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ORIGINAL ARTICLE

Fracture resistance of the veneering on inlay-retained zirconia ceramic fixed partial dentures

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Abstract

Aims. The aim of this *in vitro* study was to evaluate the fracture load of zircon frames veneered with a polymer glass holding box inlay-retained fixed partial dentures (FPDs). The influence of the position of the frame and the span length was tested. Additionally, the fracture load values of zircon frames veneered with a press ceramic were evaluated. ***Material and methods.*** Box inlay cavities were prepared on mandibular molars and premolars. Forty-eight FPDs were manufactured using industrially prefabricated zircon frames veneered with the polymer glass Artglass®. Sixteen FPDs received individually manufactured CAD/CAM zircon frames veneered with a press ceramic. All FPDs underwent thermal cycling and mechanical loading (ML). The load to fracture was measured and fracture sites were evaluated. ***Results.*** Four polymer veneered FPDs showed fractures in the veneering material after ML. The mean fracture resistance ranged from 531 N to 727 N. No significant influence of frame localization could be observed. Significantly greater fracture resistance values were found in the ceramic veneered FPDs (1276 N to 1413 N). There was no significant effect of span length in the polymer veneered group or in the all-ceramic group, with the exception of a significant peak in fracture load value for intermediate span lengths in the polymer group with a localized occlusal zircon frame. ***Conclusions.*** Polymer veneered FPDs with Y-TZP frames showed acceptable fracture resistance values, but they cannot yet be unreservedly recommended for clinical use. Fracture values for CAD/CAM manufactured Y-TZP frames combined with a press ceramic deserve further clinical investigation.

Key Words: *All-ceramic, fpds, inlay, polymer glass*

Introduction

Metal inlay-retained fixed partial dentures (FPDs) seem to be clinically successful [1], but the visibility of metal-retainer and the change in natural tooth translucency are esthetically unfavorable. The desire for natural looking restorations has encouraged research on metal-free, tooth-colored materials for inlay-retained FPDs. Promising *in vitro* [2] and *in vivo* [3] results have been reported for fiber-reinforced composites (FRCs). However, some authors do not recommend inlay-retained FRC-FPDs for permanent restorations [4,5] because of their unstable esthetics, their increased wear [6], and their liability to fiber exposure accompanied by increased plaque accumulation [7].

Another metal-free alternative may be the use of all-ceramic inlay-retained FPDs, but the brittleness of ceramic has long provided problems in their development [8]. All-ceramic inlay-retained FPDs of the early generation often failed to withstand posterior mastication forces and their indications were limited by the special mechanical properties of the material [9].

With the introduction of dense sintered yttrium-tetragonal zircon polycrystal (Y-TZP) and the ability of Y-TZP to transform, thus preventing crack propagation [10], the production of stable inlay anchored FPDs has become possible. However, zircon-based materials have an esthetic disadvantage 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.

Furthermore, with the introduction of the polymer glasses, it seems possible to eliminate the previously described disadvantages of composites and to exploit their advantages, including the simple laboratory

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procedure, the lower costs, and the possibility of repair. Additionally, the polymer glasses have produced promising *in vitro* results with respect to color change [11], wear [12], and fracture resistance [13].

In the present study, the hypothesis was tested that combinations of polymer glass/Y-TZP and Y-TZP/experimental press ceramic would be strong enough to withstand normal occlusal forces (500 N [14]) when designed as box inlay FPDs. Furthermore, the influence of span length in both groups and the influence of the position of the Y-TZP frame in the polymer-veneered group were evaluated.

Material and methods

Inlay-retained FPDs were made of an industrial prefabricated Y-TZP frame veneered with the polymer glass Artglass® (group P); all-ceramic inlay-retained FPDs were made of an individually manufactured CAD/CAM Y-TZP frame veneered with an experimental press ceramic (group A).

Preparation of the abutments

Defined box inlay cavities were made on a 2nd mandibular molar and a 1st premolar of the Frasaco study model (Frasaco, Tettnang, Germany). The cavities extended 2 mm in the occluso-cervical direction, 2 mm for the premolar and thus 3 mm for the molar in the mesio/distal to central direction and 2 mm for the premolar and thus 3 mm for the molar 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/b).

To ensure standardization of the preparation, the tooth preparations were made with diamonds with a 3-degree angle of incidence (Komet, Lemgo, Germany) using a parallelometer.

Preparation of the models

The prepared teeth were duplicated (Alpa Sil, Alpina, Unterhaching, Germany) and then 64 identical abutment teeth were cast out of the Co-Cr alloy Remanium 2000® (Dentaurum, Ispringen, Germany). To simulate physiological tooth mobility, the metal teeth were covered with heat shrink tubing (CPX 55; Poliolefin, Kentec, Waldshut, Germany).

One molar and one premolar were embedded in PMMA-resin (Palapress®; Heraeus Kulzer, Hanau, Germany) using a preoperatively prepared polysiloxane clef for exact localization. The distance between the abutments was set at 7 mm to represent the loss of a premolar (only applicable to group P), at 12 mm to represent the loss of a molar, and at 19 mm to represent the loss of a premolar + molar (Table I).

a

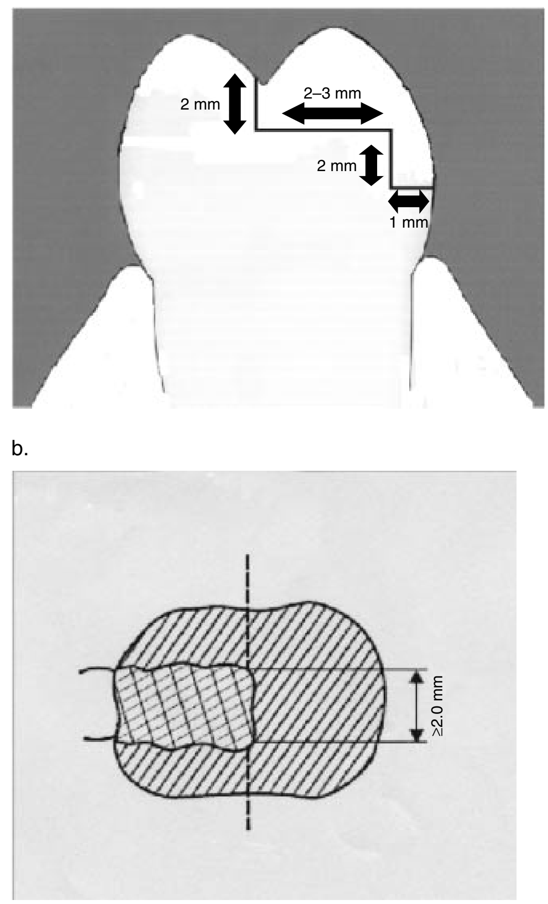

Figure 1. Dimensions of the box-inlay preparation.

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

(a) Preparation of the 48 polymer-veneered FPDs. For preparation of group P, an industrially wrought zircon

Table I. Fabrication of the FPDs

	Group P box inlay cavities						Group A box inlay cavities	
Span length	7 mm		12 mm		19 mm		12 mm	19 mm
Frame localization	O	P	O	P	O	P		
Group	PO-7	PP-7	PO-12	PP-12	PO-19	PP-19	A-12	A-19
n	8	8	8	8	8	8	8	8

O = frame localization in the occlusal box of the cavity; P = frame localization in the proximal box of the cavity; N = number of bridges.

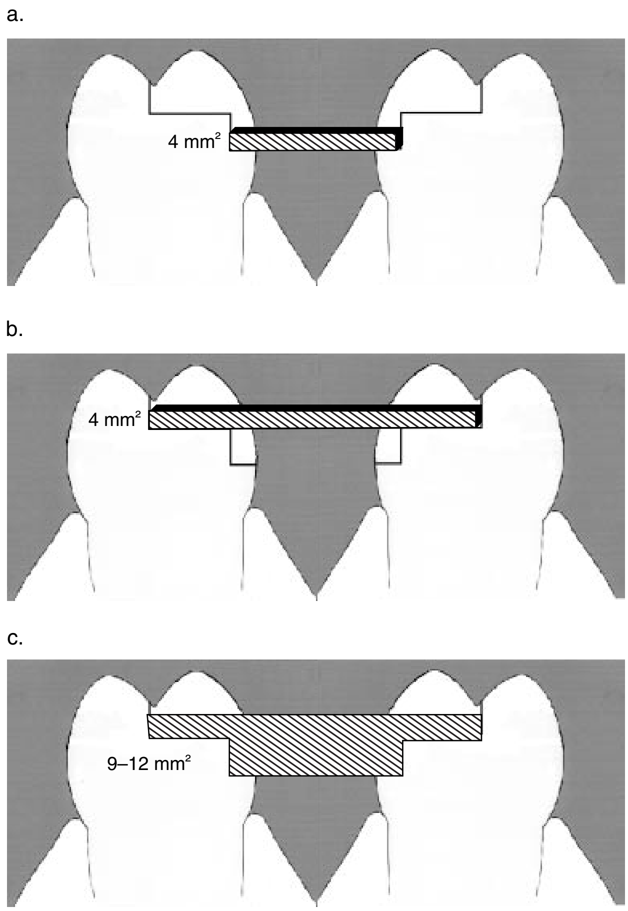

Figure 2. Frame support situation. (a) Frame adjustment in the proximal box of inlay cavity (group PP). (b) Frame adjustment in the occlusal box of inlay cavity (group PO). (c) Individual frame adjustment in the all-ceramic group (group A).

frame (Y-TZP) was adjusted to the abutments, preserving minimum dimensions of 2 mm in height and 2 mm in width. In half of the FPDs, the frame was adjusted to the occlusal box (group PO), while in the remaining FPDs the frame was adjusted to the proximal box (group PP) (Figure 2a/b). In the subgroups, the distance between the abutments was set at 7 mm (groups PO-7 and PP-7), 12 mm (groups PO-12 and PP-12), and 19 mm (groups PO-19 and PP-19).

After coating the working dies with Palaferm® (Heraeus Kulzer, Hanau, Germany), the frames were sandblasted with 50 µm Al_2O_3 at 2 bar.

A polymer single thickness film was initially applied to the working dies to fix the zircon frame. Then the frames were veneered in several steps to reduce polymerization shrinkage. Each step involved polymerization for 90 s in a UniXS light polymerization unit (Heraeus Kulzer), followed by final polymerization of 180 s.

The FPDs were finished and polished with the Artglass toolkit® (Heraeus Kulzer) at low speed, the minimum thickness of veneering polymer material being maintained at 1 mm using a preoperatively prepared polysiloxane clef. After removal of the FPDs, the inner surface was sandblasted with 50 µm Al_2O_3 prior to cementation.

(b) Preparation of 16 all-ceramic FPDs. The frameworks were prepared from pre-fabricated ceramic blanks made of 3 mol% yttrium-stabilized zirconium using a DCS milling machine (DCS Dental AG, Allschwil, Switzerland). The connectors had a minimal cross-section of 9 mm^2 and maximal cross-section of 12 mm^2; its occluso-gingival height was maximally 3.8 mm and its bucco-lingual width was maximally 3.6 mm. In half of the FPDs, the distance was set at 12 mm (group A-12) and in the other half at 19 mm (group A-19).

Before veneering the TZP framework, the frames were sandblasted with 110 µm at 2.5 bar. The TZP frames were veneered with one type of an experimental fluorapatite glass ceramic (IPS e.max ZirPress®; Ivoclar Vivadent, Ellwangen, Germany) [15]. The protocol included liner firing, wash firing, main firing, and polishing firing. The temperature and pressure gradient was used in accordance with the manufacturer's protocol (Ivoclar Vivadent).

After finishing, the surfaces of the all-ceramic FPDs were not manipulated, except for sandblasting of the inner surface with 50 µm Al_2O_3 prior to cementation.

Cementation

All metal abutments were cleaned with 90% ethanol, tribochemically silico-coated (Rocatec®; 3M Espe, Seefeld, Germany) and additionally silanized (Monobond S®; Ivoclar Vivadent, Ellwangen, Germany). All FPDs were seated with dual-cured resin composite (Variolink®, Ivoclar Vivadent) in accordance with the manufacturer's instructions.

Cyclic mechanical loading and thermal stress

All FPDs underwent 10 000 thermal cycles between 6.5° and 60°C (TC) and 600 000 cycles of mechanical loading (ML) with 50 N at the center of the pontic (frequency: 1.66 Hz).

Fracture strength

The fracture strength was determined by ML to failure in a universal testing machine (1445; Zwick Inc, Ulm, Germany). The force was applied at the center of the pontics (cross-head speed 0.5 mm/min, fracture threshold for shut-off 5 N) with a 0.5 mm tin foil between the loading element and the pontic to avoid local force peaks (Figure 3).

Fracture strength was determined by the first fracture site, evident as a peak in the protocol, defined by the fracture threshold for shut-off. The load was measured with a load cell (type U 2A) including strain gauges and was recorded with a Zwick pc-software system.

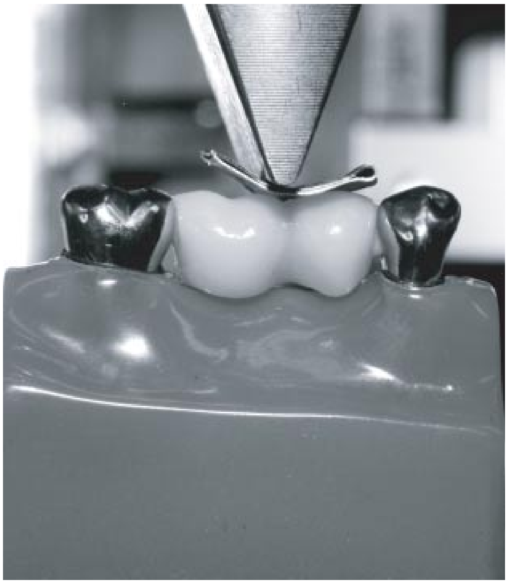

Figure 3. Loading procedure.

Statistics

Statistical analysis was performed using SPSS (version 10,07S; Ill., USA). Fracture resistance data were analysed and depicted with whisker and box plots. The box represents the 25th and 75th percentiles. Whiskers are drawn to 1.5 × interquartile range beyond the 25th and 75th percentiles. Values outside 1.5 widths or outside 3 widths of the box are marked as outliers ("o") or as extremes ("x"), respectively. The bar represents the median.

To determine the effects of differences between the groups, Kruskal-Wallis followed by Mann-Whitney U tests were performed.

Results

None of the FPDs debonded after thermal cycling (TC) or ML. Fractures in the veneering material could be observed in four polymer-veneered FPDs of group A after ML (one of each of groups PP-7, PO-12, PP-12, and PO-19). The fracture load values of those FPDs were included in the further analysis and the value was therefore set at 50.00 N corresponding to the load value during ML.

Fracture mode

No frame fracture was observed in either group examined, and none of the FPDs debonded. In most of the fractures in group P, the fracture lines ran buccal or basal to the pontic, except in the PO-7 group; here and in group A the fracture lines run beveled outwards, starting from the load locus. Detailed fracture modes are given in Table II and Figures 4a–c and 5a/b.

FPDs of group PO-7 showed a mean fracture load value of 531 N (standard deviation (SD): 137 N); FPDs of the PO-12 group showed a mean value of 728 N (SD: 284 N); for FPDs of the PO-19 group the mean fracture value was 451 N (SD: 185 N). The mean fracture load values of group PP ranged from 614 N (SD: 289 N) for FPDs of the group PP-7 over 626 N (SD: 360 N) for FPDs of the group PP-12 to 598 N (SD: 149 N) for FPDs of the group PP-19. Results are given in Figure 6.

Different frame localization in group P had no significant influence on fracture load values. However, the mean fracture load values of group A ranged from 1414 N (SD: 327 N) for FPDs of group A-12 to 1276 N (SD: 350 N) for FPDs of group A-19, and were significantly higher than values of the corresponding FPDs in group P ($p=0.001$ for group PO-12, $p=0.002$ for group PP-12, and $p<0.001$ for group PO-19 as well as for group PP-19). The fracture load values did not decrease significantly in group A with increasing span length.

No significant influence of the span length could be detected in group PP, but the values of group PO increased significantly when the 7 mm span length was replaced by a 12 mm span length ($p=0.021$), and decreased significantly if the 12 mm span length was replaced by a 19 mm span length ($p=0.007$).

Table II. Fracture mode of polymer-veneered FPDs

Span length/localization of the frame	Fracture of the frame	Fracture of veneering material buccal/basal	Fracture of veneering material occlusal	n
Group PP-7	0	7	0	7 one fracture after ML
Group PO-7	0	1	7	8
Group PP-12	0	6	1	7 one fracture after ML
Group PO-12	0	6	1	7 one fracture after ML
Group PP-19	0	8	0	8
Group PO-19	0	6	1	7 one fracture after ML

N = Number of fixed partial dentures; ML = mechanical loading.

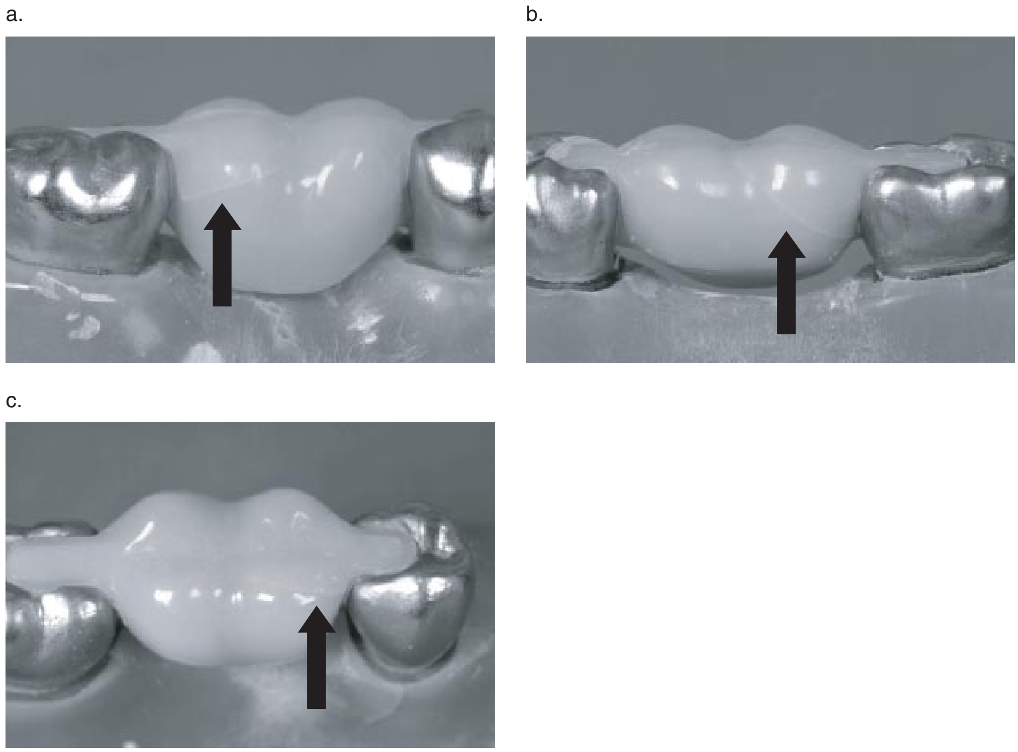

Figure 4. Fracture modus of the Artglass FPDs. Fracture line buccal. Fracture line basal. Fracture line occlusal for the premolar.

Discussion

If inlay-retained FPDs are to be used as an alternative to conventional FPDs, the inlay FPDs have to fulfill one major criterion: the material has to withstand posterior mastication forces of about 500 N [14]. On the basis of the results of the present study, polymer-veneered FPDs may be recommended for clinical use as short span FPDs. However, some critical aspects must be considered when predicting the clinical performance of these inlay-retained FPDs.

First, although a three-point bending test was applied to determine the modulus of rupture [16], the values and directions of the loads in the actual oral environment may be less favorable [17]. Secondly, supporting the restorations by metal abutment teeth to avoid variations from different dimensions, or the different elastic properties of natural teeth, might influence the fracture load values [18]. However, the fracture load values of the polymer-veneered FPDs did not attain the required 500 N for all FPD designs. In particular, FPDs of group PO-19 displayed a mean value of about 50 N below the required 500 N. Additionally, the results of the present study exhibited great variability, also influenced by the premature fracture of the veneering material after TCML.

On the one hand, the reason for the low fracture load values might be the material itself. In a study using acoustic emission signals to detect the initial fracture, Ozcan et al. [19] observed that a large amount of resin composite at the connector area may decrease strength and indicated that transmittance of the force was more

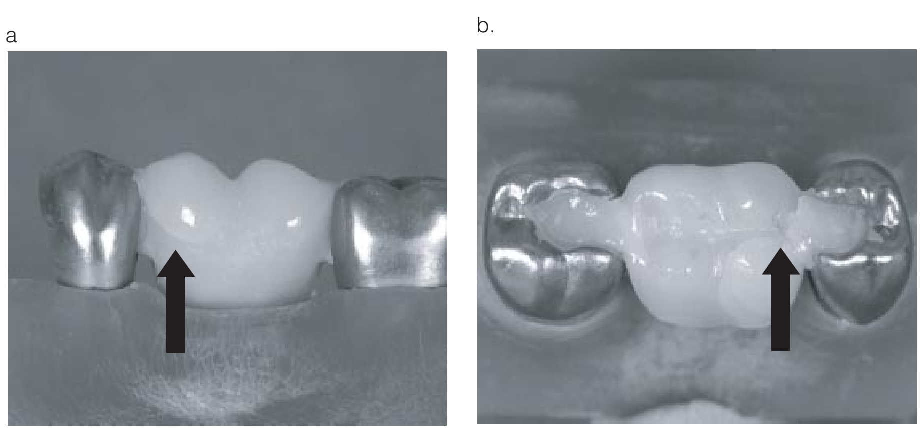

Figure 5. Fracture modus of the all ceramic FPDs. (a) Frontal view, (b) occlusal view.

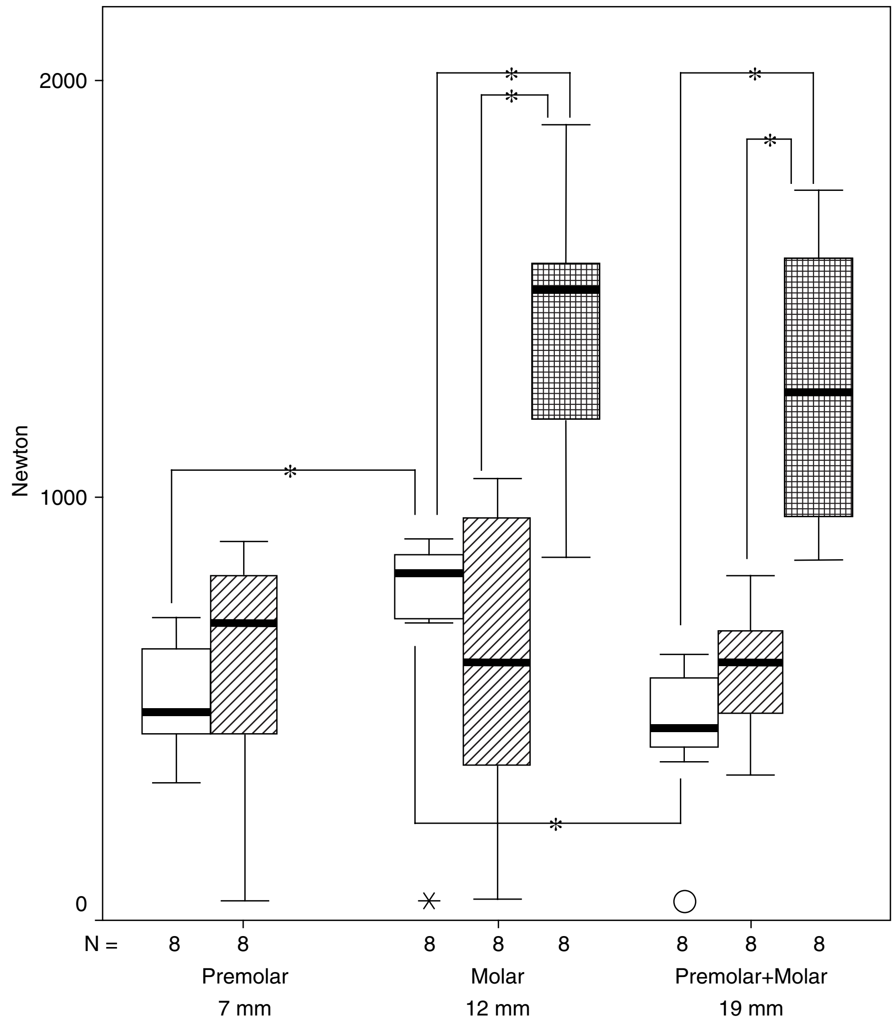

Figure 6. Fracture resistance values of all fixed partial dentures. Localization □ occlusal box, ▨ proximal box, ⊞ all ceramic, * Statistically significant.

even in the frame if the amount of veneering material was small. In the present study, the amount of veneering material varies at the loading, respectively tensile side, because of different frame localization. However, this differentiation shows no significant influence on fracture load values.

On the other hand, the reason for the low fracture load values, especially the premature failures after TCML, might be the TC. Thermo-cycling in water may affect microcracks at the interface between the filler and the matrix and induce surface stresses [20], resulting in veneering delamination.

Kawano et al. [21] showed that TC did not reduce the hardness of the material Artglass®, but did reduce the flexural strength. In the three-point bending test that we used, tensile stress will develop at the gingival surface of the pontic [22], corresponding to the most frequently observed fracture mode in the present study. Another explanation might be that the zircon-composite compound was the weak point.

In the present study, the frames were pretreated with airborne particle abrasion, the preferred surface treatment for high strength ceramic materials [23]. However, Edelhoff et al. [16] have shown that an increase in the ceramic-composite bond strength can be observed when tribochemically conditioning and silanization are used. This kind of pretreatment was used for the abutment composite compound in the present study and none of the FPDs debonded.

Furthermore, internal flaws such as porosity or undesirable microstructural features may be an explanation for the greater variation in fracture load values. Anyway, this risk applies to all manually and computer manufactured FPDs [24]. The use of industrially manufactured frames reduces the risk of manufacturing faults, but each industrially produced frame has to be adapted to the abutment tooth and to the occlusion. Consequently, machining these frames cannot be avoided and therefore faults are omnipresent.

In contrast to the results of the present study, Rosentritt et al. [25] found a median fracture value of about 900 N for zircon frames veneered with the composite material Sinfony. Although a direct comparison of the studies is not possible, because of different study conditions and different materials, those values were consistent with clinical application.

However, it has to be considered that the span length was less and connector size was higher than in our study, which also may have influenced fracture load values. In the present study, the connecting area of the zircon frame was reduced to a minimum (4 mm^2) in order to test this reduced connector design with a view to extending the strict patient selection for fabrication of a metal-free FPD [26], of preserving the esthetics, and of supporting interdental cleaning to maintain periodontal health [27].

The fracture load values for all-ceramic FPDs in the present study exceeded the maximum expected mastication forces. Available results from other comparable all-ceramic inlay-retained FPDs showed fracture resistance values of about 900 N [28]. Although a direct comparison of the fracture load values is not possible because of variations in study design (the span length was greater and the cross-section of the connecting area smaller in the present study), the fracture load values of the present study were higher (1270 N).

The differences in the fracture load values of all-ceramic and polymer-veneered FPDs in the present study may be attributed on the one hand to the dissimilar connector size [29] and on the other to the dissimilar elastic properties of these two materials. Furthermore, adhesive failure of veneer and core ceramic does not occur in the presence of a good bond between compatible ceramic core and veneer material [30] because of the possibility of elements unique to the core. Additionally, it is possible that veneering glass diffused across the interface or a layer of excess infiltration glass developed on the core surface [31].

It has to be noted that the results of the present study were limited by the size of the groups. Additionally, the results were found in an *in vitro* test. The clinical conditions might be less favorable; for example, the preparation of the abutment teeth or the dryness during adhesive cementation, which may have an influence on long term clinical success. Additionally, clinical factors, such as antagonist teeth, masticatory forces, or shock absorbance characteristics, influence the success of dental restorations. Thus, further investigations for polymer/zirconium inlay-retained FPDs are required before an unrestricted clinical investigation can be recommended.

In conclusion, within the limitations of the present study, the results indicate that polymer FPDs with a Y-TZP frame showed acceptable fracture load values for FPDs replacing a premolar or a molar, but not for greater span lengths. The results of all-ceramic FPDs encourage further clinical investigation for FPDs up to a span length of 19 mm.

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