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ORIGINAL ARTICLE

Fracture behaviour of zirconia ceramic cantilever fixed dental prostheses *in vitro*

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Abstract

Objective. Evaluation of the fracture resistance of all-ceramic cantilever fixed dental prostheses (FDPs) manufactured from zirconia frameworks and veneered with a press ceramic. ***Material and Methods.*** Two mandibular premolars were prepared either with a box inlay cavity or with a full crown chamfer preparation and then duplicated. 40 three-unit cantilever FDPs replacing one premolar, with a group size of eight for each design, were manufactured. In group i-i the cantilever FDPs were retained by two inlays, in group i-c by an inlay–crown combination, and in group c-c by two crowns. The frameworks in groups i-c-R and c-c-R were reinforced by an additional shoulder on the oral side of the zirconia frameworks. All FDPs were subjected to thermal cycling (TC) and 600,000 cycles of mechanical loading (ML) with 50 N. The load to fracture was measured and fracture sites were evaluated. ***Results.*** The mean fracture values ranged from 172 N to 792 N. Fracture-strength values were significantly lower for the i-i retained FDPs than for the i-c and c-c combinations. There was no significant effect of the reinforcing shoulder in groups i-c-R and c-c-R. For FDPs with a crown on the terminal tooth, fractures were usually within the distal wall of the distal crown. ***Conclusions.*** Inlay–inlay retained cantilever FDPs cannot withstand the mastication forces expected. Fracture load values for inlay–crown and crown–crown-retained FDPs encourage further clinical investigation. The mode of fracture indicates that reinforcement of the distal crown wall might enhance fracture resistance.

Key Words: *All-ceramic, CAD/CAM, fdp, fracture strength, Yttrium-TZP*

Introduction

Cantilever fixed dental prostheses (FDPs) have been defined as multiple retainers with one or more unsupported free-end extensions. They seem to be clinically successful [1,2] in a shortened dental arch (SDA) when one or more tooth gaps are available and replacement of the missing teeth by implants is contraindicated for anatomic, medical, economic, or psychological reasons. A cantilevered FDP is favoured for replacing molars and to avoid unilateral removable partial dentures (RPDs). The pontic prevents over-eruption of an opposing tooth and contributes to the stability of opposing removable dentures [3]. The restorative material used for the FDP must be extremely rigid and metal alloys are currently the gold standard. To achieve maximum retention and resistance, complete crowns are the preferred retainers for the cantilever FDP [4]. Some authors recommend at least two abutments and the replacement of one tooth only [5–7], whereas others advise use of three abutment teeth [8]. An esthetic drawback of the metal-based cantilever FDP is the visibility of the metal-retainer and the change in natural tooth translucency. The desire for natural looking and metal-free restorations has encouraged research on metal-free, tooth-colored materials.

One metal-free alternative may be use of all-ceramic cantilever FDPs. Early-generation all-ceramic FDPs often failed to withstand posterior mastication forces, and their use was limited by the mechanical properties of the material.

With the introduction of densely sintered yttrium-stabilized tetragonal zirconia polycrystals (Y-TZP) and the ability of Y-TZP to transform, thus preventing crack propagation [9], fabrication of stable FDPs has become possible, even if a less invasive inlay-preparation design is used [10]. Zirconia-based materials have an esthetic disadvantage, however,

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because of the high opacity of the material. An alternative approach is to combine the excellent esthetics of conventional press ceramic with the high strength of Y-TZP. This study was performed to test the hypothesis that all-ceramic FDPs with one distal cantilever combined with Y-TZP/ IPS e.max ZirPress (Ivoclar Vivadent) would be strong enough to withstand expected posterior mastication forces. The effects of abutment preparation design and the dimensions of the zirconia frameworks were evaluated.

Material and methods

Forty cantilever FDPs were made from an individually manufactured CAD/CAM Y-TZP framework veneered with IPS e.max ZirPress.

Preparation of the abutments

Defined box inlay cavities were made on 1st and 2nd mandibular premolars of the Frasaco study model (Frasaco, Tettnang, Germany). The occlusal box extended 2 mm in the occluso-cervical direction, 2 mm in the mesio/distal to central direction, and 2.5 mm in the facial to oral direction. The proximal box had to extend 2 mm apically to the occlusal box and 1 mm in the mesial to distal direction (Figure 1a). Crown preparation with a chamfer finishing line was also performed, again on 1st and 2nd mandibular premolars, combined with an occlusal reduction of 1.5 mm and a circular reduction of 1 mm (Figure 1b). To ensure standardized preparation, the teeth were prepared with diamonds with a three-degree angle of incidence (Komet, Lemgo, Germany), using a parallelometer (S-3 Maser; Schick, Schemmershofen, Germany).

Preparation of the models

The prepared teeth were duplicated (Alpa Sil; Alpina, Unterhaching, Germany) and the 48 abutment teeth were cast from a Co–Cr alloy (Remanium 2000®; Dentaurum, Ispringen, Germany). The metal teeth were covered with heat-shrink tubing (CPX 55; Polyolefin, Kentec, Waldshut, Germany) to simulate physiological tooth mobility. Twenty-four models with 1st and 2nd premolars were embedded in PMMA resin (Palapress®; Heraeus Kulzer, Hanau, Germany); a preoperatively prepared polysiloxane splint was used for identical tooth localization. Impressions and stone dies were made to prepare 40 working models for fabrication of the FDPs. The impressions were made with polyether impression material (Impregum®; 3M Espe, Seefeld, Germany), in accordance with the manufacturer's instructions. The stone dies were then cast with suprastone material (Fujirock; GC America, Ill., USA).

Preparation of the FPDs

Five groups of eight cantilever FDPs each were tested. All 40 cantilever FDPs replaced one premolar. In group i-i the cantilever FDPs were retained on two inlays, in group i-c on a combination of an inlay and a crown, and in group c-c on two crowns (Figure 2a–c). Two additional groups of cantilever FDPs with a reinforced (R) zirconia framework including an additional shoulder 2.5 mm high

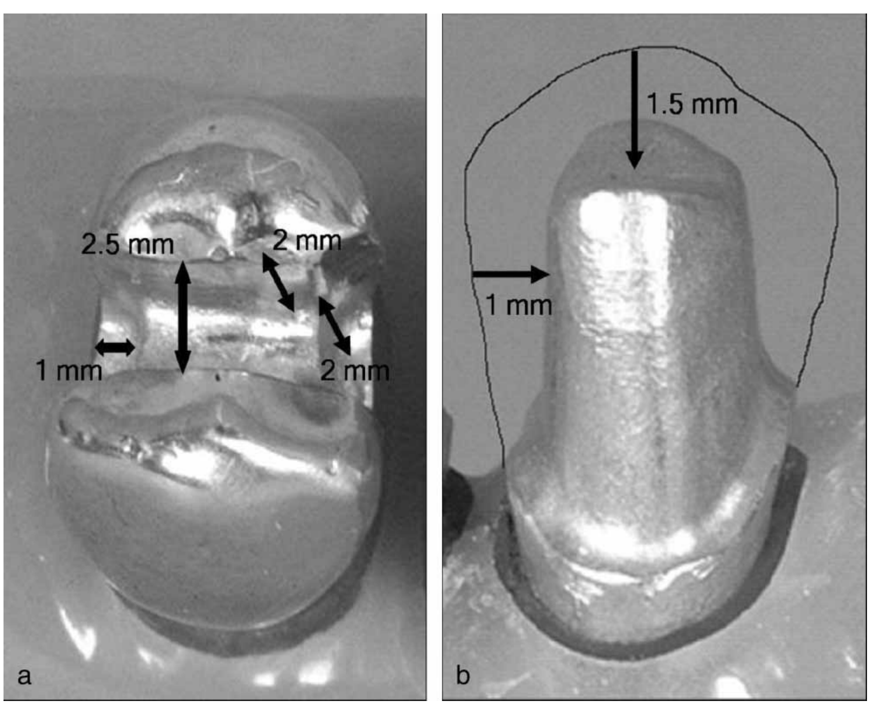

Figure 1. (a) Dimensions of the inlay preparation. (b) Dimensions of the crown preparation.

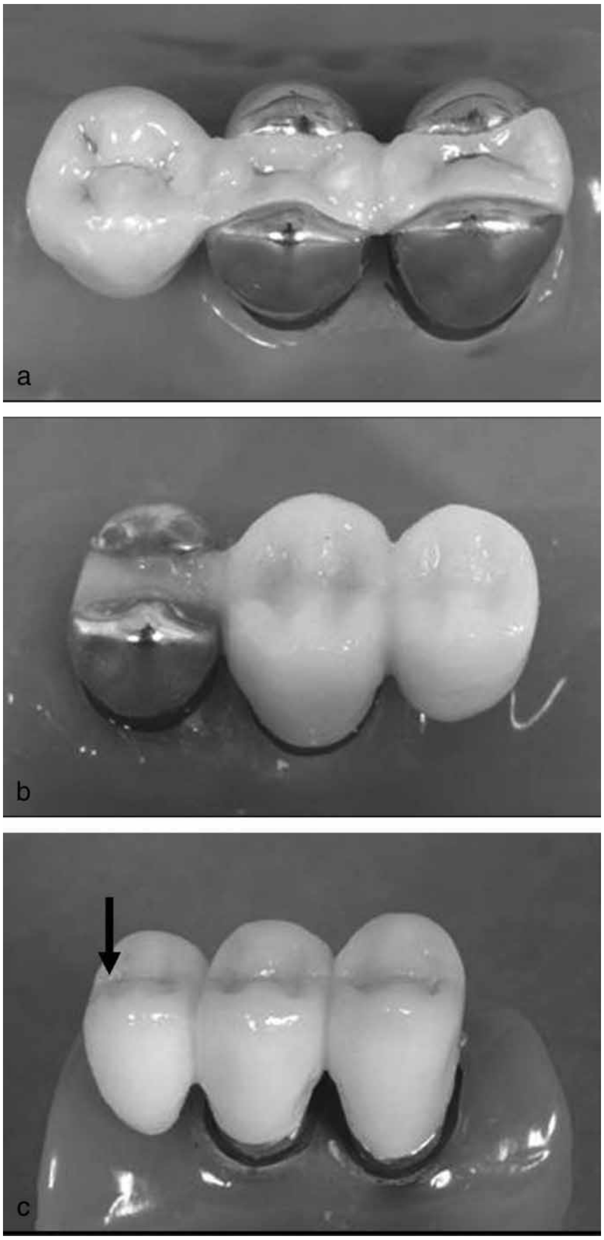

Figure 2. (a) Inlay–inlay retained FDP. (b) Inlay–crown retained FDP. (c) Crown–crown retained FDP (↓ indicates loading point).

and 1 mm wide on the oral side of the FDPs (Figure 3) were also tested. Group i-c-R was retained on a combination of an inlay and a crown and group c-c-R on two crowns. All 40 frameworks were prepared from prefabricated ceramic blanks made of 3% (v/v) yttrium-stabilized zirconia using a Cerec Scan milling machine (Sirona, Bensheim, Germany).

The connectors in groups i-c and c-c had a cross-section of 12 mm^2; those in the i-i group had a cross-section of 9 mm^2. The TZP frameworks were veneered with IPS e.max ZirPress® (Ivoclar Vivadent, Ellwangen, Germany). The procedure included liner firing, main firing, and two glaze firings. The temperature and pressure used were in accordance with the manufacturer's instructions

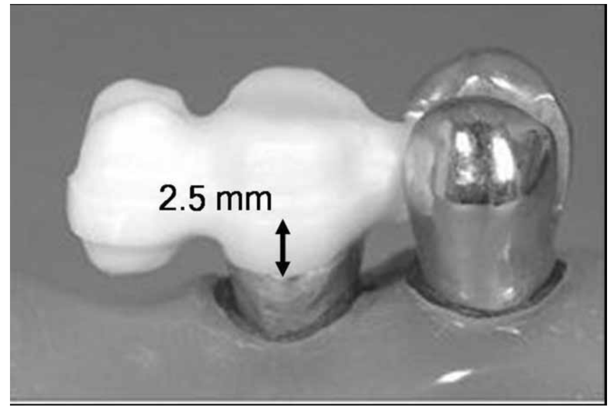

Figure 3. FDPs with a reinforced zirconia framework.

(Ivoclar Vivadent). After finishing, no further grinding or polishing of the surfaces of the all-ceramic FDPs was performed.

Cementation

All metal abutments were cleaned with 90% ethanol. All FPDs were seated with dual-cured resin composite (Variolink II®; Ivoclar Vivadent) in accordance with the manufacturer's instructions.

Cyclic mechanical loading and thermal stress

All FDPs were subjected to artificial aging. In accordance with Krejci & Lutz [11], the following conditions were chosen to simulate 2.5 years of service:

1. 10,000 $\times$ 45 s thermal cycles between 6.5 and 55°C (Willytec, Gräfelfing, Germany) to simulate physiological fluctuation of intra-oral temperatures [12].
2. 600,000 cycles of mechanical loading (ML) with 50 N at a frequency of 1.66 Hz (Willy Tec, München, Germany). The force of 50 N was chosen to simulate the physiological biting force and was applied at the center of the distal fossa of the cantilever by use of a steel ball (8 mm diameter) (Figure 2c).

Fracture strength

The fracture strength was determined by mechanical loading to failure in a universal testing machine (1445; Zwick, Ulm, Germany). The force was applied with a lever of 6 mm to the distal aspect of the pontics (cross-head speed 0.5 mm min^{-1}) (Figure 2c) with 0.5 mm tin foil between the loading element and the pontic to reduce large stresses. Failure was detected as 10% loss of the force applied. The fracture strength was determined by the occurrence of the first fracture, evident as a peak in the procedure. The load was measured with a load cell (type U 2A), including strain gauges, and was recorded with Zwick pc-software.

Statistics

Statistical analysis was performed using SPSS (version 14.01S, Chicago, Ill., USA). Fracture-resistance data were analyzed and depicted as whisker and box plots (Figure 5).

Non-parametric tests [13] (Kruskal-Wallis, Mann-Whitney U-test) were used to determine the effects of differences between the groups. The level of significance (p) was set at $\alpha = 0.05$.

Results

None of the FDPs became unbonded after thermal cycling or mechanical cycling, and no fractures were observed in the veneering material. For FDPs of group i-i the mean (SD) fracture load was 172 ($\pm$37) N; for FDPs of the i-c group the mean value was 580 N ($\pm$138); and for FDPs of the c-c group the mean fracture load was 703 N ($\pm$108). The mean fracture loads of the reinforced groups ranged from 792 N (SD $\pm$259 N) for i-c-R FDPs to 603 N (SD: $\pm$214 N) for the c-c-R FDPs.

Different preparation designs of the abutment teeth had a significant effect on fracture load. The fracture loads for group i-c ($p \leq 0.001$) and c-c ($p \leq 0.001$) were both significantly higher than that for group i-i. There was no statistically significant difference between group c-c and group i-c ($p = 0.066$). Likewise, the values for the reinforced groups i-c-R ($p \leq 0.001$) and c-c-R ($p \leq 0.001$) were significantly higher than the values for group i-i. Between the two groups i-c-R and c-c-R, however, there was no significant difference. Reinforcement of the framework with an additional shoulder in groups i-c-R and c-c-R had no significant effect on fracture load compared with the corresponding FDPs in groups i-c ($p = 0.59$) and c-c ($p = 0.29$).

No fracture of a connector was observed. All fractures in group i-i occurred at the transition from the inlay-base to the proximal box. Twenty-eight FDPs with a crown on the terminal tooth fractured within the distal wall of the distal crown. Fractures within the veneering material of the pontic were observed in four cases only. Detailed fracture modes are given in Figure 4a–c.

Discussion

If all-ceramic cantilever FDPs are to be used as an alternative to metal–ceramic FDPs in the posterior region, the FDPs must fulfil one major criterion — the material must withstand posterior mastication forces of approximately 500 N [14]. On the basis of results from this study, zirconia ceramic cantilever FDPs may be considered for further clinical evaluation if the final abutment tooth, at least, has received a crown preparation.

Predicting the clinical performance of these all-ceramic cantilever FDPs is challenging, because available data for maximum clinical forces in the posterior region and, in particular, on cantilever FDPs vary widely. Some authors have reported mean maximum bite forces in the molar region of 847 N for men and 597 N for women [15], or 859 N for non-bruxers and 806 N for bruxers [16]. For cantilever FDPs, maximum local forces between 150 N [17] and 700 N were reported.

In this study, some of the cantilever FDPs did not survive the required *in vivo* fracture strength values of approximately 500 N [14]. In particular, for cantilever FDPs of group i-i the mean value was approximately 330 N below the required 500 N; this is not sufficient for clinical application.

The fracture load of all-ceramic cantilever FDPs with a crown as terminal abutment exceeded the maximum expected mastication forces of 500 N. The authors could not find any comparable data for all-ceramic cantilever FDPs, but results from this study were lower than available values from other all-ceramic end-to-end FDPs. For end-to-end FDPs with inlays as retainers the mean fracture load was 1,270 N [10].

One reason for the low fracture-load values might be the material itself. Internal flaws, for example porosity or undesirable microstructural features, may be an explanation for the greater variation of fracture-load values. This risk applies to all manually and computer manufactured FDPs, however [18]. Use of industrially manufactured frameworks reduces the risk of manufacturing faults, but each industrially produced frame must be adapted to the abutment tooth and to the occlusion. Consequently, machining these frames cannot be avoided and faults are, therefore, omnipresent.

Another explanation for the results could be the experimental arrangement. Simulating physiological tooth mobility has been shown to result in lower fracture-load values [19]. Use of premolar instead of molar abutment teeth also reduces stability [20], because the smaller surface of the root allows greater deflection of the tooth under load. This leads to higher stresses because of shear forces in addition to the bending forces. The load was, in addition, applied only to the pontic; this has been shown to be the arrangement resulting in greatest stress [7].

All FDPs underwent artificial aging. Strub & Beschnidt found that fracture-load values for all-ceramic crowns were significantly reduced after thermal cycling [21], in contrast to Attia & Kern, who did not observe a general decrease [22]. Likewise, depending on the level and duration of the load, mechanical cycling may lead to a reduction of the fracture-load values of TZP FDPs.

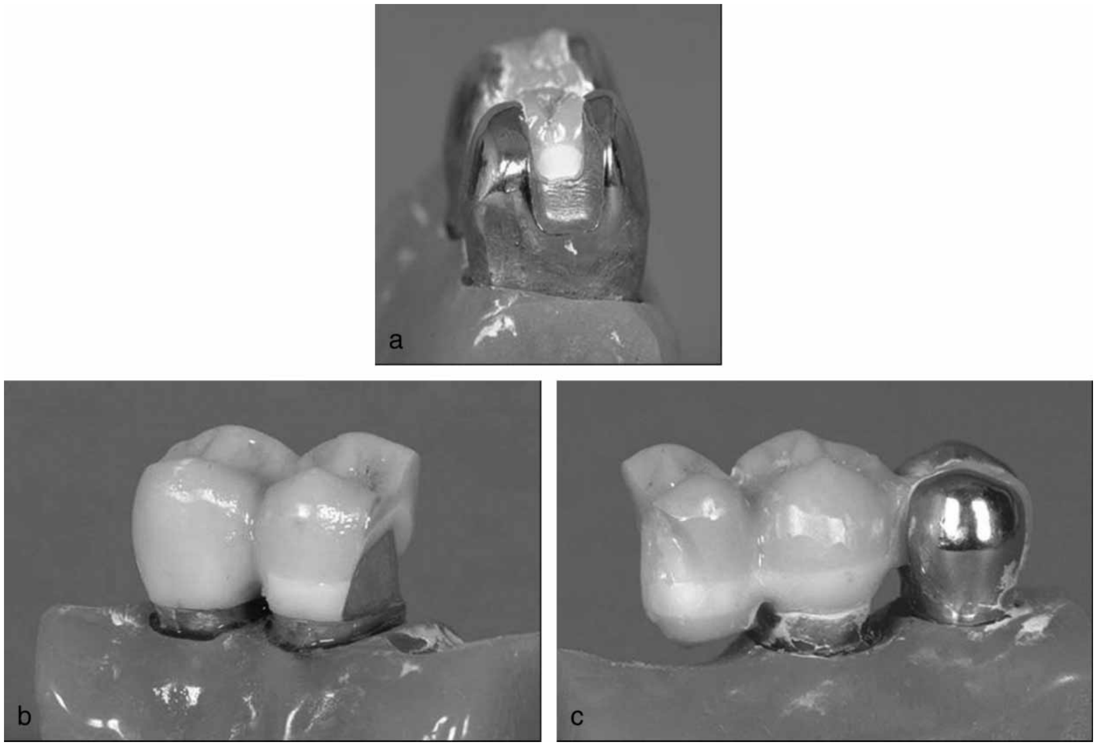

Figure 4a–c. Fracture modes of the cantilever FDPs.

The fracture mode of the FDPs with a crown on the distal abutment tooth was fairly consistent. In this study, a reduced connecting area of the zirconia framework of 12 mm^2 was tested. This relatively gracile design facilitates manufacture of an esthetic product and enables the patient to maintain periodontal health by efficient interdental cleaning. In none of the specimens did the loading result in fracture of the connector, which for a long time has been regarded as the most vulnerable part of FDPs. Although, in this study, only eccentric loading was applied to the distal extension, even the all-ceramic connectors with reduced cross-sections withstood the resulting forces.

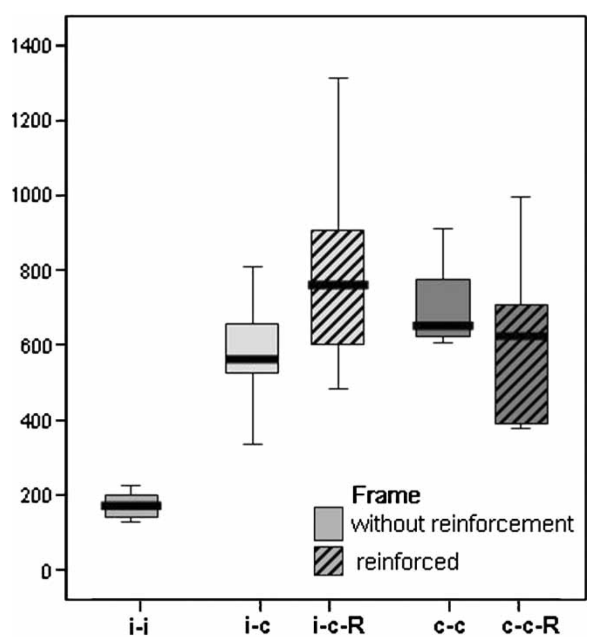

Figure 5. Fracture load values for all FDPs. The boxes represent the 25th and 75th percentiles. Whiskers are drawn to 1.5 × the interquartile range beyond the 25th and 75th percentiles. The bars represent the medians.

Only four FDPs failed because of fractures within the veneering material of the pontic. Twenty-eight from a total of 32 FDPs fractured through the distal wall of the terminal abutment, the region with the highest stress concentration in cantilever FDPs [23]. Other authors have reported the greatest strain to be mesial to the most distal retainer [24]. The observations in this study are in accordance with those of Wylie & Caputo [8] and Awadalla et al. [25], and can be explained as follows. The axial load on the cantilever extension produced reactive forces distal to the terminal abutment–compressive strain in the lower part of the distal crown wall and tensile loading in the upper part. Likewise, in a finite element analysis, Romeed et al. [7] found the maximum principal stresses concentrated subjacent to the occlusal surface of the connector between the retainer and pontic of cantilever FDPs. The fracture mode in this study indicates that tensile loading led to breakaway of the distal crown wall. In fact, with a thickness of at least 0.5 mm the frame of the distal crown met the manufacturer's requirements; despite this, it could not withstand the extraordinary tensile stresses resulting from a cantilever extension.

This estimate is confirmed by the fracture-load values for group i-c-R and c-c-R. Although the lower part of the distal crown wall was clearly reinforced, the upper part and, especially, the frame thickness in the area of tensile loading were kept unchanged. Consequently, this frame configuration did not result in significantly higher fracture-load values than the corresponding FDPs without an additional shoulder.

The FDPs in group i-i had a minor cross-section of 9 mm^2 in the connecting area compared with 12 mm^2 in groups i-c and c-c. The fracture mode in the i-i group indicates that this large difference had no effect on fracture-load values, because all the FDPs of group i-i fractured at the transition from the inlay-base to the proximal box. The cross-section at this part of the framework, 1 mm inside the tooth, was only 2 mm^2 (Figure 4a). Because of the inlay-preparation design, reinforcement of this part of the framework is infeasible and thus inlay retained all-ceramic zirconium-based cantilever FDPs cannot be recommended for clinical investigation.

It has often been claimed that appropriate crown-preparation design is necessary if retention and resistance of cantilever FDPs are to be maximized [3,4]. It has been claimed that the tooth furthest from the pontic, especially, should be extremely retentive to resist vertical dislodgement [3]. Use of a less invasive inlay preparation for the mesial abutment tooth in this study led to promising results. Despite the lifting tendency on the furthest retainer [25] imparted by the apical directed forces on the cantilever, none of the FDPs became unbonded after thermal cycling or mechanical loading. These findings, however, were from an *in vitro* test on metal abutment teeth and must be verified for natural human teeth. The loading procedure, with only axial forces, was not, moreover, a realistic scenario. Non-axial loadings and the resulting deflections of the teeth may lead to greater stresses at the interfaces between tooth-cement and FDP, and thus an increased risk of cement failure [20].

It must be noted that the results of this study were limited by the sample size of the groups. The results were also obtained in an *in vitro* test and the only loading direction simulated was vertical. Lateral forces which also cause stress on restorations are very important for their clinical success and should be studied further. Clinical conditions might be less favorable; for example, preparation of the abutment teeth or dryness during adhesive cementation, which may affect long-term clinical success. Clinical factors, for example antagonist teeth, periodontal bone loss, crown-root ratio, vitality and inclination of abutment teeth, and masticatory forces or shock absorbance characteristics, may also affect the success of dental restorations. Further investigation of all-ceramic zirconia-based cantilever FDPs is therefore required before recommendation for clinical use.

Within the limitations of this study, the results indicate that all-ceramic cantilever FDPs with a Y-TZP framework replacing one premolar resulted in acceptable fracture-load values if the final abutment tooth, at least, received a crown preparation. The fracture mode indicates that reinforcement of the distal crown wall might enhance fracture resistance.

In conclusion, inlay–inlay retained cantilever FDPs cannot be recommended, whereas the results for crown-retained all-ceramic cantilever FDPs encourage further clinical investigation.

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