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Effect of framework design on fracture resistance of zirconium oxide posterior fixed partial dentures

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ABSTRACT

Introduction: The effect of framework design modifications in all-ceramic systems is not fully understood. The aim of this investigation was to evaluate the effect of different framework designs on fracture resistance of zirconium oxide posterior fixed partial dentures (FPD).

Materials and Methods: Thirty two posterior zirconia FPD cores were manufactured to replace a second premolar. The specimens were divided into four groups; I: 3×3 connector and standard design, II: 3×3 connector and modified design, III: 4×4 connector dimension, and standard design and IV: 4×4 connector dimension and modified design. After storing for one week in artificial saliva and thermocycling (2000 cycles, 5-55°C), the specimens were loaded in a universal testing machine at a constant cross-head speed of 0.5 mm/min until failure occurred. The Weibull, Kruskal-Wallis, and Mann-Whitney tests were used for statistical analysis ($\alpha = 0.05$).

Results: The mean fracture resistance of groups with 4×4 mm connector was significantly higher than groups with 3×3 mm connector ($P < 0.001$). Although, the fracture resistance of the modified frameworks was increased in the present study (1.1 times), they were not significantly different from anatomic specimens ($P = 0.327$).

Conclusions: The fracture resistance of the zirconia posterior-fixed partial dentures was significantly affected by the connector size; it was not affected by the framework modification.

Key Words: Ceramic, dental materials, dental porcelains, dental prosthesis design, fixed bridges, zirconia

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INTRODUCTION

The use of all-ceramic restorations has become more popular in clinical dentistry due to the universal necessity to use a biocompatible and high esthetic prosthetic material.[1-5] However, the most dental ceramics are brittle and susceptible to breakage.[6,7] As an alternative for these restorations, the zirconia-based materials have been introduced as a strong core for esthetic fixed prostheses with good esthetics, excellent biocompatibility, and high mechanical properties.[8-10] Fracture pattern of all-ceramic crowns and fixed partial dentures (FPD) are nearly different.[11] Crowns fail from the cementation surface that is opposed to the occlusal surface, whereas fracture of all-ceramic FPDs has a relatively high incidence in the connector areas.[6,7,12,13] This high incidence in the posterior region is due to lower inter-occlusal space and short clinical crowns of molar abutments.[4,14] It has been said that with smaller cross-sectional area of connectors, the needed load for fracture of core will be decreased.[4] Another procedure to minimize the connector fracture in all-ceramic restoration is to make broadly curved connector that has been proved to be relatively safer than sharply curved connector.[6,15] However, the shape and size of zirconia connectors is still a controversial subject.[4,6,12,15] Another justification for importance of the connector size in the zirconia FPDs is this

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fact that the connectors of zirconia frameworks are usually bulkier than the veneering porcelain and may produce residual stresses due to differential heating and cooling rates.[16] These stresses can explain the high level of porcelain veneering delamination from the zirconia sub-structures.[17]

Many design modifications have been suggested previously for improving the strength of metal ceramic restorations to increase the support of veneering porcelain.[6,15,18] *In vitro* studies have shown that the alterations in metal ceramic restoration design such as increasing the radius of curvature at the gingival embrasure of the connector or adding a high lingual shoulder connected to a proximal strut can affect their resistance to fracture.[14,18,19] Adding lingual shoulder and proximal strut was used for improving the mechanical properties of all-ceramic single crowns,[6,14] although this design did not improve the mechanical properties of the all-ceramic systems used in the study by Lorenzoni *et al*.[2] However, the effect of using this design in all-ceramic FPDs is not completely investigated.[7] Most of the available studies about the zirconia framework design are about its effects on the marginal adaptation.[20-22]

The aim of this investigation was to evaluate the effects of framework design on fracture resistance of zirconium oxide posterior FPDs.

The null hypothesis was that the framework design did not have any effect on the fracture resistance of zirconium oxide posterior FPDs.

MATERIALS AND METHODS

In this experimental-laboratory study, a maxillary typodont model was used (AG-3; Frasaco Franz Sachs and Co., Tettnang, Germany) for making a nickel–chromium (Ni–Cr Formula 45+Ti, Neodontics, USA) master die. The right first premolar and first molar were prepared so that the finish line was a circular shoulder (1.2 mm depth) with rounded internal axio-gingival line angles and comprised 2 mm of occlusal reduction. The preparation was assessed by a dental surveyor (Ney Co, Hartford. CT, USA) to ensure a bucco-lingual and mesio-distal occlusal convergence angle of 8°. The occluso-gingival preparation height was 5 mm. The residual ridge in the pontic area (second premolar) was reduced 2 mm in height to create adequate occluso-gingival clearance for next study steps. Impression of this model was made using a polyvinyl siloxane impression material (Panasil, Kettenbach, Eschenburg, Germany) and was poured with a type IV dental stone (GC Fujirock EP, GC Dental Products Co., Tokyo, Japan) to make a working cast. The dies were scanned (Cercon Eye, DeguDent, Hanau, Germany) and four different framework designs were prepared using computerized software (Cercon art 2.2, Degudent, Germany) as follows [Figure 1]:

Group I: The copings with 3 × 3 connector dimension and standard design
Group II: The copings with 3 × 3 connector dimension and modified design
Group III: The copings with 4 × 4 connector dimension and standard design
Group IV: The copings with 4 × 4 connector dimension and modified design.

Standard design was an uniform core with an even thickness of 0.7 mm. Modified design consisted of a standard 0.7 mm thickness core increased in thickness by adding a 1 mm thick lingual margin of 2.0 mm height, connecting to full contour proximal struts with 3.5 mm height.[2,14] In all designs a depth and width of 0.2 mm × 3 mm indentation was prepared in the center of pontic area to use it in force application step.

Ceramic blocks (Cercon Base ceramic, Degudent, Germany) were milled using a milling unit (Cercon Brain unit, DeguDent, Germany) according to aforementioned designs. All the specimens were sintered to full density in the Cercon Heat furnace (a 6-h sintering program with a maximum temperature of 1350°C). At last, 32 zirconia FPD frameworks were prepared according to four different designs (n = 8). The fabricated frameworks were checked to see their fit at the margins using a silicone material (Fit Checker, GC Co, Tokyo, Japan). Adaptation was evaluated by one person under magnification × 10 (MBC-10, St. Petersburg, Russia).

After storing for one week in artificial saliva at 37°C, the specimens were subjected to thermocycling for 2000 cycles at temperatures alternating between 5 and 55°C for 30 s each, with an intermediate pause of 15 s. Before loading of specimens, a Teflon cylinder (3 mm diameter and 3 mm height) was placed in the center of each pontic by the means of cyanoacrylate adhesive. This cylinder was placed to prevent any contact damage during the loading and to provide homogeneous load distribution on the pontic unit.[13]

The specimens were loaded in an universal testing machine (Electromechanical low-capacity testing Machines, walter + bai, AG, Switzerland) at a constant crosshead speed of 0.5 mm/min until failure occurred. The maximum force to fracture was recorded in Newton.

The collected data were analyzed (SPSS/PC 13.0; SPSS Inc, Chicago, IL, USA) using Weibull, Kruskal−Wallis, and Mann−Whitney tests with the level of significance at 0.05. The Weibull parameters, characteristic force at failure (F_0), and the Weibull modulus (m) were determined for each test group by fitting a Weibull distribution to each respective dataset. F_0 is load bearing capacity for the specimens at 63.2% probability of failure, whereas the modulus m is an indication of the scattering in the force at failure and of the reliability of the material investigated.[9,13,23]

RESULTS

The mean and standard deviations of the load to fracture of the zirconium oxide posterior FPD specimens are shown in Table 1. Because the groups did not meet the assumption of homogeneity of variances, the effect of framework design on fracture resistance of zirconium oxide posterior FPDs was evaluated by Kruskal−Wallis test. This statistical test revealed that there was a significant difference among all experimental groups ($P < 0.001$) [Table 1]. The Mann−Whitney test showed that the mean fracture resistance of group IV was significantly higher than group I ($P < 0.001$) and group II ($P < 0.001$), but there was not any significant difference between group IV and group III ($P = 0.156$). The mean value of the load to fracture of the zirconium oxide posterior FPD specimens was affected significantly by the connector size ($P < 0.001$), but it was not affected by the framework modification ($P = 0.327$).

The average load bearing capacity F_0 and the reliability m are listed in Table 1. The highest load bearing capacity was seen in the group IV frameworks and the lowest is seen in the group I frameworks. The group II frameworks were the most reliable, showing the smallest scattering of the measured load-bearing capacities [Figure 2] Other group frameworks suffered from a broader load-bearing scatter which was showed by the lower Weibull moduli.

DISCUSSION

In this *in vitro* study, the fracture resistance of three-unit zirconium oxide posterior FPD was investigated. There was a significant difference among all experimental groups ($P < 0.001$), so the null hypothesis that the framework design did not

Table 1: The mean values (SD) of the loads to fracture (N) the zirconia bridges, and the Weibull parameters F_0 (characteristic force at failure) and m (the Weibull modulus)

Framework design	Mean(SD)*	Min	Max	F_0 (N)	M
Group I (3×3)	2127.50 (490.03) A	1170	2700	2350.74	4.06
Group II (3×3 modified)	2366.35 (214.07) A	2080	2700	2462.76	12.04
Group III (4×4)	3658.75 (942.43) B	2420	4800	4043.97	3.99
Group IV (4×4 modified)	4371.25 (928.84) B	2800	5500	4759.93	4.90

*The groups with similar letters did not have any significant difference (Mann-Whitney test, $p<0.05$)

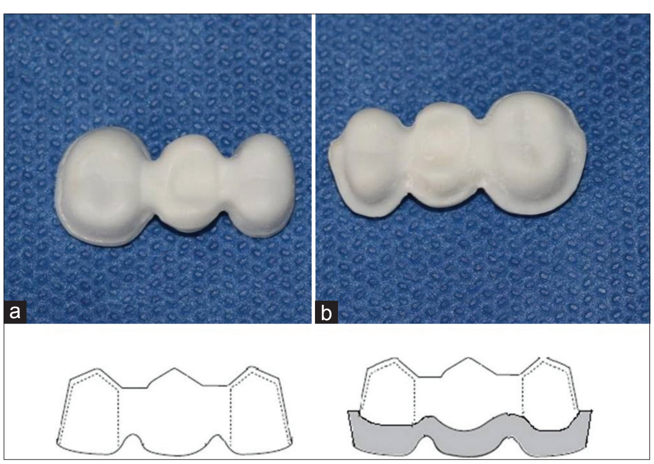

Figure 1: Specimens with anatomical (a) and modified design (b) a.Occlusal view, b.Lingual schematic view

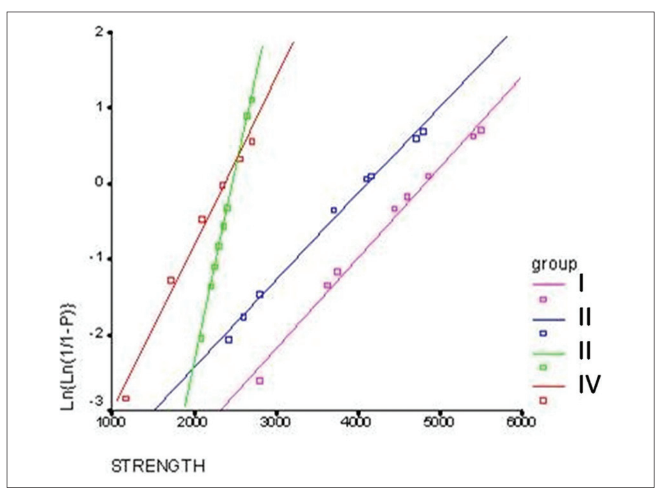

Figure 2: Weibull diagram

have any effect on the fracture resistance of zirconium oxide posterior FPDs was rejected.

The effect of framework design modifications in all-ceramic systems is not fully understood. Many of the present data about the framework design is about metal ceramic frameworks wherein core fracture is not an important issue.[14,24] Using proposed core-design modifications in metal ceramic restorations may improve the framework strength in all-ceramic restorations.[14,18]

Some authors stated that coping design can affect the fracture load.[14,18] Marchack *et al*.[24] showed that extending the core 2 mm above crown margins can decrease the veneer failure rates in zirconia-supported crowns. But Lorenzoni *et al*.[2] stated that framework design modification did not improve the fatigue life of the veneered yttria-stabilized tetragonal zirconia polycrystals (Y-TZP) crowns.

In the present study, the mean value of the load to fracture of the specimens was affected significantly by the connector size ($P < 0.001$), but it was not affected by the framework modification ($P = 0.327$) [Table 1]. However, the fracture resistance of the modified frameworks increased in the present study (1.1 times), but it was not significantly different from anatomic specimens. It can be due to small sample size in this study.

The connector is the thinnest section of a framework and will bend more easily than the pontic and abutment areas. To ensure optimum strength, the connectors must be of adequate dimensions.[4,12,13,25-27] Connector dimension was evaluated for each specific all-ceramic system. The 3 × 3 mm[28] and 4 × 4 mm[25,29,30] dimensions were proposed for the most all-ceramic systems. But Augereau *et al*.[26] stated that increasing the dimensions of the connector is not necessary to increase resistance to fracture. On the other hand, dental technicians tend to make connectors with sharp gingival embrasures to have more esthetic FPDs.[27] Theses sharp line angles can increase the failure rate of all-ceramic restorations in the connector areas.[31] Some authors showed that increasing the radius of the gingival embrasures of the connectors increased the fracture strength of Y-TZPFPD.[6,15,32]

It is believed that design modifications can be useful in such cases because having ideal connector dimension is impossible due to hygienic and periodontal conditions, high functional forces, long span FPDs, deep over bite, and parafunctional habits.[25,33] However, in the present study, the Kruskal–Wallis test showed that there was not any statistical difference between the groups with same connector size but different core designs [Table 1]. It is said that in such cases, materials other than all-ceramics should be considered.[25]

Comparing the load-bearing capacities F_0 for four study groups [Table 1], it becomes obvious that the groups with 4 × 4 mm connector possess an approximately 1.5 to 2 times higher load-bearing capacity than groups with 3 × 3 mm connector dimension. It is well known that the load-bearing capacity for FPDs depends not only on the properties of the ceramic material, but also to a high extent on the size, shape and position of the connectors, as well as on the span of the pontics.[13,34] The increased load-bearing capacity found in the groups III and IV was due to the larger size of the connectors of the test specimens. The slightly higher F_0 for group IV when compared with group III can be considered negligible. By enlarging the cross-section at the point of failure (at the connectors) the maximum stress at the connector was reduced and thus for a given load the failure probability was also reduced. Load-bearing capacities F_0 for all groups in the present study was higher than some other studies.[4,13] However, the values calculated in the study by Kokubo *et al*.[18] were higher than present study. It can be due to difference in connector size, length of span, periodontal ligaments (PDL) simulation around study abutments, and high elastic modulus of abutment teeth.

Weibull moduli ranges between 6.1-8[23] and 4.5-5.7[9] for zirconia-based FPDs and about 7 for zirconia frameworks[13] were reported earlier, whereas the present study exhibited Weibull moduli of 4.06, 12.04 for groups I and II, and 3.99, 4.90 for groups III and IV respectively [Figure 1]. The results of this study do not concur with the literature, where an increase in the Weibull modulus was reported by increasing connector dimension.[12,13,25,35] The Weibull modulus is related to the distribution of flaws in a brittle material and it is not related to the size of the flaws. If the flaws are evenly distributed throughout the specimen, the resulting data will show little statistical scatter and result in a high value of the Weibull modulus.[36] In this study, the highest modulus (m) was seen in the group II frameworks. Hence, it can be concluded that the specimens in this group have more uniform flaw distribution than other groups.

The abutment teeth were made of nickel–chromium alloy which had an elastic modulus higher than dentin. Using supporting material with high elastic modulus resulted in increasing fracture strength.[9,11,25] Therefore, fracture forces evaluated in the present study might have been higher than in clinical practice. However, studies using resin material for the abutment teeth reported similar fracture forces for zirconia-based FDP.[23]

On the other hand, it is said that analysis using immobile retainers produces a much higher failure load than found when testing with mobile posts.[9] However, increasing the complexity of the model and simulating the periodontal membrane, did not increase the validity of the results.[25] Beside, physical properties of PDL substitute materials were approximate and were not considered as real values.[37]

Another aspect to consider is the shape of the specimens to be tested. Ideally, the FPD should be anatomically shaped with dimensions similar to those of FPDs in clinical use.[25] In the present study the anatomically shaped and size specimens were used.

The fracture strength of the zirconia specimens can be increased considerably after veneering.[38] However, in some studies, the fracture resistance of zirconia-based FPDs was decreased after veneering.[39] However, the fracture resistance values of veneered specimens can be due to veneer material fracture instead of core fracture. So, in the present study unveneered cores were used.

To simulate aging, three main factors have to be present [9,23] which are as follows:

- Presence of water which propagates crack growth in ceramics.
- Repeated thermal stressing which leads to a decrease in the mechanical properties of prosthetic restorations.
- The repeated application of mechanical force for simulating the chewing strokes.

Therefore the specimens were stored in artificial saliva for one week and then were subjected to thermocycling for 2000 cycles (5-55°C) according to ISO TR No. 11450.[9,23] But one limitation of the present study was using static instead of dynamic loading.

CONCLUSIONS

Within the limitations of this *in vitro* study, one can conclude that:

1. The mean fracture resistance of groups with 4×4 mm connector was significantly higher than groups with 3×3 mm connector.
2. Although the fracture resistance of the modified frameworks increased in the present study (1.1 times) it was not significantly different from anatomic specimens.

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