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Effect of the radius of curvature of embrasure in connector area on the fracture resistance of three-unit zirconia fixed partial dentures

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ABSTRACT

BACKGROUND: Most clinical studies indicate that fixed partial denture's (FPDs) have an approximately high failure rate, especially around the connector area between the retainers and pontics. The configuration of the prosthesis components can have a significant impact on the stress distribution as the contour changes. Increase the fracture resistance of three units of FPD by modifying the connector configuration in sections where the stress pattern occurs with extreme stress. **OBJECTIVES:** This study aimed to evaluate the effect of the curvature radius on the connector embrasure areas on the fracture resistance of three-unit all-ceramic FPDs. it was made from IPS e.max ZirCAD. *Methods:* A total of 20 all-ceramic three-unit FPDs were fabricated from IPS e.max ZirCAD. They were divided into four groups with different connector's designs. All FPDs were adhesively bonded to the abutment teeth using resin cement. **RESULTS:** The mean fracture loads were highest for Group A (0.60 mm r OE and r GE). The results showed that a significant difference among fracture loads of different types of connector design. **CONCLUSIONS:** It can be concluded that change of the radius of curvature of the connector area showed to be critical for the fracture resistance of all-ceramic fixed partial dentures. The fracture resistance can be improved by constructing larger radii of gingival and occlusal embrasure in the connector area.

Keywords: fracture resistance, Zirconia, gingival, occlusal embrasure radii

INTRODUCTION

The weakened part of posterior all-ceramic fixed partial dentures (FPDs) is the connector area mentioned by many investigators. The occlusal loads are more significant in the posterior area, and the connector's elevation was restricted by the short crowns of the molars (Steyern et al., 2001 and Sorensen et al., 1999). In the three-unit FPDs, the connector areas perform stress concentrations because the construction of an FPDs compound has many concavities and convexities, relying on the teeth restored and their alignment. In addition, dental technicians prefer to make sharp embrasure types to boost esthetics at the connector site (Kokich et al., 1999). These factors can increase stress levels, increasing the likelihood of all-ceramic FPDs failing at connectors, frequently at locations with insufficient connector height. The connector area should be planned to help with stress management; an alteration in the components can affect the stress distribution in the ceramic body. The finding is a sudden contour change (Kamposiora et al., 2000). The magnitude of local changes can significantly increase stresses from abrupt stresses. Brittle materials such as porcelains can exacerbate stress with cracks of varying sizes and locations (Thompson et al., 1994).

Many researchers use numerous ways to measure the fracture strength of all-ceramic three-unit posterior FPDs. Uniaxial flexure tests (such as three- and four-point flexure tests), biaxial flexure tests, and steel ball loading were performed at a crosshead speed of 0.5 mm/min by a 14.3 mm diameter steel ball. Still, any single one of these means imitates the typical clinical conditions (Won et al., 2002; Tinschert et al., 2001;Tinschert et al., 2000).

The measurement of ceramics' since ceramic stress levels can vary by mode of size, the acquisition of relevant clinical data will be complicated (Burke 1999 and Malament et al., 1999). Whereas in bar and disk experiments, fracture stress and fracture toughness values are plotted and assigned the geometric parameters (size and location), FPD stress concentrations cannot be directly measured or calculated as these geometric parameters. We may perform on dental stress raisers, accurately describing all FPD risk factors (Li 1999). The design of the stress relief can increase the fracture toughness of three segments. -previous research demonstrated more uniform stress distributions in more comprehensive geometries (Kamposiora et al., 2000; Pospiech et al., 1996). However, there hasn't been enough data collected in the connector area to provide suggestions for FPD construction.

This study was carried out to determine the magnitude of the curvature radius's effect at the connector's embrasure regions on the fracture resistance of three-unit all-ceramic FPDs made of IPS e.max ZirCAD.

MATERIAL AND METHODS

A model (Columbia Corp, USA) assumed the mandibular first molar as missing and the mandibular second premolar, the second molar was prepared for three units of FPD. The teeth were prepared for the complete crown with 6° occlusal convergence and a tooth reduction of 1.0-1.5 mm axially and 2.0 mm occlusal. The finish line was formulated deep chamfer using taper with round end diamond bur (Mani, Inc, Japan), standard grit diamond and fine grid diamond for smoothing. In addition, the line of the angles was rounded. The height of the premolar was 6 mm and the molar was 5 mm. A dental surveyor (Dental farm, Italy) was used for standardization.

Five all-ceramic three-unit FPDs were created for each of the four different connector designs from IPS e.max ZirCAD (Ivoclar Vivadent, Schaan, Liechtenstein). The occlusal embrasure radius (r OE) and the gingival embrasure radius (r GE) were created with two radii (r) of curvature: 0.60 mm and 0.25 mm.

Four distinct groups were distinguished as follows: Group A = r OE 0.60 mm and r GE 0.25 mm; Group B = r OE and r GE (0.60 and 0.25 mm); while Group C = r OE and r GE (0.25 and 0.60 mm); and Group D = r OE and r GE was 0.25 mm. A total of 20 IPS e.max ZirCAD FPDs were CAM fabricated. First, the IPS e.max ZirCAD FPDs were sintered in a sintering furnace (Programat S1 1600, Ivoclar Vivadent) at 1500 °C for two hours. After that, the FPD's were checked and glazed.

All FPDs were adhesively luted after the manufacturers to mimic clinical conditions. The inner surfaces of the crowns were sandblasted with 50 μm aluminium oxide Al_2O_3 at a pressure of 1 bar and conditioned with and Monobond Plus (Ivoclar Vivadent, Schaan, Liechtenstein) for 60 seconds and dried lightly with air. Multilink Automix (Ivoclar Vivadent) self-curing resin cement was used for the cementation of FPDs. Prior to the resin cement setting, a soft brush softly removed any extra cement near the margin. The FPDs were kept in place by finger pressure for the entire set time. The cemented prostheses were preserved for 24 hours in distilled water at 37°C before testing and subjected to thermal cycling (Haake, Karlsruhe, Germany). Two hundred cycles were performed between 5 and 55 ° C (30 s dwell time at each temperature).

All samples were loaded using a tungsten ball with a diameter of 5 mm at a cross heads peed of 0.5 mm per minute in a universal testing machine (Instron Corp, Canton, Mass) until fracture occurred. The tungsten ball was positioned in the middle of the pontic. A polyethene foil (1.0 mm thickness) was located among the ceramic specimen and the opposing tungsten ball to produce more identical stress dissemination amongst the tungsten ball and the ceramic surface. Loading was persisted to the stage of fracture, and the peak load was recorded at the fracture point. Crack location and fragmentations were located using visual examination. SPSS Software (version 21, Munich, Germany) were used for Fracture force data. Means and standard deviations (SD) were calculated for each group. A student t-test compared the results. A one-way table was prepared for analysis of variance (ANOVA) was applied. The data to disclose any significant differences at 5 % level.

RESULTS

The fracture load of all groups are shown in Table 1 and Fig 1. The mean fracture load and SD were as follows: Group A, 985 ± 94.8 N; Group B, 742 ± 81.3 N; Group C, 979 ± 96.1 N; and Group D, 691 ± 58.9 N. The mean fracture loads were highest for Group A where the lowest was for Group D. The ANOVA revealed a statistically significant difference (P=0.002) in the mean fracture loads of various connector designs. In addition, the loads to failure were significantly greater for designs A and C than for designs B and D. while in the failure loads of designs A and C or B and D were not significant differences.

Visual examination of the fractured samples revealed that all cracks occurred between both the second premolar and the first molar (10 samples) in addition between the first molar and the second molar (10 samples)

Table I. Fracture loads, Mean and Standard deviation values for the different connector designs

Group	No	Mean force of fracture (N)	Standard deviation (S.D.)	Location of fracture	
				Mesial	Distal
A	5	985	94.8	2	3
B	5	742	81.3	3	2
C	5	979	96.1	3	2
D	5	691	58.9	2	3
ANOVA P value		26.2 0.002*			

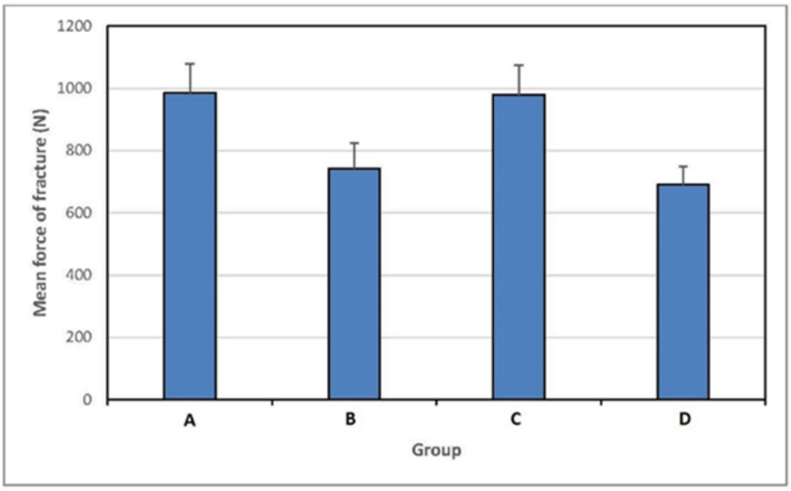

Figure 1: Fracture resistance of different connector designs

DISCUSSION

The shape of the framework, especially of the interface of the pontic connector, affects all-ceramic FPDs' stress distribution, fracture strength and fracture mode (Tsumita et al., 2008). In reducing fracture probability in test specimens, the morphology of the connector configuration at the gingival embrasure is critical (Saran Babu et al., 2019; Kamposiora et al., 2000). In the current research, Group design A, with a larger radius of curvature at the gingival embrasure, had a lower tensile stress strength at the gingival embrasure than Group design B. Group designs A and C with a higher GE curvature demonstrated substantially higher resistance to fracture than group designs B and D with a lower GE curvature. In comparison, there was no critical impact of curvature at the OE on fracture resistance. For designs C and D, the mean failure loads were identical to those for designs A and B. These results mean that as the curvature radius at the GE increased, the failure load increased significantly. This is consistent with finding by Kamposiora et al.(1996) and Pospiech et al.(1996), who reported an increased stress concentration in the connector, and the stress concentration was more significant when the connector height was reduced and the connector width was increased for FPDs (Kamposiora et al., 1996, and Pospiech et al., 1996). However, Bahat et al. (2009) reported that the curvature radius of the gingival embrasure plays an essential role in the load-bearing power. Plengsombut et al.(2009) said that the nature of the connector affects the fracture resistance of milled ceramic, not pressed ceramic. In 2018, Heintze et al. concluded that a decrease in the size of the connector resulted in a substantial reduction in resistance to fatigue. For the longevity of FPDs, appropriate connector sizes are vital. Hafezeqoran et al. (2020) concluded in their study that intelligent embrasure design is not recommended for high-stress areas with minimal occlusogingival height. They also added that the 9-mm^2 connector size recommended by the manufacturer for 3-unit monolithic zirconia fixed dental prothesis should be used more carefully. Moreover, Miura et al. 2017 added that the design with expanded framework height and framework width could stop stress concentration and protect the abutment. At the same time, Bakitian et al.(2019) found that Sharp embrasures significantly reduce the load-bearing capacity of monolithic zirconia FPDs compared to FPDs with blunt embrasures. Junker et al. (2019) supported the theory of decreased connector diameters of the three-unit lithium disilicate FPDs, resulting in reduced mean fracture resistance. Most previous studies have shown that fixed partial dentures with small gingival embrasure radii are exposed to high-stress concentrations during loading in the connector area, compared to FPDs with large embrasure radii.

Anusavice (1996) suggests that an internal compressive stress pattern with tensile force at the maxilla and a load applied to the peri-alveolar portions of the gonial sockets and on the gum hasculus, and through a three individual loading fixture in the facial portions of the dentalium prosthesis creates a stable posteriorly compressed condensation of pressure distribution inside the FPD, the dentalium. Tensile stress has led to fractures that can be lower when dealing with ceramic joints, so this can be used to inform the design of three-part ceramic FPDs. In addition, expanding the cervical flapped brackets without sacrificing esthetics isometry (and vice versa) may increase the interproximal occlusal flange joint surface so that the maximum curvature can be reached (Peterson et al., 1998).

According to the findings of this study, cracks occurred nearly equally on the mesial and distal sides of the pontic. According to the study by Kelly in 1999, the moment arm's location (relative to the points of contact) affects the length of the occurrence of a fracture. Cracks can sometimes develop in a connector area further away from the area of the actual contact as well. Though the significance of the stress distribution has not been tested in the current study, it is expected that the occlusal contact expansion will improve its expression. We should be looking to expand the existing amount of occlusal contacts. When a medical student at Tokyo University carried out an experiment on horse legbones in 2011, they discovered fractures near the abutment end of the bridge segment.

In vitro studies did not resemble repeated dynamic functional loading. Additionally, the direction and mode of loading can differ significantly between individuals and the same individual. Hamza et al., in 2016, agreed with that and added that such in vitro studies allow tested groups to be compared but cannot yet be correlated with the clinical conditions. Zarkovic et al., 2019 added that a more comprehensive approach must be used in future trials, such as the standardized abutment material, simulation material for periodontal support, and modern digital methods with long-term follow-up more clinical studies. In this study, all the loads were applied to the middle of the pontic. In all-ceramic three-unit FPDs, clinical contact areas on the pontic and the adjacent abutments can establish differences in the mode of failure. The thermal and mechanical ageing procedure used is well-known and simulates five years of clinical service (Nothdurft et al., 2010). The measured results may not be directly comparable to intraoral measurements, and ceramic prostheses may fail at values other than those measured in this study. Additional clinical trials are necessary to confirm current findings.

CONCLUSIONS

The limitations of this study, as well as the conclusions drawn from this in vitro study, were as follows:
1.The connector design affects the fracture resistance of all-ceramic fixed partial dentures.
2. It has been demonstrated that altering the radius of the embrasure curvature in the connector area significantly affects the fracture probability of all-ceramic FPDs.
3. The radius of connector curvature at the gingival embrasure significantly affected the fracture resistance of the all-ceramic FPDs.
4. It was discovered that as the radius of the gingival embrasure decreased, the stress concentration values increased.

CONFLICTS OF INTEREST

The authors declare no conflicts of interest

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