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FRACTURE RESISTANCE EVALUATION OF CAD/CAM MONOLITHIC ZIRCONIA FDPS WITH DIFFERENT SPLIT PONTIC DESIGNS RESTORING TILTED MOLAR ABUTMENTS

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ABSTRACT

Purpose: The purpose of the study was to evaluate the fracture resistance of CAD/CAM fabricated monolithic zirconia FDPs with differently designed split pontics under simulating ageing conditions.

Methods: A total of 21 FDPs with split pontics were constructed over 21 dental epoxy resin casts representing missing mandibular first molar with a second premolar abutment of 0° angulation and a tilted second molar abutment of 25° angulation. The FDPs were divided into three groups (n=7) according to the split pontic design; Group (K): Keyhole, Group (B): Bone, Group (R): Relief cut. The cemented FDPs were subjected to thermocycling and mechanical loading in a chewing simulator equivalent to one-year clinical service. Fracture resistance of aged specimens was measured by a universal testing machine and followed by fracture mode detection by scanning electron microscope. One-way ANOVA and Bonferroni post-hoc tests were utilized to compare among groups. The significance level was set at $P \leq 0.05$.

Results: There was a statistically significant difference among groups (*P*-value <0.001, Effect size =0.852). Pair-wise comparisons among the groups revealed that group (K) showed the statistically significantly highest mean fracture resistance (2148.6 ± 193 N), and followed by group (R) (1605.4 ± 286.2 N). While group (B) showed statistically significantly lowest mean fracture resistance (1027.6 ± 91.8 N).

Conclusions: The tested different split pontic designs of monolithic zirconia FDPs for tilted molar abutments demonstrated acceptable fracture load values.

KEYWORDS: Tilted molar; split pontic; non-rigid connector; monolithic zirconia and ageing.

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INTRODUCTION

Tilted molar abutments are a challenge esthetically and periodontally. The main problem to treat tilted molar situation with a fixed dental prosthesis (FDP) is a proper common path of insertion. Orthodontic uprighting of tilted abutments can be considered as a treatment approach. This could also be solved aggressively by intentional endodontic treatment if the preparation can't solve it alone. But the mechanical solutions of the split pontic, locked attachment (Non-Rigid Connector-NRC) and the telescopic retainer are available and must be considered.[1]

Split pontic is known as NRC placed entirely in the pontic and is useful for tilted abutments. In Layman term, one half of the pontic (e.g., the bottom half) is attached to a retainer, while the other half of the pontic (e.g. the top half) is attached to another retainer. When the individual retainers are placed, their extensions form the entire pontic. The two pieces of pontic are not cemented together for relieving stress at midspan on long pontics or to align the distal path of insertion of tilted abutment. The split pontic stress-releasing effect acts as "safety valves" against the connectors leverage forces as they give chance for rotation and resiliency between the prosthetic restoration and abutment teeth. Split pontics are expected to provide promising design for prosthesis stability and retention with accepted esthetics.

All ceramic FDPs are more utilized in clinical domain due to their esthetic values and excellent biocompatibility. Ceramic strength and fracture toughness are continuously under improvement to overcome their weakness. The application of the computer aided design/computer aided manufacturing (CAD/CAM) technology [2,3] has made zirconia popular in dentistry and recently, the ability to fabricate NRCs with different designs and dimensions from zirconia has been suggested.[4-6] Split pontics as a modification for NRCs with the aids of designing software libraries can be fabricated with different designs and dimensions from zirconia.

However, there has not been any investigation into the impact of different split pontic designs on the fracture strength of monolithic zirconia FDPs. Hence, the purpose of the present study is to evaluate the fracture resistance of CAD/CAM fabricated monolithic zirconia FDPs with different split pontic designs under simulating ageing conditions in a form of thermocycling and cyclic loading. The null hypothesis was that different split pontic designs have an influence on increasing the fracture resistance of the FDPs.

MATERIALS AND METHODS

Sample size calculation and grouping

The power analysis used the fracture resistance as the prime outcome. The effect size f = 0.7897 was calculated based upon the results of a pilot study conducted on three specimens for each group utilizing alpha (α) level of (5%) and Beta (β) level of (20%) i.e. power = 80%; the least estimated sample size was a total of 21 specimens (7 specimen s per group). Pilot study specimens were included in the main study. Sample size calculation was carried out by using power analysis software (G*Power; Version 3.1.9.2, HHUD, Germany).

Twenty-one 3-unit monolithic zirconia FDPs with split pontics replacing a missing mandibular first molar were fabricated and divided in accordance with the split pontic design into three groups (7 specimens for each): Group (K): Keyhole (4.8 mm buccolingual dimension-BL), group (B): Bone (4.4 mm BL), group (R): Relief cut (3.5 mm BL).

Preparation of abutment teeth and fabrication of experimental casts

To simulate the clinical missing mandibular first molar, an ivorine (Model #R861; Columbia Dentoform Corporation, Long Island City, NY, USA) mandibular second premolar of 0° angulation and tilted mandibular second molar of 25° angulation

[7] were placed in a mandibular arch model that were guided during placement by a plastic angle guide sheet. The distance between abutment teeth was 11.6 mm, equivalent to the regular first molar length[8]. Silicon indices for the unprepared teeth were used to check preparation depths [9]. Preparation of abutment teeth was standardized to receive full coverage all ceramic retainers with 12° as a total convergence angle, a rounded shoulder margin of 1.0 mm thickness circumferentially, 1.5 mm anatomical occlusal reduction, an axial reduction of 1.0-1.5 mm, and a preparation height of 3.0 mm. All sharp points and line angles were rounded off [10].

Twenty-one half-arch impressions were made for the prepared teeth with a polyvinylsiloxane impression material (President Coltène, Coltène/whaledent GmbH, Germany) utilizing one-step technique. The impression was inspected under 4x magnification fiber optic microscope for any tears or defects which if observed, the impression was retaken again. Twenty-one epoxy resin casts were fabricated accurately, on which the retainers were cemented, utilizing a dental epoxy resin material (Die Epoxy Type 8000 System, American Dental Supply, Inc, Allentown, PA, USA) that was blended following the manufacturer manual and poured into twenty-one half arch impressions [11].

Designing and fabrication of zirconia FDPs

Zirconia FDPs were fabricated from high translucent zirconia blanks (GC Initial Zirconia Disk; 18 mm CIP HT, GC America INC., ALSIP, IL) that are ideal for full contour and monolithic restorations. The zirconia blanks are composed of cold isostatic pressed (CIP) zirconium silicate crystalline white powder to optimize physical properties and organic binders enabling to press the powder into blanks. Fabrication took many phases; acquisition was done occlusally, buccally, lingually, and proximally utilizing a 3D dental scanner (Identica hybrid; MEDIT corp., Korea) and scanning software (collab Scan, V 2.0.0.3; MEDIT corp., Korea) to have 3D model of preparation and followed by FDP design utilizing a software (exocad Dental CAD; exocad GmbH, Germany) where all the "Biogeneric" preparation margins and base lines for FDP had been entered. Then, the pontic location center and design were chosen. The axis for each tooth individually was defined where the software stipulates that both levels must be positioned vertically to the insertion axis (Fig. 1).

The cement space was set to 100 µm. While the occluso-gingival dimension, mesio-distal dimension, and two halves interface space of the split pontic for all specimens were set at 8 mm, 6 mm, and 100 µm respectively. While the split pontic buccolingual dimension was set according to split pontic design into: Keyhole (K) (4.8 mm), Bone (B) (4.4 mm), Relief cut (R) (3.5 mm) (Fig.2-4).

The stereolithography (STL) file was sent to a milling machine (K5; vhf camfacture AG, Germany) for milling the designed FDPs followed by sintering at 1550°C in a special sintering furnace (Tegra speed; Teknik Dental, Turkey) for 9 hours. After milling, the FDPs components were detached from their milled blanks. Each specimen was finished properly by applying the glaze (Clear glaze; CERABIEN ZR, Kuraray Noritake Dental Inc., Chiyoda City, Tokyo, Japan) and baking according to manufacturer instructions. After sintering (Fig 3), the marginal fit and fitting surface of the finished FDPs were checked for defects under 4x magnification fiber optic microscope.

Cementation

Before bonding, the epoxy resin abutment teeth were sandblasted with 50 µm aluminum oxide particles at 2.8 bars for 13 sec at a distance of 10 mm, then cleaned and air-dried [12]. The internal surfaces of the retainers of zirconia FDPs were air abraded with 110 µm aluminum oxide particles at 2 bars for 10 seconds from a perpendicular distance of 10 mm [13]. All FDPs were cleaned in an ultrasonic cleaner for 5 minutes and air- dried [14]. Each FDP then was cemented to its corresponding epoxy master die with a self-adhesive resin cement (RelyX

Unicem, 3M, ESPE, St. Paul MN, Germany) that was mixed according to the fabricator specifications and seated on the prepared dies with an especially designed static loading device of 10 kg and a metal bar with metal-acrylic cementation index that was adapted on the occlusal surface of the FDP for 10 minutes to ensure an effective flow of cement and the maximum adaptation of the retainers [15], surplus cement was cut off, and then a definitive curing took place by light-curing of each surface for 20 seconds.

Artificial ageing program of the cemented FDPs

The specimens underwent a combined chewing simulation in a multifunction chewing simulator (ROBOTA chewing simulator; ROBOTA Model ACH-09075DC-T, LTD., Germany) integrated with thermo-cyclic protocol operated on servo-motor (Model ACH-09075DC-T; AD-TECH TECHNOLOGY CO., LTD., Germany). The tested specimens were exposed to 240,000 chewing cycles imitating clinical one-year service [16] with a steatite antagonist (diameter: 5.6 mm) at a load of 49 N and 1.6 Hz frequency. The point of load was concentric in the masticatory center (central pit of the pontic). A sideward movement (1mm; oral-vestibular) was executed. Only specimens of the same group were loaded simultaneously in the chewing simulator. The load application was associated with thermocycling procedure, including the immersion in cold/hot water bath for 1200 thermo-cycles representing one year clinically [17] with temperature variation 5˚C/55˚C and dwell time 60 seconds.

Measurement of fracture resistance after aging

The load-to-failure test was done utilizing a universal testing machine. Each specimen was mounted onto the lower fixed compartment of a computer-controlled testing machine (Model 3345; Instron Industrial Products, Norwood, MA, USA) with a load cell of 5 KN and data were recorded utilizing computer software (Bluehill Lite Software; Instron, PA, USA). A specially designed load appliance of one head metallic attachment (5.6 mm ball) was used for compression load until failure at a crosshead speed 1mm/min placed centrally at the occlusal surface of the pontic [18]. A tin foil sheet was placed in-between to obtain consistent stress distribution and minimizing the transmitted local force peaks. The failure load was observed by an audible crack and approved by a sudden drop at load-deflection curves recorded with computer software. The load required to fracture was recorded in Newton.

Post testing fractographic analysis

Fractured pieces of tested specimens were scanned under scanning electron microscope (SEM) (JEOL 6390 LV, Jeol, Peabody, MA, USA) to analyze the fractured segments.

Statistical analysis

Numerical data were investigated for normality by checking the distribution of data and utilizing normality tests (Kolmogorov-Smirnov and Shapiro-Wilk tests). Data showed normal (parametric) distribution. Data were presented as mean and standard deviation values. A one-way ANOVA test was utilized to compare among the three groups. The Bonferroni post-hoc test was used for pair-wise comparisons. The significance level was set at $P \leq 0.05$. Statistical analysis was performed with statistics software (IBM SPSS Statistics for Window; Version 20.0, IBM Corporation, Armonk, NY, USA).

RESULTS

Fracture resistance

There was a statistically significant difference among the groups (*P*-value <0.001, Effect size = 0.852). Pair-wise comparisons among the groups revealed that group (K) showed the statistically significantly highest mean fracture resistance value. Group (R) showed statistically significantly lower mean value. While group (B) showed statistically significantly lowest mean fracture resistance value (Table 1).

Fracture mode

During the thermocycling and mechanical loading (TCML), none of the investigated FDPs failed. Visual inspection of the specimens revealed that the fractured modes could be classified as chipping and catastrophic fracture (Fig. 6a-d). For all split pontics fractures, the fracture was originated in the top half, close to its lingual surface in different manners in three groups (Table 2).

SEM at 120 and 200 x power for a typical fracture surface of a failed FDP showed cracks originated at the loading point then spread radially with a surrounding smooth mirror region (Fig. 7), hackle patterns in the periphery (Fig. 7-10), and arrested lines finally ended at the edge of the pontic (Fig. 10). The presence of compression curls on the fractured surface opposite to the crack origin was noticed also (Fig. 7).

TABLE (1) Descriptive statistics and results of one-way ANOVA test for comparison among fracture resistance of the three groups.

Group (K) (n = 7)	*Group (B) (n = 7)*	*Group (R) (n = 7)*	*P-value*	*Effect size (Eta squared)*
Mean (SD)	**Mean (SD)**	**Mean (SD)**		
2148.6 (193) A	1027.6 (91.8) C	1605.4 (286.2) B	<0.001*	0.852

****: Significant at P ≤ 0.05, Different superscripts are statistically significantly different.***

TABLE (2) Fracture mode for differently designed split pontic specimens.

Group	*Specimen No.*	*Mode of Fracture*
K	4	A catastrophic pontic and molar retainer fracture and debonding.
	3	A catastrophic fractured pontic and debonding.
B	5	A catastrophic fracture where cracks ran through the split pontic and debonding.
	2	A chipping in the top half of split pontic.
R	4	A catastrophic fracture in in the top half of split pontic and debonding.
	3	A chipping in the top half of split pontic.

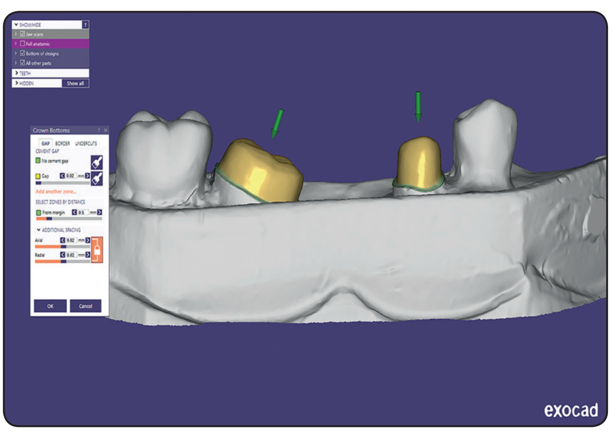

Fig. (1) Definition of the axis for each tooth individually with the insertion axis tools.

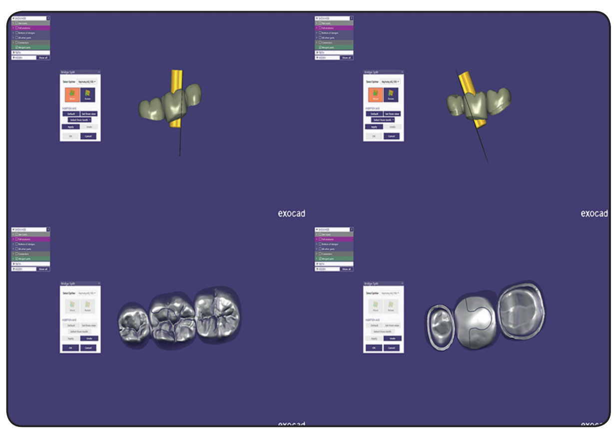

Fig. (2) Keyhole split pontic designing.

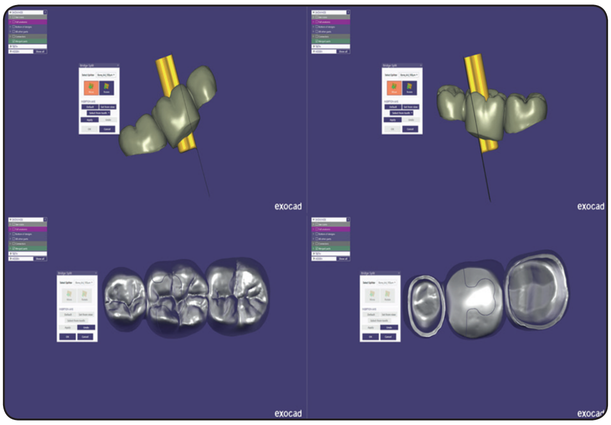

Fig. (3) Bone split pontic designing.

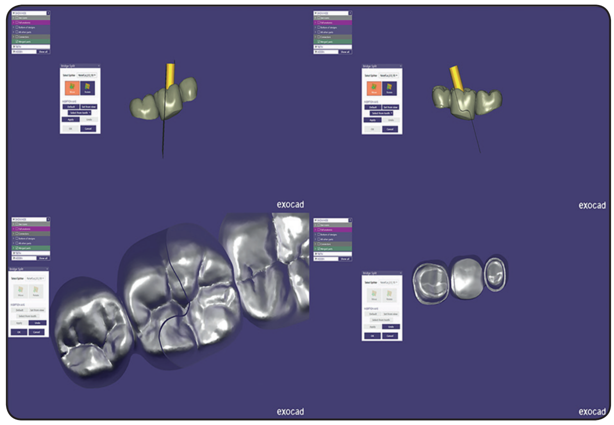

Fig. (4) Keyhole split pontic designing.

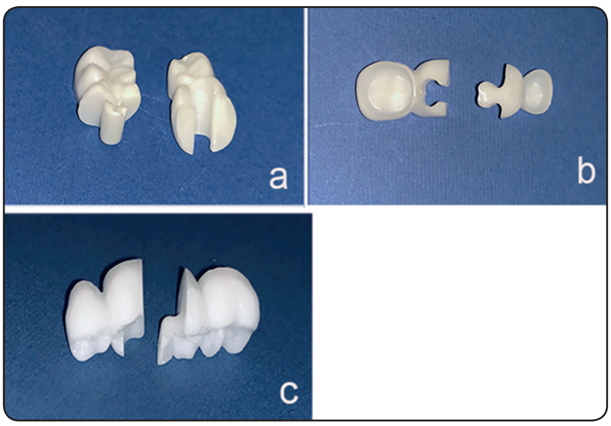

Fig. (5) Finished differently designed split pontic FDPs a) Keyhole split pontic design, b) Bone split pontic design and c) Relief cut split pontic design.

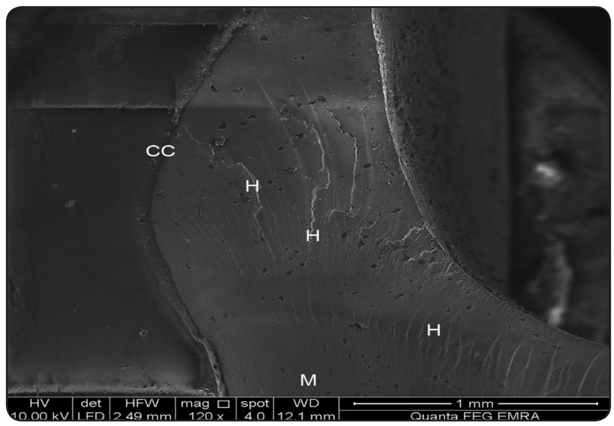

Fig. (7) SEM photograph (120x) of a fracture surface in split pontic showing a smooth mirror region (M), hackle twist lines (H) and compression curls (CC).

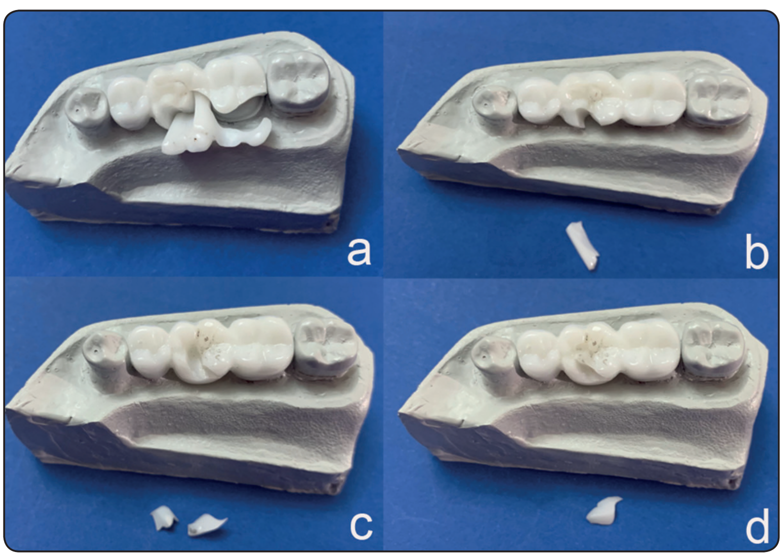

Fig. (6) Fractures of differently split pontic designed FDPs; a) Key hole design specimen showed a catastrophic pontic and molar retainer fracture, b) Bone design specimen showed a catastrophic fracture where cracks ran through the split pontic, c) Relief cut design specimen showed a catastrophic fracture in in the top half of split pontic and d) Relief cut design specimen showed a chipping in the top half of split pontic.

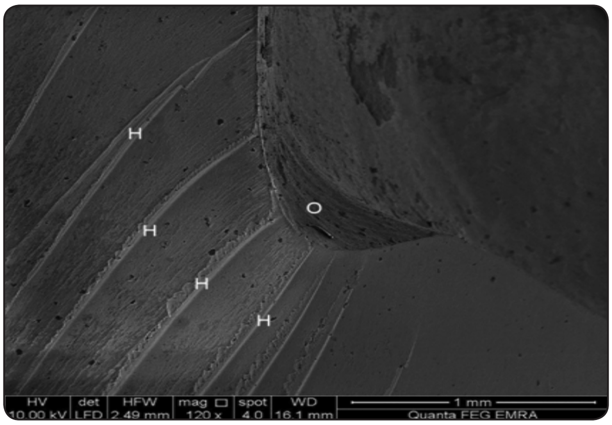

Fig. (8) SEM photograph (120x) representing hackle twist lines (H) having a river like pattern can be observed, converging towards the origin (O) located at the surface in contact with the indenter.

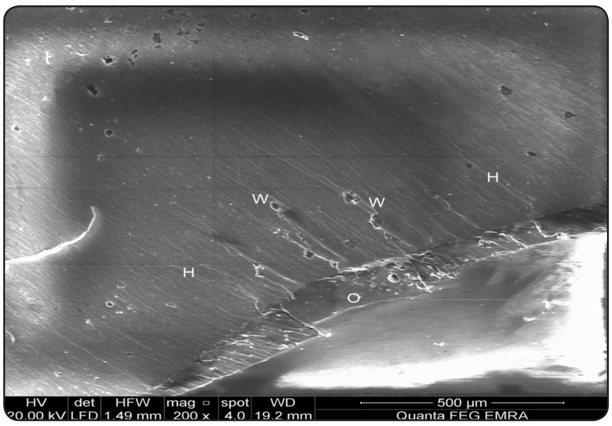

Fig. (9) SEM photograph (200x) representing hackle twist lines (H) converging towards the origin (O) and wake hackle (W) is noticed.

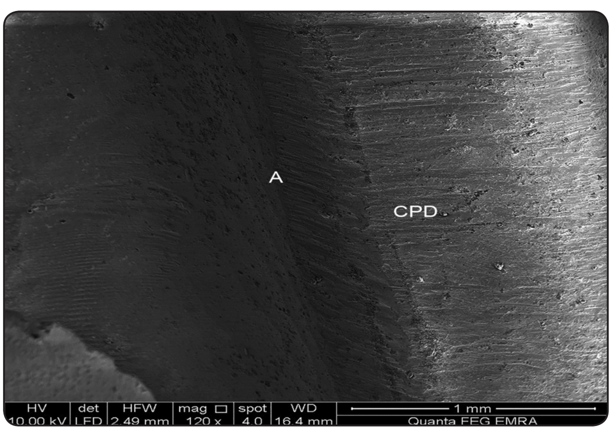

Fig. (10) SEM photograph (120x) of a fracture surface in split pontic showing crack propagation directions (CPD) with arrest lines (A).

DISCUSSION

The massive posterior loads with tilted molar abutment and the sharp embrasures at connector areas result in concentrated stresses, which increase the all ceramic FDPs failure chance. Split pontic as a mechanical solution after the preparation of abutment teeth according to their long axes is a treatment choice for mesially tilted molar abutment. There are always trials to optimize the connector dimensions, but the connector area was still the most determining factor [19]. The modification of the connector design as the alternative orientation on the NRC where the key and keyway were reversed. The key was attached to the distal surface of the anterior retainer, while the keyway was incorporated in the pontic. This orientation leads to conservative tooth preparation without more space for the NRC ensuring retention, solving the angulation flexibility, paralleling the conflict in the NRC to a mesially inclined posterior abutment and physiologic axial contouring as the NRC are totally within the pontic contour [20] avoiding an extensively large cross-sectional area in an FDP connector isn’t recommended morphologically, biologically and esthetically [19].

It was reported that the zirconia based three-unit FDPs fracture strength was significantly higher than those constructed with IPS Empress (Ivoclar-Vivadent) and In-Ceram Alumina (Vita) ceramics [21]. That study [21] and others [22-24] have stated that zirconia-based FDPs have the tolerance capacity to physiologic occlusal forces exerted in the posterior region and may be impressive substitutes to ceramo-metallic protheses. A loading point and one or both connectors fractures are the most common fracture patterns for tested zirconia-based FDPs [25, 26]. Therefore, the connector design seems to be crucial for the fracture resistance and durability of zirconia-based FDPs. When FDP for the molar region is prepared with zirconia-based ceramics, it is recommended a connector area of 9-16 mm^2 to be more than 3.4 mm occluso-gingivally and buccolingually [27]. Therefore, the different split pontic designs used in this study aimed to avoid that drawback of limited connector design space by using an inverted orientation with a maximum occluso-gingival height (8mm) as some studies and manufacturers suggest larger vertical connector dimensions (minimum 4.0 mm) than horizontal ones as the main load directed vertically [19,28.] However, other authors have shown that horizontal forces sometimes significantly affect [28,29]. The buccolingual dimensions were set according to different designs used (K: 4.8 mm, B: 4.4 mm, and R: 3.5 mm).

Fracture forces assessed in the present study might have been higher than in clinical practice. Since the abutment teeth were made of epoxy resin with 20 GPa elastic modulus which is high in comparison to dentine (12 GPa) where Scherrer and Rijk stated that the fracture strength is directly proportioned to the elastic modulus of the supporting material [30]. However, studies utilizing resin material for the abutment teeth reported similar fracture forces for zirconia-based FDP [31].

Clinically, all ceramic restorations fail through slow crack growth as a result of many cycles of stress, so-called fatigue failure [32]. The moisture and a controlled temperature to the test conditions in this study was designed to simulate the intraoral environment [1]. For the applied forces simulating the posterior teeth physiological biting forces, the magnitude, duration and frequency values were comparable to ones reported in the literature to the pattern of load application [33, 34]. In vitro studies found that the functional forces arising during mastication usually range between 2 and 50 N [4,35]. 240,000 fatigue cycles were applied by using the chewing simulator and they correspond to a period of 1 year of clinical service [33].

In the present study, tested groups were subjected to central load-to-failure and contacts were developed as clinically relevant as possible [1]. A 5.6 mm diameter ball was used for load application similar to that of a molar cusp. Hence, the wear facet is an area (0.5-3.0 mm) rather than a point and the contact area and the magnitude of force have an influence on failure mechanisms during function.

The maximum bite force varies among patients, intra-individually by the time and significantly from one area to another intra orally, being ~ 90-340 N in the anterior region, 220-450 N in the premolar region, and 400-900 N in the molar region [36]. In this study, the single load-to-failure values were 2148.6 ± 193 N for group (K), 1027.6 ± 91.8 N for group (R) and 1605.4 ± 286.2 N for group (B). All values were still much higher than 1000 N, in agreement of being as the minimum required material resistance to apply at the posterior area [21,37,38] and support our suggested split pontic designs impact on restoration fracture resistance.

Higher stress values obtained by centrally loaded tested groups were observed in comparison to values found clinically [33]. However, the single loading-to-failure test does not necessarily reflect clinical conditions since few clinical fractures were observed after a single loading [39]. Still, it can help as a starting point for investigation of new materials or concepts, allowing certain factors standardization that are difficult to standardize clinically [40]. Moreover, differences in loading dynamics and stress distribution often lead to overestimation of the fracture load observed under laboratory conditions [41].

The null hypothesis was accepted as a significant difference among groups was detected. In comparison among tested groups, fracture resistance load values for group (K) showed the statistically significantly highest values (2148.6 ± 193 N). While group (R) showed statistically significant lower mean value (1605.4 ± 286.2 N). Group (B) showed the statistically significant lowest mean fracture resistance value (1027.6 ± 91.8 N). This is explained by the fact of torque and shear stresses increasing leads to fracture resistance decreasing eventually. While increasing of buccolingual dimensions leads to enlarging the surface area, bulk, retention, and so rising the resistance to the torque and fracture.

However, no available reference in dental literature was found in relation to the effect of the design of the split pontic on the stress distribution. On the other hand, group (R) with 3.5 mm buccolingual dimension showed significant higher fracture resistance load values (1605.4 ± 286.2 N) than group (B) with 4.4 mm buccolingual dimension; this could be explained by the design geometry with an area of constriction in group (B) (1027.6 ± 91.8 N) that could be a weakness area due to decreased width. The previous statement could be also supported by group (K) results (2148.6 ± 193 N), as there was not a sudden constriction as found in group (B) (1027.6 ± 91.8 N).

One of the important factors in ceramic materials is the bulk as its increase leads to higher fracture resistance and vice versa. According to K and B pontic designs, there was decreasing in material thickness (bulk) in the top part of split pontic buccally and lingually to accommodate the bottom half. Applied forces during chewing simulation resulted in fractures mainly cited in the lingual part that can be explained by decreased anatomical axial contour compared to the buccal aspect of group (K) and (B). While in group (R), a catastrophic fracture or chipping occurred in the top half of split pontic as there was a support from the bottom half and this supported the effect of the vertical bulk thickness in increasing fracture resistance. As it had been hypothesized that fracture initiation sites in dental ceramics could be controlled by changing the ceramic thickness [42].

For most specimens, the fractures occurred at the point of loading and through the top half of split pontic. When occlusal forces are directed through the long axis of zirconia FDP pontic, compressive stresses evolve at the occlusal aspect of the pontic at the marginal ridge, and tensile stresses develop at the gingival surface of the top half which contributes to the propagation of microcracks located at the gingival surface in an occlusal direction and may eventually result in fracture. The fractures, however, were perpendicular to the mesial-distal axis of the FDPs in a smooth curve between the point of loading and the connector gingival side. Those findings were approved as shown in SEM findings (Fig. 7-10).

According to the present study, it is recommended to use Keyhole designed split pontic in clinically tilted abutment cases due to high fracture load value. However, clinical studies are desired for different designs, to prove the findings of this in vitro study. Furthermore, more split pontic designs with different dimensions utilizing different designing softwares should be thoroughly investigated. In addition, using periodontal ligament simulating condition should be evaluated as it may have a significant effect on fracture resistance and fracture modes.

CONCLUSIONS

The tested different split pontic designed monolithic zirconia FDPs for tilted molar abutment demonstrated high fracture load values where Keyhole design showed the highest value. This seems to be an encouraging treatment substitute for patients with tilted abutments.

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