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Fracture resistance of different zirconium dioxide three-unit all-ceramic fixed partial dentures

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Abstract

Objective. This study evaluated the fracture resistance of different zirconia three-unit posterior all-ceramic fixed partial dentures before and after fatigue loading. ***Material and methods.*** Forty-eight zirconia three-unit posterior all-ceramic fixed partial dentures were fabricated using different manufacturing systems and conventionally cemented on human teeth. The restorations were divided according to the system used for manufacturing the frameworks into 3 groups of 16 specimens each (DCS, Procera and Vita CerecInlab). Half of each group was artificially aged through dynamic loading and thermal cycling, whereas the other half was left with no artificial aging. Afterwards, all specimens were tested for fracture resistance using compressive load on the occlusal surface. Non-parametric ANOVA using the Kruskal-Wallis and Wilcoxon rank tests was performed to test for differences in fracture resistance values with a global significance level of 0.05. ***Results.*** All artificially aged specimens survived with no failures. The median fracture resistance values (before; after artificial aging) were: DCS (1683 N; 1618 N), Procera (1522 N; 1256 N), and Vita CerecInlab (1702 N; 1556 N). No significant differences were found for comparisons between different groups before artificial aging. Artificial aging did not significantly influence the fracture resistance of different groups. After artificial aging, group Procera showed significantly lower fracture resistance than the Vita CerecInlab ($p = 0.015$) and DCS ($p = 0.038$) groups. ***Conclusions.*** All tested restorations have the potential to withstand occlusal forces applied in the posterior region and can therefore represent interesting alternatives for replacing porcelain-fused-to-metal restorations. Further assessment is needed before recommending such restorations in daily practice.

Key Words: *All-ceramic, fatigue loading, fixed partial denture, fracture resistance, zirconia*

Introduction

The porcelain-fused-to-metal (PFM) technique has been extensively used in the fabrication of fixed partial dentures (FPDs) [1]. The familiar fabrication techniques and the acceptable esthetic outcome have rendered PFM the most popular treatment option for the majority of dentists [2]. Clinical studies on FPDs report survival rates between 72% and 87% after 10 years, between 69% and 72% after 15 years, and 53% after 30 years [1–3]. However, gingival discoloration and allergic contact gingivostomatitis from gold alloys have been reported with PFM restorations [4–6]. Such adverse effects have accelerated the development of alternatives to metal-based dental restorations. The introduction of all-ceramic materials enabled the fabrication of restorations with optimal esthetics and characteristics, such as color stability, high wear resistance, and low thermal conductivity [7,8]. All-ceramic FPDs fabricated using different ceramic core systems have been evaluated in a limited number of clinical studies, demonstrating that the success rate of posterior all-ceramic FPDs is between 73.4% and 93.3% after observation periods of between 3 and 5 years [8–11]. However, low fracture resistance and relatively low flexural strength have limited the possibility of manufacturing posterior all-ceramic FPDs using such core systems [12]. Today, zirconium-dioxide-based ceramics, specifically yttrium tetragonal zirconia polycrystal ceramics (Y-TZP), are being introduced into dentistry because new computer-aided manufacturing technologies (CAM) have made an application possible [13]. With a

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flexural strength of 900 to Pa, a fracture toughness of 10 $MPa/m^{0.5}$ and a modulus of elasticity value of 210 GPa, this oxide ceramic exhibits better mechanical performance and superior strength and fracture resistance than other core ceramics [14,15]. The high strength of the material can be explained by a phase transformation toughening mechanism. This involves the transformation of metastable tetragonal grains (t) to the monoclinic phase (m) under stress at the crack tip, which, accompanied by volume expansion, induces compressive stresses that counteract crack propagation [14]. Y-TZP ceramic is being used today as a material to fabricate root canal posts, frameworks for crowns and FPDs, implant abutments, and implants [16–19]. A limited number of short-term clinical studies on posterior Y-TZP-based FPDs have shown a success rate of 100% with observation periods between 2 and 3 years [20,21]. Y-TZP ceramic has been reported to be prone to aging, however [22–24]. Spontaneous or slow $t \rightarrow m$ transformation, called low-temperature degradation (LTD), can cause mechanical property degradation [14,24,25]. This transformation is affected by temperature, vapor, grain size, micro and macro cracking of the material, concentration of stabilizing oxides, and fabrication and veneering techniques [14,23,24,26,27].

The fabrication of Y-TZP frameworks can be carried out using two major techniques. The first includes milling enlarged frameworks out of homogenous blanks, which are usually delivered in a non-sintered (green body) or in different pre-sintered stages. The milled frameworks are then sintered and shrunk to the desired dimensions [12,28]. The second technique employs milling the frameworks directly with the final dimensions out of highly dense sintered prefabricated blanks [12,20]. With such variations in fabrication and processing techniques of Y-TZP frameworks, differences in the LTD resistance and the long-term stability of the restoration can be expected [24].

The different fabrication techniques of Y-TZP frameworks have not yet been compared for their effect on the fracture resistance and long-term stability of Y-TZP-based FPDs. It can be hypothesized that these techniques produce frameworks with different aging stability and fracture resistance properties. The purpose of this *in vitro* study was to evaluate the fracture resistance of zirconia three-unit posterior all-ceramic FPDs fabricated with three different processing techniques before and after fatigue loading.

Material and methods

Forty-eight caries-free human mandibular premolars and 48 molars were selected, cleaned by scaling, and stored in 0.1% thymol solution at room temperature throughout the study. Pairs of premolar and molar teeth were randomly assigned into three test groups (DCS, Procera, and Vita CerecInlab) of 16 samples each. A representative clinical model with a missing first molar was then selected. The mesiodistal width of the pontic was 11 mm. In order to imitate physiological tooth mobility and to conform to the biological width, all roots of the abutment teeth were covered 2 mm apically from the cemento-enamel junction (CEJ) with an artificial periodontal membrane made out of a gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar GmbH, Hilden, Germany). The teeth were then fixed into the sample-holder using an autopolymerizing acrylic resin (Technovit 4000; Wehrheim, Germany). The resin has a modulus of elasticity of approximately 12 GPa, which approximates that of human bone (18 GPa). The abutment teeth were prepared using diamond burs (No 386.023, 8368.023, 837KR.012, 8837KR.012; Gebr. Brasseler, Lemgo, Germany) with water spray application. A 1.2-mm deep circumferential chamfer preparation with a reduction of 1.5 mm for the occlusal surface and 1.2 mm for the axial surfaces was made. The preparation's convergence angle was 6°. Because of different clinical crown heights, the heights of the prepared premolars and molars were 6 mm and 5 mm, respectively. This difference was later adjusted through the veneering process to make sure that the restoration parts were at the same horizontal level. All transitions from the axial to the occlusal surfaces were rounded and homogeneous and smooth surfaces were achieved (Figure 1). Afterwards, impressions of the prepared abutments were taken using the Putty-Wash-Technique on custom-made plastic trays. Light-body silicone impression material (Dimension Garant L; 3M-Espe, Seefeld, Germany) was syringed around the abutments, and putty material (Firmer Set Putty; 3M-Espe, Seefeld, Germany) was used on the tray. Four hours after removing the impression from the abutment teeth, a die was poured using a type 4 dental stone (Fujirock type 4; GC, Tokyo, Japan) for the DCS and Procera groups and a scanable stone (Dentona esthetic-base gold; Dentona, Detmold, Germany) for group Vita CerecInlab. After a setting time of at least 45 min, the die was removed from the impression and a total of 16 frameworks were fabricated for each test group. The frameworks of all groups had a connector height of 3 mm, a connector width of 3 mm, an occlusal thickness of 0.7 mm, and a wall thickness of 0.5 mm (Figure 2).

Fabrication of DCS frameworks

Stone dies of the DCS group were scanned using a laser scanner (Preciscan; DCS Precident, Allschwill, Switzerland). The collected data were then digitized and transmitted to a computer, where the framework was designed and calculated using special software (DCS Dentform; DCS Precident, Allschwill, Swit-

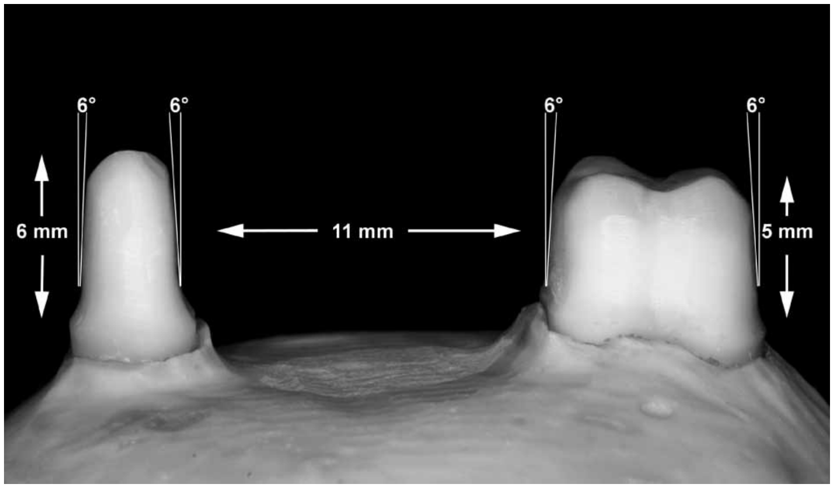

Figure 1. Dimensions of the test model used.

zerland). The data were subsequently forwarded to the machining center (Precimill, DCS Precident, Allschwill, Switzerland), where the framework was milled out of a fully sintered Y-TZP blank (DC-Zirkon; DCS Precident, Allschwill, Switzerland). This highly dense blank does not need any further heat treatment and is manufactured by sintering and hot isostatic pressing (HIP) of Y-TZP material. Afterwards, the framework was removed from the milled blank and further manual precision adjustments using diamonds (FG Diabolo burs; Bredent, Senden, Germany) were made.

Fabrication of Procera frameworks

Stone dies of the Procera group were scanned using a tactile scanner (Procera Forte; Nobel Biocare, Göteborg, Sweden). The data were transferred to a computer and the design of the framework (CAD) took place using the Procera software. The information was then compressed and transferred via modem line to the production unit (Procera; Nobel Biocare, Göteborg, Sweden), which used the information to calculate the anticipated shrinkage and fabricate an enlarged die. Non-sintered Y-TZP ceramic (green stage), which is low in density and did not undergo previous heat treatment, was then dry pressed against the enlarged die, and the temperature was raised to a level similar to the pre-sintering stage. At this point, the enlarged and porous framework is stable. The outer surfaces of the framework were then milled to the desired shape. After removing from the enlarged die, the milled framework was inserted into the furnace and fired to full sintering. During this cycle, the framework shrunk to fit the dimensions of the original working die.

Fabrication of Vita frameworks

Dies of the Vita CerecInlab group were scanned using the Cerec Inlab scanner (Sirona, Bensheim, Germany). The data were then digitized and transmitted to a computer where the framework was designed and calculated. Pre-sintered Y-TZP blocks (Vita In Ceram 2000 YZ-Cubes; Vita Zahnfabrik, Bad Säckingen, Germany) were milled in the CAM unit (Cerec Inlab, Sirona, Bensheim, Germany). The unit mills an enlarged framework out of the pre-sintered block to compensate for later sintering shrinkage. Afterwards, the milled framework was

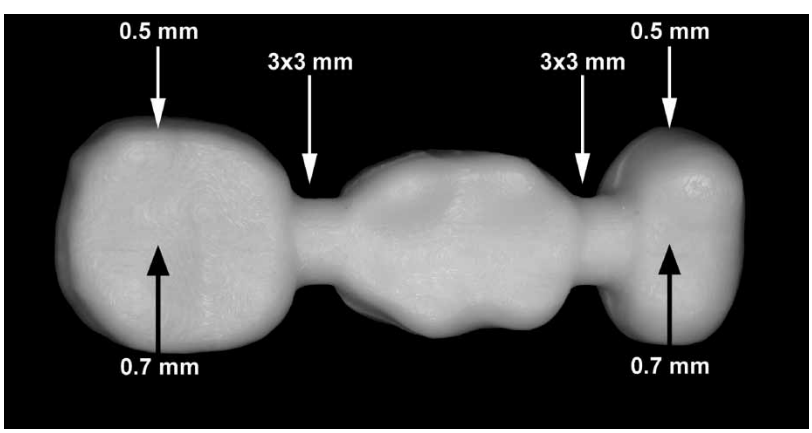

Figure 2. A representative Y-TZP framework before veneering (Vita CerecInlab group). The dimensions of the connectors and the thickness of axial surfaces are referred with white arrows. The thickness of the occlusal surfaces is referred with black arrows.

carefully removed from the machine and separated from the block-holder at the milled side using diamond cutting instruments. The framework was then post-sintered using a high temperature furnace (Vita ZYrcomat; Vita Zahnfabrik, Bad Säckingen, Germany). In accordance with the manufacturer's recommendations, the temperature in the firing chamber did not exceed 1600°C. Subsequent fitting adjustments after sintering were made using rotary diamond instruments with water cooling (FG-Diabolo burs; Bredent, Senden, Germany).

After try on and fitting, frameworks of all groups were veneered using silicate-based veneering ceramic (e.max Ceram; Ivoclar-Vivadent, Schaan, Liechtenstein). The parameters used for the veneering process are listed in Table I. Then, all FPDs were definitively placed on the prepared teeth using glass ionomer cement (Ketac Cem; GC, Tokyo, Japan) with finger pressure for 3 min.

Eight of 16 specimens in each group were exposed to 1,200,000 cycles of thermomechanical fatigue in a computer-controlled dual-axis artificial oral environment (Willytech, Munich, Germany) to simulate 5 years of clinical function. Using a 6-mm diameter ceramic ball (Steatite; Hoechst Ceram Tec, Wunsiedel, Germany), a load was applied in the center of the occlusal surface of the FPD pontics at a frequency of 1.6 Hz. The ceramic ball has a Vicker's hardness similar to that of enamel. A loading force of 49 N was chosen to simulate a load within the clinical range. During testing, all specimens were subjected to simultaneous thermal cycling of between 5° and 55°C for 60 s each, with a dwell time of 12 s, maintained by a thermostatically controlled liquid circulator (Haake, Karlsruhe, Germany). Finally, all specimens were loaded compressively in a universal testing machine (Z010/TN2S; Zwick, Ulm, Germany) with force application perpendicular to the occlusal surface and a cross-head speed of 2 mm/min. The loading stamp was centrally positioned over the occlusal surface of the first molar (pontic). One-millimeter-thick tin foil (Dentaurum, Ispringen, Germany) was placed between the loading stamp and pontic to achieve homogenous stress distribution. The load required to fracture the specimen was recorded using x-t recording software (Zwick testXpert V7.1; Zwick, Ulm, Germany).

Results of the load-to-fracture test were recorded with the help of box plots. Non-parametric ANOVA using the Kruskal-Wallis and Wilcoxon rank tests were performed to test for differences in fracture resistance values with a global significance level of 0.05.

Results

All specimens subjected to aging survived 1,200,000 cycles of dynamic loading and thermal cycling in the artificial oral environment. No chipping of the veneering ceramic or decementation of the restorations was recorded. The median fracture resistance values for the non-aged groups showed that the highest value occurred in group Vita CerecInlab followed by groups DCS and Procera. After artificial aging, the highest median fracture resistance value was observed in group DCS followed by groups Vita CerecInlab and Procera (Table II). The fracture resistance values before and after artificial aging are represented in the box plots (Figure 3). Compared to values at the initial stage, artificial aging reduced the fracture resistance by 5%, 13%, and 7% for groups DCS, Procera, and Vita CerecInlab, respectively. Comparisons of the fracture resistance values obtained before and after aging of each group were not statistically significant for all groups ($p > 0.05$). Similarly, no significant differences were found for the fracture resistance comparisons between different groups before artificial aging (Kruskal-Wallis test: $p = 0.3$). After artificial aging, the Kruskal-Wallis test showed a significant group effect ($p = 0.03$). The Wilcoxon rank test revealed significantly lower fracture resistance of the Procera group compared to the Vita CerecInlab (Wilcoxon test: $p = 0.015$) and DCS (Wilcoxon test: $p = 0.038$) groups. All other comparisons showed no statistically significant differences ($p > 0.05$).

The location and mode of failure of the three test groups with and without thermomechanical fatigue are listed in Table III. The fractures were located in the loading point and through one or both connectors. The fracture modes of the specimens of all test groups were similar. The failures were perpendicular to the mesial-distal axis of the restorations in a smooth curve between the loading point and the gingival side of the connector.

Discussion

The design and tests conducted in this study were chosen to better simulate clinical conditions. Extracted human teeth were used as abutments

Table I. Firing chart for the veneering ceramic

Process	Preheating (°C)	Heating rise (min)	Vacuum in (°C)	Vacuum out (°C)	Holding time (min)	High temp (°C)	Heating rate (°C/min)
Opaque	403	6	450	799	2	800	50
Dentin	403	6	450	759	2	760	50
Glaze	403	6	450	729	1	725	50

Table II. Groupwise fracture resistance in N after the load-to-fracture test

Group	Min.	1st quartile	Median	Mean	3rd quartile	Max.	SD
DCS/initial	1278	1489	1683	1659	1845	1993	245
DCS/aged	1175	1522	1618	1580	1713	1804	197
Procera/initial	1105	1278	1522	1496	1727	1800	260
Procera/aged	1044	1131	1256	1297	1348	1783	242
Vita CerecInlab/initial	1472	1659	1702	1713	1776	1946‘	142
Vita CerecInlab/aged	1394	1479	1556	1593	1663	1854	174

because their modulus of elasticity, bonding characteristics, thermal conductivity, and strength are closer to the clinical situation than those of metal, plastic, or animal teeth [29]. The abutment's mobility has been demonstrated as a decisive factor in the evaluation of dental restorations [29,30]. The analysis with immobile abutments was found to produce a higher failure load than when testing with mobile abutments, and led to overestimation of the load-bearing capacity of the restoration [28,30]. A thin layer of gum resin, as used in the present study, can simulate tooth mobility similar to physiological tooth movement and thus provide better clinically relevant data [29–31].

In this study, half of the specimens of each test group were exposed to the artificial oral environment. The main objective of this device is to introduce a comparable cyclic fatigue component. The number of specimens tested and the use of water rather than artificial saliva during testing are limitations of this study that may affect the interpretation. The parameters used for the cyclic fatigue loading in an artificial oral environment were limited to the physiological values found in the literature [32,33]. Several *in vitro* studies used a cyclic loading force of 49 N for testing the fracture resistance of dental restorations [29,31]. These studies have considered the functional forces that arise during mastication or swallowing, which usually range between 2 and 50 N [32,33]. Hence, a cyclic loading force of 49 N was applied so as to approach a clinically relevant situation.

Natural teeth and dental restorations are subjected on a daily basis to masticatory loading, which places them under repeated loading throughout their service life. Studies have shown that humans have an average of 250,000 masticatory cycles per year. Comparing different *in vitro* studies, the number of applied cycles varies according to the selected service time [27,29,31]. In the present study, 1,200,000 masticatory cycles were performed in the artificial oral environment to simulate a service time of 5 years.

It has been reported that masticatory loads range from 50 to 250 N, while parafunctional behavior such as clenching and bruxism can produce loads between 500 and 800 N [30,34–36]. In the most severe cases, the forces can exert loads of up to 1000 N [37,38]. In this study, all test groups before and after fatigue loading showed minimum fracture resistance levels of greater than 1000 N. Thus, all test specimens exceeded the limits of the fracture resistance for posterior restorations.

The fracture resistance of Y-TZP based all-ceramic FPDs was evaluated in several *in vitro* studies. These studies reported values between 800 and 3480 N

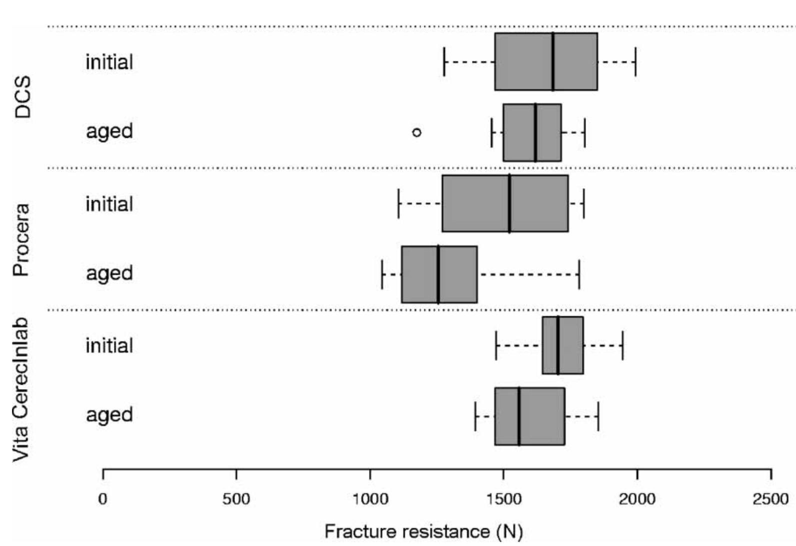

Figure 3. Box plots of the results after the load-to-fracture test of different groups in N ($n = 8$).

Table III. Location and mode of failure after the load-to-fracture test

Group	Abutment fracture	Distal connector	Mesial connector	Both connectors
DCS/initial	1	7	–	–
DCS/aged	–	4	–	4
Procera/initial	–	2	1	5
Procera/aged	–	8	–	–
Vita CereInlab/ initial	–	4	3	1
Vita CerecInlab/ aged	–	3	–	5

[23,27,39]. Different experimental design and testing parameters, such as fatigue loading, abutment material, and mobility, were used in the previous studies, so a comparison between the results of these studies and those of the present one is rather difficult.

The exposure to fatigue loading reduced the fracture resistance of all test groups. This reduction, however, was statistically not significant. Differences in the reduction rate of the fracture resistance may be explained by different fabrication techniques for each system. Milling a framework out of a pre-sintered Y-TZP blank and subsequent post-sintering will probably produce surface flaws and residual/compressive stresses different from those produced in a Y-TZP framework milled out of a fully sintered blank [22–24,40,41]. This will in turn lead to differences in the LTD resistance between different systems [24]. Other comparative studies employing surface quality and structure detecting methodologies like X-ray Diffraction (XRD), Atomic Force Microscopy (AFM), and Optical Interferometer (OI), may elucidate the effect of different fabrication techniques on the aging sensitivity of Y-TZP ceramics.

Another issue that should be addressed is the effect of the veneering ceramic on the fracture resistance of the restorations. Temperature and moisture are known as factors that enhance t-m phase transformation of Y-TZP ceramics and affect their mechanical properties [42–44]. Thus, veneering cannot be excluded as a factor that might affect the fracture resistance of a Y-TZP based restoration. In an *in vitro* study, Y-TZP specimens subjected to heat treatment or veneering exhibited significantly smaller fracture resistance than those of specimens tested as delivered after milling [23]. The authors explained the reduction in the fracture resistance by the effect of heat treatment and moisture during veneering on the relaxation of residual stresses and/or phase transformation and/or alteration of the grain size of Y-TZP ceramic. In this study, the frameworks were veneered using one type of veneering ceramic and its effect on the fracture resistance of the tested groups was not evaluated. Further studies are needed to evaluate the effect of veneering ceramics/procedures on the fracture resistance of Y-TZP frameworks.

The fracture patterns of the test specimens of all groups were located in the loading point and through one or both connectors. These patterns are similar to those reported in previous *in vitro* studies of 3-unit zirconia-based all-ceramic FPDs [23,27,28,39]. It was difficult to assess the exact origin of the fractures that occurred in the present study; whether they started at the loading point or at the connectors. However, they were perpendicular to the mesial-distal axis of the frameworks in a smooth curve between the loading point and the gingival side of the connector. Previous studies have shown that the fractures of all-ceramic FPDs are usually located in the area between the retainers and the pontics, and that the gingival side of the connectors can be the area where high tensile stresses are located [8,23,28,39,45]. In addition, the dimensions of the connector area are crucial for the resistance and longevity of all-ceramic FPDs [21]. A small irregularly designed connector area will reach the critical strain faster than a thick well-designed connector. The material's properties, anatomical limitations, hygienic reasons, and esthetic expectations are important factors that should be considered for the design of the connector area [12]. In the present study, connector dimensions of 3×3 mm were used. Several *in vitro* studies have investigated the fracture resistance of Y-TZP based all-ceramic FPDs with connector dimensions of 3×3 mm and showed good results [20,23]. In a clinical study, Y-TZP all-ceramic FPDs with connector dimensions of 4×4 mm showed a success rate of 100% after an observation period of 2 years [21]. No clinical studies are available on the clinical success of Y-TZP all-ceramic FPDs with connector dimensions of 3×3 mm. Therefore, the long-term success of these restorations should be evaluated before recommending for clinical application.

Conclusions

Within the limitations of this *in vitro* study, it can be concluded that all tested Y-TZP all-ceramic FPDs have the potential to withstand physiological occlusal forces applied in the posterior region, and can therefore be interesting alternatives replacing PFM restorations. Issues like connector shape and size, veneering ceramic, aging behavior, and long-term clinical performance need to be further assessed before Y-TZP all-ceramic FPDs can be recommended in daily practice.

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