto S, Sato T, Sugihara N. Effect of framework design on fracture resistance in zirconia 4-unit all-ceramic fixed partial dentures. Bull Tokyo Dent Coll 2013;54:149-56.
17. Esquivel-Upshaw JF, Young H, Jones J, Yang M, Anusavice KJ. Four-year clinical performance of a lithia disilicate-based core ceramic for posterior fixed partial dentures. Int J Prosthodont 2008;21:155-60.
18. Fayez SH. Katamish, Elguindy J. Effect of connector dimension on the fracture characteristics of cantilever posterior zirconia bridges. Ainshams Dent J 2017:71-84.
19. Fontijn-Tekamp FA, Slagter AP, Van Der Bilt A, Van ‘T Hof MA, Witter DJ, Kalk W, Jansen JA. Biting and chewing in overdentures, full dentures, and natural dentitions. J Dent Res 2000;79:1519-24.
20. Kelly JR, Denry I. Stabilized zirconia as a structural ceramic: an overview. Dent Mater 2008;24:289-98.
21. Sakaguchi RL, Powers JM., Craig’s restorative dental materials. 13th ed. Mechanical properties. St. Louis; Mosby; 2012. p. 355-67.
22. Salimi H, Mosharraf R, Savabi O. Effect of framework design on fracture resistance of zirconium oxide posterior fixed partial dentures. Dent Res J (Isfahan). 2012;9:764-9.
23. Oilo M, Hardang AD, Ulsund AH, Gjerdet NR. Fractographic features of glass-ceramic and zirconia-based dental restorations fractured during clinical function. Eur J Oral Sci 2014;122:238-44.