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Effect of Connector Designs on Flexural Strength of Gradient Versus Translucent Monolithic Zirconia Fixed Dental Prosthesis: An in Vitro Study

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ABSTRACT

Background: Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) has been used due to its high strength and unique transformation toughening properties. Despite the exceptional mechanical properties of using zirconia, its opacity limits its use only in posterior regions. Improvements in translucency were done in monolithic zirconia "gradient zirconia" restorations since several clinical data reported a high failure rate of (FDP) around connector areas. ***Aim of the study:*** to evaluate the effect of connector designs on the flexural strength of gradient zirconia in comparison to translucent zirconia fixed dental prosthesis. ***Materials and Methods:*** A total of twenty-three-unit fixed dental prosthesis were fabricated for the current study. The FDP will be divided into two main experimental groups (n =10). With 4 subgroups as the following: Group B (BruxZir zirconia) with subgroups: SubGroup BS: BruxZir Zirconia with sharp connector design. SubGroup BR: BruxZir Zirconia with round connector design. Group G (Gradient zirconia) with subgroups: SubGroup GS: Gradient zirconia with sharp connector design. SubGroup GR: Gradient zirconia with round connector design. ***Results:*** Revealed that the lowest flexural strength mean value was related to group B BruxZir with sharp connector design with a value (578.77), followed by Group B BruxZir with round connector design (709.10), while the highest flexural strength value was found in ZirCAD group with round connector designs (964.78). ***Conclusions:*** Flexural strength of gradient 3Y & 5Y zirconia is higher than translucent 4Y zirconia. Round connector designs show higher flexural strength than sharp designs.

Keywords: Connector Design, Flexural Strength, Gradient Zirconia

INTRODUCTION

Yttria-stabilized tetragonal zirconia has been used for ceramic fixed dental prostheses (FDPs) due to its high strength, acceptable esthetics, and biocompatibility. It has been performed as the material of choice in high-stress regions and is now approved as a high-strength material with its unique transformation toughening properties.[1,2]

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Several studies have settled the use of zirconia for long-span FDPs in high-stress clinical situations as a core material.[3] It has a superior aesthetic appearance compared to metal-ceramic restorations; however, layering with more translucent materials is necessary for improving the restoration's appearance regarding its opacity.[4] Superficial Chip-off fractures of the veneer porcelain are considered an inherent weakness in such veneered Y-TZP restorations as it has been reported as a serious common issue.[1,5]

Translucent zirconia has been used in different clinical situations as another alternative, providing simpler clinical steps than the construction of multilayer restoration with opaque zirconia cores veneered with translucent feldspathic.[6] Ultra-translucent zirconia is drawing attention due to its exceptional esthetic appearance. However, the high translucency came at the expense of its mechanical properties, which can be attributed to the reduced ability of transformation toughening due to the increased cubic zirconia content.[7] The quest to reach an ideal representative material with dual advantages has led to the introduction of gradient zirconia material, which benefits the strength of 3-mol% yttria stabilized tetragonal zirconia polycrystal (3Y-TZP) and provides the good esthetics of 5-mol% yttria-stabilized tetragonal zirconia polycrystal (5Y-TZP).[7,8]

Since several clinical data have reported a relatively high failure rate of FDP consistently around connector areas between retainers and pontics, different modifications have been introduced to eliminate this problem.[8] One of these modifications is altering the stress pattern of the connector design in regions where maximum stress occurs, in which to improve the flexural strength of three-unit FDPs.[9]

However, the scientific data for this material is very limited to support their indications. Therefore, the aim of this in vitro study is to evaluate the effect of connector designs on the flexural strength of gradient zirconia in comparison to translucent zirconia Fixed dental prosthesis. The hypothesis of the study is that there will be a difference in flexural strength under load between different connector designs of gradient zirconia and translucent zirconia in three-unit fixed dental prosthesis.

MATERIALS AND METHODS

Materials

1- Gradient zirconia: IPS Emax ZirCAD disc 16 mm thickness.

2- Translucent zirconia: BruxZir Shaded 16 PLUS disc 14 mm thickness.

Sample size calculation was performed

using IBM® SPSS® SamplePower® Release 3.0.1 based on the results of Onodera K et al (2011). [10] A total of twenty-three-unit Fixed dental prosthesis were fabricated for the current study. The specimens will be divided into two main experimental groups (n =10). With 4 subgroups as the following: **Group B** (BruxZir zirconia) with subgroups: **SubGroup BS**: BruxZir Zirconia with sharp connector design. **SubGroup BR**: BruxZir Zirconia with round connector design**.** **Group G** (Gradient zirconia) with subgroups: **SubGroup GS**: Gradient zirconia with sharp connector design. **SubGroup GR**: Gradient zirconia with round connector design. Two standardized stainless-steel master dies were designed to simulate a mandibular left first premolar and first molar prepared to receive zirconia frameworks. The stainless-steel dies and the base were fabricated by computer numerical control (CNC) machining. The stainless-steel master dies were milled with a 1.0- mm-deep chamfer finish line width, 4.5-mm occlusal-gingival height, and a total convergence angle of 12 degrees.[11] To simulate a 3-unit (FDP) from a mandibular first premolar to a first molar, the stainless-steel dies were screwed in pairs in a custom-designed stainless-steel holder with 17.5 mm between the centers of the master dies. Then, it was coated with Shera scan spray, a very thin anti-reflective white in color scan powder, in order to obtain a 3D image and transmit it to the designing software using the Medit i700 scanner. The generated standard tessellation language (STL) file was checked for inaccuracies and then transported on a flash drive to the computer for the software design.

Designing the fixed dental prosthesis was done via Exocad Dental CAD2.2 software. An occlusal thickness of 2 mm from the center of the abutments was set to standardize the production of the BruxZir and ZirCAD fixed dental prostheses. The connector size was chosen according to the manufacturer’s instructions, with 3x3mm for the molar abutment and the premolar abutment. A standardized wall thickness of 1 mm and 100 µm cement space was provided .[11] Bigeneric copy mode was used for the construction of the fixed dental prosthesis to ensure standardization of the restorations. Then, round and sharp connector designs were set using the software. Zirconia fixed dental protheses were milled using CAD/CAM milling machine Ronald DWX-52D 5-Axis dental machine. zirconia was sintered at temperatures between 1350 and 1550 °C with holding durations of 2 to 4 hours. This was performed by using a Ceramic furnace in fire HTC speed. Polishing for zirconia was done first by eZr™ complete system kit, then glazing of zirconia restorations was done using Multimat® Easy, Dentsply furnace. The glaze paste was applied using a ceramic brush in a thin coat over the whole surface of each zirconia FDP. Typically, glazing is done for 1-2 minutes at a temperature of between 850 and 900 C. The prostheses are then bench-cooled to room temperature after being chilled to about 600 °C.[12]

Restorations were not cemented on the stainless-steel dies to exclude the effect of cement on the flexural strength. Both groups FDP were jigged and put on a universal testing device. One layer of 1-mm-thick tin foil was sandwiched between the ceramic specimen and the opposing steel ball in order to create a more even distribution of stress between the two surfaces; then, a vertical load was applied through a 3 mm diameter steel bar at a crosshead speed of 0.5 mm/min occlusal on the mid-way pontic area.[11] The flexural strength of the fixed dental prostheses was maintained. The highest load that caused a bridge to collapse was measured in MPa. Flexural strength values were calculated from the following formula[13]:

$$Flexural\ Strength = \frac{3xLoadxLength}{2xWidthxThickness^2}$$

$$\sigma \frac{3PL}{2bd^2}$$

Where:

F= Loading force at fracture point

L= Length of supporting span “33 mm”

b= Width “8 mm”

d= depth of FDP specimen “3mm”

Statistical analysis

Statistical analysis was performed using SPSS 20®, Graph Pad Prism® and Microsoft Excel 2016. All quantitative data were explored for normality by using Shapiro Wilk and Kolmogorov Normality test and presented as minimum, maximum, means, and standard deviation (SD) values. All data were presented in tables and graphs. There was an insignificant interaction between the connector design and the material, as P=0.11 had 2.29% total variance.

RESULTS

There were statistically significant differences between the tested groups. Comparison between different groups was performed using the Two Way ANOVA test as listed in **Table 1,** which revealed a highly significant difference between them as P<0.0001. Tukey`s Post Hoc test was performed for multiple comparisons as listed in **Table 2** and revealed that the lowest flexural strength mean value was related to group B BruxZir with sharp connector design with value (578.77), followed by Group B BruxZir with round connector design (709.10).

Table 1: Comparison between flexural strength of all groups using Two Way ANOVA test followed by Tukey`s Post Hoc test for multiple comparisons.

Flexural strength						
Material	**Connector design**	**Min**	**Max**	**Mean**	**SD**	**P value**
Group B (Bruxzir)	**BS**	541.40	603.40	578.77 a	32.90	<0.0001*
	BR	661.30	804.10	709.10 b	82.27	
Group G (Zircad)	**GS**	695.70	820.60	744.07 b	67.04	
	GR	906.80	1002.65	964.78 c	50.99	

Min: *minimum* ***Max:*** *maximum* ***SD:*** *standard deviation*
**significance at P ≤ 0.05.*

Table 2: Tukey's multiple comparisons test of flexural strength**.**

Tukey's multiple comparisons test						
	Mean 1	Mean 2	Mean Diff.	SE of diff.	95.00% CI of diff.	P Value
Group I BS vs. Group I BR	578.8	709.1	-130.3	49.91	-290.2 to 29.50	0.1151
Group I BS vs. Group II GS	578.8	744.1	-165.3	49.91	-325.1 to -5.467	0.0429*
Group I BS vs. Group II GR	578.8	964.8	-386	49.91	-545.8 to -226.2	0.0003*
Group I BR vs. Group II GS	709.1	744.1	-34.97	49.91	-194.8 to 124.9	0.8941
Group I BR vs. Group II GR	709.1	964.8	-255.7	49.91	-415.5 to -95.85	0.004*
Group II GS vs. Group II GR	744.1	964.8	-220.7	49.91	-380.5 to -60.88	0.0096*

**significance at P ≤ 0.05.*

In contrast, the highest flexural strength value was found in ZirCAD group with round connector designs (964.78) followed by ZirCAD with sharp connector designs (744.07). Effect of ZirCAD and BruxZir material, there was a significant difference between different materials as P<0.0001, with 49.72 % total variance.

DISCUSSION

Due to its good biocompatibility and high aesthetic potential, ceramic fixed dental prostheses FDPs have seen a surge in use over the past 10 years. However, they have poor flexural strength, especially when utilized in the posterior region. High-strength oxide ceramics, like zirconium dioxide, have been developed in an effort to increase flexural strength; these materials have mechanical properties that are superior to those of other all-ceramic materials currently on the market. The development of these materials, combined with modern CAD/CAM technology, has led to a wide range of applications for dental restorations.[14] The weakest zone of all-ceramic (FDPs) and the location of the majority of clinical failures is the connector area.[15] By changing the connector's design and size in the zones of greatest tension, 3-unit FDP can achieve greater long-term succ-ess.

It is challenging to standardize the connectors size, making it difficult to test the biomechanical performance of the connectors in relation to their dimensions in clinical studies.[16] As a result, the 3-unit FDP design was standardized in this study with a connector diameter of 3x3 mm in accordance with the manufacturer's instructions. Furthermore, two connector designs round and sharp were developed. The same design was used by Plengsombut et al.[14] in a prior in-vitro study; however, restorations are subjected to cyclic loading in the oral cavity that changes in magnitude and direction as a result of a complex chemical environment, and the loci of stress concentration for FDP used in flexural strength tests used in the dental laboratory are different from those in FDP in the oral cavity.[17]

The abutments teeth in the current study were constructed of stainless steel, which has an elastic modulus of 200 GPa, which is greater than that of dentin, which has an elastic modulus of 12 GPa.[18] Their benefit is that they can all have the same physical characteristics and measurements, like a taper and finish line. In several studies, fracture testing of ceramic crowns and FDPs has been done using steel or resin dies.[19] However, steel or resin abutments do not replicate the actual force distribution that happens on crowns cemented to natural teeth. It can be argued that a standard steel or resin die, ensures constant preparation shape and equal physical quality of the abutments under loading.

Although most papers choose specimens to test the flexural strength of connector design, in this study twenty 3-unit fixed dental prosthesis were constructed according to shakal et al.[13] and Bahraini et al.[20] Bigeneric copy mode was used for the construction of the fixed dental prosthesis to ensure standardization of the restorations. FDP were not cemented on their corresponding dies to eliminate the cement variable.[21] The FDP were then placed into a universal testing device. Steel balls, were used to apply the load perpendicularly to the middle portion of the FDP pontic at a cross head speed of 0.5 mm/min until failure occurred. One layer of 1-mm-thick tin foil was sandwiched between the ceramic specimen and the opposing steel ball in order to create a more even distribution of stress between the two surfaces. When the force was 1% below the peak level recorded during the test prior to fracture detection and testing interruption, the loads necessary to fracture the FDP were automatically recorded. Visual examination of fracture patterns was performed on representative fractured specimens from each test group.

The hypothesis of this study was accepted that there will be difference in flexural strength under load between different connector designs of gradient zirconia (3Y&5Y) and translucent zirconia (4Y) in three-unit fixed dental prosthesis.

In the current study, the mean and standard deviation (SD) values of flexural strength for each group were calculated. The ZirCAD round connector group had the greatest values, with a mean of 964.78, and the BruxZir sharp group had the lowest values, with a mean of 578.77. These findings are consistent with Hamza et al.[22] which state that the flexural strength of the round connector design is higher than that of the sharp design. Also, Y-TZP based ceramics is affected by the connector dimension and design.

In the current investigation on connector shape, all restorations with sharp connector designs had fracture patterns that were less angularly inclined toward the pontic than those with circular connector designs. The difference in stress levels between connector designs with rounded and sharp edges may help to explain this finding.[14,22] The acquired results are consistent with earlier preliminary investigations, which found that connections with a round design are less angular and exhibit lower levels of stress.[23,24] According to Tsumita et al.[25], the design of the framework in ceramic FDP, particularly the pontic connector interface, significantly influences how stress is distributed within the framework. Oh and Anusavice and Dornhofer et al.[26] have said that the radius of curvature at the gingival embrasure significantly influences the flexural strength of FDPs, hence the radius of the gingival embrasure should be as large as feasible.

Oh and Anusavice and Sundh et al.[27] also found results that were similar to those of this study; as a result, the connector's crosssection diameter, shape, and location are crucial for zirconia FDP to succeed over the long term and should be chosen by the material's properties, anatomical constraints, and aesthetic preferences. Varying processing factors that cause varying levels and types of surface damage can have a direct impact on zirconia's strength.[28] Surface flaws on the core can be caused by industrial fabrication, sintering processes, cooling, various surface treatments, and modest clinical changes, which compromise the reliability and long-term performance of these materials.

Even though they are tiny in nature, areas of surface defects created during milling can operate as stress concertation sites and as prospective sites for crack initiation and propagation.[29] In order to create the stresses necessary for the transformation, which maintains the surface under good compressive stress and greatly boosts flexural strength, grinding and airborne-particle abrasion have been proposed.[30]

The large discrepancy in maximum failure load between various connector designs may be explained in the following manner, according to the findings of the current study. First, earlier investigations using finite element analysis have revealed higher stress levels for the sharp connector design.[27] Second, the milling process has a detrimental effect on the structural dependability and strength of blocks made of industrially prepared ceramic; nonetheless, the potential for defects and flaws to be introduced into the ceramic increases with the sharpness of the milling bur configuration used to generate the acute notch of the connector.[14]

To more effectively generalize the results of this study, these characteristics should be taken into consideration in further clinical studies. Future research should incorporate thermo-mechanical loading and model the oral environment more accurately. The absence of periodontal ligaments, which would have been able to absorb masticatory stresses intraorally, is a drawback of the current study, which used metal abutments. However, the biomechanics of the natural periodontal ligament are extremely intricate to imitate due to its complicated structure (collagen fibers, blood vessels, nerves, and fluids).[14,31] Nevertheless, it has been claimed that replicating the periodontal membrane does not improve the data' validity.

The present study has limitations, including a number of issues that make direct comparisons between the findings and clinical settings challenging. Under clinical conditions, an anatomical restoration is subjected to cyclic loading that changes in amplitude and direction while being influenced by a complex chemical environment. This study used specimens in a single load to failure. Additionally, a zirconia restoration is supported by the tooth structure, as opposed to the test specimens, which were supported by the attachment unit's stiff fixation.

CONCLUSION

1- Flexural strength of gradient 3Y & 5Y zirconia is higher than translucent 4Y zirconia.

2- Round connector designs show higher flexural strength than sharp designs.

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CONFLICTS OF INTEREST

The authors claim that they have no conflicts of interest.

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