**A Novel Design Modification to Improve Flexural Strength of Zirconia Framework: A Comparative Experimental In Vitro Study**

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**Aim:** Zirconia-based restoration is successfully replacing metal ceramic restorations in posterior areas. Although higher mechanical properties of zirconia, their use in compromised situation is questionable. Hence, there is a need to modify the design which to strengthen the framework. The aim of this in vitro study was to evaluate the influence of lingual collar design on the flexural strength of CAD/CAM-fabricated posterior three-unit zirconia framework.

**Materials and Methods:** A mandibular metallic stainless steel master mold is designed for a three-unit fixed partial denture framework. All CAD-milled 20 samples are divided into two groups based on the design. Group A—with collar (10 samples) and Group B—without collar (10 samples), tested using universal testing machine to calculate the mean fracture load and flexural strength.

**Statistical Analysis Used:** Descriptive statistics and independent sample t test were used to find the difference between the groups, and simple linear regression was used to find the relationship between load and displacement between the groups.

**Results:** The result of the mean flexural strength for Group A was 11328.06 ± 3770 MPa and for Group B was 7633.95 ± 3196 MPa; the mean fracture strength observed for Group A was 1274.04 ± 424 MPa and for Group B was 858.80 ± 359 MPa. A statistically significant difference was observed in flexural strength between Groups A and B (P < 0.05).

**Conclusion:** Zirconia framework with connector dimension of 7 mm² with lingual collar design can be successfully incorporated in compromised situation where an ideal connector dimension of 9 mm² cannot be placed.

**Keywords:** CAD/CAM zirconia, connector, lingual collar

**INTRODUCTION**

The last few decades have witnessed an era of aesthetic and biomimetic dentistry wherein; there has been an increase in the desire and demands of patient’s to have posterior metal-free restorations. An equilibrium is desired between conservation of tooth structure and resemblance to natural dentition in terms of form, appearance, and strength.

The field of restorative dentistry has seen a better alternative in the form of yttria-stabilized tetragonal zirconia polycrystalline (Y-TZP) because of its excellent biocompatibility, esthetics, and transformation toughening property, enabling it to have a high flexural strength (up to 1200 MPa) and fracture toughness (6–9 MPa m¹/²) when compared to other all ceramic systems.\(^{[1]}\)

Recently, CAD/CAM technology has led to the development of high-strength ceramics, thereby...
facilitating the fabrication of well-fitting frameworks having increased flexural strength and fracture toughness.

Zirconia-based long-span frameworks had problems initially with regard to the fit of the framework, leading to some clinical complications. However, due to recent advancements in CAD/CAM technology these problems have been resolved.\cite{2} Wimmer et al.\cite{2} found that conventional fixed dental prostheses (FDPs) were better CAD/CAM in relation to load values. Even though Y-TZP-based restorations had higher mechanical properties, clinical failures commonly occurred at the connector area of fixed dental prostheses.\cite{1}

In fixed partial denture (FPDs), connector thickness are relatively thin when compared to other areas of the framework. Hence during loading especially in the posterior FPDs, stress concentration occurs at the connector area leading to failure of the prosthesis.

The clinical recommendations for Y-TZP-based FDP connector thickness varies from 2 to 4mm in height and width.\cite{1} Larsson et al.\cite{2} recommended a minimum diameter of 4mm. Schmitter et al.\cite{2} recommended the use of YTZP-based FDPs with a connector dimension of 9mm². Onodera et al.\cite{2} proposed a connector with a cross-sectional area of 7.0 or 9.0mm². Kamosiora et al.\cite{2} in their two-dimensional finite-element stress analysis study, concluded that increasing the connector height dramatically reduces the stress levels within the connectors.

Studies show crack propagation which led to initial fracture originated from the gingival embrasure of connector.\cite{2} Certain design modifications such as lingual collar with proximal struts were incorporated within crowns to enhance strength of zirconia-veneered and metal-ceramic crowns. These crowns were then subjected to loading in the marginal ridge areas. It was concluded that zirconia veneered crowns presented decreased fracture rates when compared to metal ceramics regardless of framework design.\cite{2}

Although there is specification for the ideal connector designs, there is no agreement in literature, with regard to posterior FPD connector design in situations such as decreased clinical crown height and compromised ridge contour wherein the ideal connector dimensions are difficult to maintain. There is a need to strengthen the connector and to improve the flexural strength of zirconia three-unit framework in these compromised situations. Further there is no evidence in literature that cutback copings with lingual collar extending from below the pontic up to the connectors would strengthen the connector areas which in turn increases the flexural strength of the prosthesis.

Therefore, this study is undertaken to determine the effect of lingual collar design on the flexural strength of CAD/CAM-fabricated posterior three-unit zirconia coping framework.

**Materials and Methods**

This study was approved by YEC-1 (Yenepoya Ethics Committee-1) with the protocol number 2017/031 titled “Evaluation of the influence of collar design on the flexural strength of CAD/CAM fabricated posterior three-unit zirconia framework.”

**Preparation of the metallic master model**

A mandibular metallic stainless steel master mold with retrievable dies of second premolar and second molar as abutment teeth is designed for a three-unit FDP framework. The diameter of the abutment will be 7.0mm and 11.0mm, corresponding to the second premolar and second molar, respectively. The axial surface of the abutments will have a taper of 6° and a height of 5.0mm. A deep chamfer finish line having a curvature radius of 1.0mm. The distance between the two abutments is 20.0mm.

**Preparation of posterior three-unit zirconia coping framework**

The master mold base will be duplicated using polyvinyl siloxane impression material and poured with scannable Type IV gypsum. The stainless steel dies with duplicated mold will then be scanned with a Ceramill Map400 Scanner, Amann Girrbach AG (Koblach, Austria). Zirconia-coping framework corresponding to two design groups will be fabricated using Ceramill Mind software, Amann Girrbach AG (Koblach, Austria) and STL file, Amann Girrbach AG (Koblach, Austria) is obtained. The designed framework is nested in Zi Blank (12mm), Amann Girrbach AG (Koblach, Austria) and connectors are designed carefully to maintain the collar dimensions during the milling process.

![Figure 1: Connector placement of 7 mm² cross-sectional area](image-url)
All 20 samples of CAD-milled posterior three-unit zirconia cutback coping framework will have an even thickness of 0.6 mm and cement space of 0.05 mm [Figure 1].

These 20 samples are divided into two groups based on the design.

The two designs fabricated are as follows:

- **Group A (with collar):** Connector cross-sectional area of 7 mm$^2$ with lingual collar of 1 mm thick and 2 mm height (10 samples) extending from below the pontic up to the connector [Figure 2]
- **Group B (without collar):** Connector cross-sectional area of 7 mm$^2$ (10 samples) [Figure 3]

A total of 20 samples of three-unit zirconia copings ($n = 10$ for Group A and $n = 10$ for Group B) will be milled from White zirconia blanks using CAD/CAM system. All the milled copings will be then placed on a firing tray and sintered in a furnace for 10 h with an average rise in temperature of 8°C/min and peak temperature of 1450°C with a holding time of 2 h. Each framework was cemented to the test model according to the manufacturer’s instructions using self-adhesive resin cement (3M ESPE AG (Seefeld, Germany) self-adhesive resin cement), then the cemented framework was immersed in distilled water (37°C) for 24 h [Figure 4].

**Evaluation of the flexural strength**

All samples were tested using universal testing machine and loaded axially at the center of the pontic by using steel ball of diameter 3 mm with a constant cross-head speed of 1.0 mm/min using universal testing machine. The load will be applied at the center of the pontic axially until it fractures. The peak load $F (N)$ was recorded at the fracture point and the fracture modes was observed; flexural strength $\sigma$ (MPa) was calculated using the following formula:[i]

$$\sigma = \frac{F \times \text{span length}}{\text{radius}^3},$$

where $F$ is the maximum load at fracture (N), $L$ is the length of the specimen (mm), and $R$ is the radius of the connector.

**Statistical analysis**

Convenient sampling method was used with a minimum sample size required 10 per group, total of 20 samples with $\alpha = 5\%$ level of significance, power $\beta = 80\%$, using G*Power 3.1 software, (Dusseldorf, Germany).
Descriptive analysis and independent sample t test were used to see the difference between two independent groups (Group A—with collar and Group B—without collar). Simple linear regression was used to find the relation between load and displacement between the two groups.

**RESULTS**

Results showed that the mean fracture load for Group A was 1274.12 N and for Group B was 859.19 N. The mean flexural strength recorded for Group A was 11328.06 ± 3770 MPa and for Group B was 7633.95 ± 3196 MPa. The mean fracture strength observed for Group A was 1274.04 ± 424 MPa and for Group B was 858.80 ± 359 MPa. A statistically significant difference was observed in flexural strength between Groups A and B (P < 0.05). Also the displacement of framework in Group A (0.004 units) was lesser compared to Group B (0.006 units) when loaded axially.

**DISCUSSION**

Esthetics has played a great role in the demand by patients to have posterior tooth color restorations. Earlier posterior all ceramic restorations lacked strength; hence, its use was limited to single crowns, or in areas where there was less functional force. The flexural strength of earlier all ceramic crowns ranged from 120 to 150 MPa, which was much below the compressive force experienced by natural teeth. As the flexural strength of all ceramic materials was less, this led to the introduction of Y-TZP (900–1000 Mpa). These CAD/CAM-milled restorations are predominantly used nowadays as fixed bridges to replace posterior teeth.

The mean adult occlusal force ranges between 400 and 800 N in the posterior area, 300 N in premolar area, and 200 N in the anterior area. As occlusal loads are maximum for posterior area, there is a need to strengthen the framework to withstand high occlusal load. Fully stabilized zirconia core is the material of choice with high values of flexural strength and fracture toughness.

In spite of zirconia material, possessing high flexural strength there have been instances of fracture of these crowns and frameworks in certain clinical conditions such as decreased clinical crown height and compromised ridge contour. These fractures originated from the gingival embrasure up to the occlusal surface at the connector site between the pontic and retainer.

Fardin et al. incorporated a design modification to reduce fracture rates in single crowns and came to the conclusion that the proposed design modification did not influence the fracture rates.

In this study, a similar design modification proposed by Fardin et al. for metal ceramics was incorporated to a posterior three-unit zirconia framework to check its influence on fracture rates.

This study was conducted in CAD/CAM dental laboratory of Yenepoya University; all the 20 samples of CAD-milled Y-TZP posterior three-unit zirconia cutback coping framework had an even thickness of 0.6 mm and connector dimension of 7 mm. These 20 samples consisted of two groups: Group A (with collar—10 samples) with modified connector design and Group B (without collar—10 samples) with conventional connector design. These 20 samples were luted to the metal die and checked for fracture strength under universal testing machine. The values obtained are recorded in Table 1 (with collar) and Table 2 (without collar).

In Table 1, Group A (with collar), the highest load recorded to fracture the specimen was 212 kg and the lowest was 53.22 kg. The highest fracture strength recorded was 2088.23 MPa and lowest was 521.91 MPa. The highest flexural strength recorded was 18561.95 MPa and the lowest was 4639.11 MPa.

| Sample | Load (kg) | Fracture strength (MPa) | Flexural strength (MPa) | Displacement (mm) |
|--------|-----------|-------------------------|------------------------|------------------|
| 1      | 130.66    | 1281.34                 | 11389.6                | 1.11             |
| 2      | 118.13    | 1158.46                 | 10297.33               | 1.00             |
| 3      | 156.03    | 1530.13                 | 13601.15               | 1.08             |
| 4      | 212.94    | 2088.23                 | 18561.95               | 1.18             |
| 5      | 53.22     | 521.91                  | 4639.11                | 0.56             |
| 6      | 167.30    | 1640.65                 | 14609.68               | 1.21             |
| 7      | 106.21    | 1041.56                 | 9258.31                | 0.92             |
| 8      | 95.61     | 937.61                  | 8334.31                | 0.89             |
| 9      | 123.09    | 1207.10                 | 10729.77               | 1.18             |
| 10     | 136.05    | 1334.91                 | 11859.46               | 1.13             |
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In Table 2, Group B (without collar), the highest load recorded to fracture the specimen was 154 kg and the lowest was 44.90 kg. The highest fracture strength recorded was 1518.17 MPa and lowest was 440.32 MPa. The highest flexural strength recorded was 13494.75 MPa and the lowest was 3911.11 MPa.

The results [Table 3] and [Graph 1] indicate that the mean and the standard deviation of load, flexural strength, and fracture strength of Group A (with collar) were higher when compared to Group B (without collar). The mean fracture load applied on the framework with lingual collar was higher than the maximum natural occlusal force in the posterior area (400–800 N).[11]

The mean fracture load of Group A was 129.9240 kg (1274.12 N) and for Group B was 87.6150 kg (859.21 N), which exceeded 800 N in all frameworks. This is due to the inherent strength of fully stabilized Y-TZP and to the use of cutback coping design, which assured even thickness of the framework. In this study, the flexural strength was calculated using a formula suggested by Hamza et al.[1] and the results of this study were in confirmation with the study conducted by Hamza et al.[1]

The results of independent sample t test performed [Table 4] indicated statistically significant difference in flexural strength between Groups A and B (P < 0.05). In this study, even though the same dimension for connector (7 mm²) with a diameter of 3 mm and round in shape was used for both the groups, the presence of lingual collar in Group A provided overall strength to the framework. The strength of the framework with collar increased due to the reinforcing effect of lingual collar which helped to withstand the higher amount of load, thereby leading to uniform distribution of stress.

The additional finding from the study was, in the case of framework with collar (Group A), for every 1 unit of load applied there was a displacement of 0.004 units, whereas in the case of specimen in (Group B) there was a displacement of 0.006 units, which is significant, indicating that framework with collar is better than that of without collar [Tables 5 and 6] [Graphs 2 and 3].

This reduced amount of displacement in Group A is due to bracing effect of the lingual collar from below the pontic up to the connectors and its extension to the mesial and distal retainer on both sides.

In this study, when compressive force was applied directly on the occlusal surface of the pontic, crack initiation occurred from within its gingival embrasure area due to tensile stress leading to fracture, in agreement with other studies.[12-15] This phenomenon occurs as the maximum stresses are seen in the connector which also happens to be the thinnest portion of the FPD.

Analysis of fracture site in Group A (with collar) revealed the mode of fracture was on the mesial side in 9 of 10 samples (90%). Interestingly samples had fractured on mesial side of the pontic [Figure 5]. The maximum natural occlusal load in the premolar area is 300 N[11] and the results of this study showed higher

| Sample | Load (kg) | Fracture strength (MPa) | Flexural strength (MPa) | Displacement (mm) |
|--------|-----------|------------------------|------------------------|-------------------|
| 1      | 67.60     | 662.93                 | 5892.62                | 0.50              |
| 2      | 44.90     | 440.32                 | 3911.11                | 0.52              |
| 3      | 52.31     | 512.99                 | 4559.82                | 0.58              |
| 4      | 82.15     | 805.62                 | 7160.97                | 0.83              |
| 5      | 154.81    | 1518.17                | 13494.75               | 1.11              |
| 6      | 110.42    | 1082.85                | 9625.33                | 0.91              |
| 7      | 131.27    | 1287.32                | 11442.75               | 1.05              |
| 8      | 106.69    | 1046.27                | 9300.17                | 0.95              |
| 9      | 68.45     | 671.27                 | 5966.75                | 0.59              |
| 10     | 57.19     | 560.84                 | 4985.24                | 0.47              |

| Descriptive statistics          | Mean   | Std. deviation |
|---------------------------------|--------|----------------|
| Load (kg)                       | Group A—with collar | 129.9240 | 43.22721 |
|                                 | Group B—without collar | 87.6150 | 36.65985 |
| Flexural strength (Mpa)         | Group A—with collar | 11328.0670 | 3770.64859 |
|                                 | Group B—without collar | 7633.9510 | 3196.50488 |
| Fracture strength (Mpa)         | Group A—with collar | 1274.0400 | 424.06851 |
|                                 | Group B—without collar | 858.8050 | 359.52983 |

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mean fracture load of 1274.12 N, which is relatively high. The reason for this mesial fracture would be due to the incorporation of lingual collar led to smaller radius of gingival embrasure on the mesial side compared to that on the distal side,\(^{16}\) which further led to stress concentration near the mesial connector leading to fracture.

However, analysis of fracture site in Group B (without collar) revealed that 8 of 10 samples (80%) showed fracture on the distal side of the connector and 2 of

### Table 4: Independent sample t test performed to check whether there is any difference in flexural strength between Group A (with collar) and Group B (without collar)

| Flexural strength (Mpa) | Sig. (two tailed) | 95% Confidence interval of the difference |
|-------------------------|-------------------|------------------------------------------|
|                         | .030              | –6978.24319 –409.98881                   |

\(t_{18} = –2.363\) and \(P < 0.05\) considered as significant

There is a difference between flexural strength in Group A (with collar) and Group B (without collar)

### Table 5: Simple linear regression analysis for Group A (with collar)

| Group A—-with collar | Unstandardized coefficients | Standardized coefficients | \(t\) | Sig. |
|-----------------------|-----------------------------|---------------------------|------|------|
| \(B\)                 | .004                        | .001                      | .834 | 4.275| .003 |
| \(P\)                 |                             |                           |      | \(< .05\) considered as significant |

### Table 6: Simple linear regression analysis for Group B (without collar)

| Group B—-without collar | Unstandardized coefficients | Standardized coefficients | \(t\) | Sig. |
|-------------------------|-----------------------------|---------------------------|------|------|
| \(B\)                  | .006                        | .001                      | .950 | 8.592| .000 |
| \(P\)                  |                             |                           |      | \(< .05\) considered as significant |

1. Graph 1: Bar chart showing mean flexural strength of Groups A and B. 1 represents Group A—-with collar and 2 represents Group B—-without collar.
2. Table 4: Independent sample t test performed to check whether there is any difference in flexural strength between Group A (with collar) and Group B (without collar)
3. Table 5: Simple linear regression analysis for Group A (with collar)
4. Table 6: Simple linear regression analysis for Group B (without collar)
10 samples (20%) on the mesial side of the connector [Figure 6], which is in agreement with a study conducted by Onodera et al.[6] This tendency of fracture is attributed to the larger distance between the center of the distal abutment and the middle of the pontic when compared to the distance between the center of the mesial abutment and the middle of the pontic.[6]

This study focuses on the cutback coping, which is anatomically similar to the final crown. In this study, cutback coping showed similar fracture strength (Group B—without collar) 858 ± 341 when compared to the offset coping without collar design used in the study conducted by Onodera et al.[6] having fracture strength of 980 ± 61 N. This indicates that cutback coping has similar fracture resistance than the offset copings without lingual collar. A study conducted by Guess et al.[17] stated that anatomic core design

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**Graph 2:** Load-displacement diagram of Group A—with collar

**Graph 3:** Load-displacement diagram of Group B—without collar

**Figure 5:** Fracture site for Group A
Modification significantly increased the reliability and resulted in reduced chip size of veneering ceramics.

This in vitro study can be correlated with the clinical situation due to its design considerations. In areas of high-stress concentration, the probability of fracture of the framework decreases as the thickness increases and acts as a barrier to interrupt crack propagation. Also, the incorporation of the lingual collar in the gingival embrasure turned the sharp radius of gingival curvature to round, which further increased its ability to withstand higher occlusal load. Therefore, the lingual collar design used in this study is ideally suited to strengthen the framework in compromised situation.

Failures in FPD are considered when the veneered ceramic fractures. The limitation of this study was the use of static model using cutback coping without ceramic layering as standardization of ceramic layering was not possible. Further research is required to check the flexural strength of the same after ceramic layering in dynamic conditions to verify the present findings.

CONCLUSION
The following conclusions were drawn based on the observation and results of this study:
1. The flexural strength of the Y-TZP-based three-unit posterior framework is affected by design.
2. The lingual collar-modified framework withstood higher occlusal load compared to the framework without collar.
3. Fracture occurred at the mesial connector between the pontic and the retainer in Group A and at the distal connector in Group B.

4. The displacement of framework in Group A (0.004 units) was lesser as compared to Group B (0.006 units) when loaded axially.

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Conflicts of interest
There are no conflicts of interest.

REFERENCES
1. Hamza TA, Attia MA, El-Hossary MM, Mosleh IE, Shokry TE, Wee AG. Flexural strength of small connector designs of zirconia-based partial fixed dental prostheses. J Prosth Dent 2016;115:224-9.
2. Partiyan A, Osman E, Rayyan MM, Aboushelib M, Ibrahim A, Jimbo R. Fracture resistance of three-unit zirconia fixed partial denture with modified framework. Odontology 2017;105:62-7.
3. Wimmer T, Ender A, Roos M, Stawarczyk B. Fracture load of milled polymeric fixed dental prostheses as a function of connector cross-sectional areas. J Prosth Dent 2013;110:288-95.
4. 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.
5. Schmitter M, Mussotter K, Rammelsberg P, Stober T, Ohlmann B, Gabbert O. Clinical performance of extended zirconia frameworks for fixed dental prostheses: two-year results. J Oral Rehabil 2009;36:610-5.
6. 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.
7. Kampsosira P, Papavasiliou G, Bayne SC, Felton DA. Stress concentration in all-ceramic posterior fixed partial dentures. Quintessence Int 1996;27:701-6.
8. Bahat Z, Mahmood DJ, Vult von Steyern P. Fracture strength of three-unit fixed partial denture cores (Y-TZP) with different connector dimension and design. J Prosth Dent 2009;103:149-59.
9. Fardin VP, de Paula VG, Bonfante EA, Coelho PG, Bonfante G. Lifetime prediction of zirconia and metal ceramic crowns loaded on marginal ridges. J Prosth Dent 2016;32:1543-54.
10. Raghavan RN. Ceramics in dentistry. Dental surgeon. Chennai, India; 2012.
11. Yilmaz H, Aydin C, Gul BE. Flexural strength and fracture toughness of dental core ceramics. J Prosth Dent 2007;98:120-8.
12. Plengsombut K, Brewer JD, Monaco EA Jr, Davis EL. Effect of two connector designs on the fracture resistance of all-ceramic core materials for fixed dental prostheses. J Prosth Dent 2009;101:166-73.
13. Taskonak B, Yan J, Mecholsky JJ Jr, Sertgöz A, Koçak A. Fractographic analyses of zirconia-based fixed partial dentures. J Prosth Dent 2008;109:1077-82.
14. Heinze SD, Cavalleri A, Zehlheger B, Büchler A, Zappini G. Fracture frequency of all-ceramic crowns during dynamic loading in a chewing simulator using different loading and luting protocols. J Prosth Dent 2008;109:1352-61.
15. 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.
16. 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.

17. Guess PC, Bonfante EA, Silva NR, Coelho PG, Thompson VP. Effect of core design and veneering technique on damage and reliability of Y-TZP-supported crowns. Dent Mater 2013;29:307-16.