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# Fracture resistance of monolithic and veneered all-ceramic four-unit posterior fixed dental prostheses after artificial aging

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**Abstract:** This study compared the fracture resistance of monolithic and veneered all-ceramic four-unit posterior fixed dental prostheses (FDPs) generated by computer-aided design/computer-aided manufacturing (CAD/CAM) after aging in a mastication simulator. Four-unit FDPs were designed from six different all-ceramic systems: 1) monolithic lithium disilicate (M-E), 2) monolithic zirconia (M-TZI), 3) veneered zirconia by conventional layering (V-L), 4) veneered zirconia by lithium disilicate pressing (V-P), 5) veneered zirconia by lithium disilicate fusing (CAD-F-E), and 6) veneered zirconia by feldspathic ceramic cementing (CAD-C-CB). The specimens were divided into control and aging groups ( $n = 10$  per group). The aging process included both thermocycling and mechanical loading and was followed by fracture resistance testing. All specimens in the M-E, M-TZI, and V-L groups survived; however, all specimens in the V-P group were fractured during artificial aging. The highest fracture resistance values were observed in the M-TZI group. According to the fracture resistance test, connector fractures were the most frequent type of failure. M-TZI and M-E FDPs revealed no failures during aging and showed higher

fracture resistance than the veneered groups. Among the veneered zirconia framework groups, V-L FDPs showed the highest success rate during aging, while the fracture resistance was similar among all the veneered zirconia groups.

Keywords: CAD-on; lithium disilicate; monolithic; press-on; zirconia.

## Introduction

For decades, veneered metal alloys have been the ‘gold standard’ for posterior fixed dental prostheses (FDPs), because of their good mechanical properties and high survival rates after long observation periods (1,2). Recently, all-ceramic materials with high aesthetics and good biocompatibility have gained popularity as the demand for metal-free reconstructions increases (3). With the extensive use of computer-aided design/computer-aided manufacturing (CAD/CAM) technology to fabricate ceramic restorations, various all-ceramic materials have been developed for crowns, FDPs, and frameworks (4).

Zirconia is a widely used framework material for all-ceramic FDPs because of its excellent mechanical properties and biocompatibility (5,6); however, the clinical success of zirconia-supported FDPs has been compromised by chipped veneering porcelain, especially in posterior restorations (7-10). To overcome the problem of porcelain chipping, improved veneering methods, including modified firing and over-pressing, have been developed. Recently, the “CAD-on technique” was introduced (8,11,12). In this technique, both the zirconia

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framework and the veneer (either lithium disilicate or feldspathic ceramic veneer) are fabricated using CAD/CAM technology, and the two components (zirconia framework/ lithium disilicate veneer or zirconia framework/ feldspathic ceramic veneer) are bonded with low-fusing ceramic or adhesive cementation procedures (3,13,14). This new technique may be less time-consuming and less technique-sensitive than conventional ceramic layering, and it offers more reliable restorations due to the superiority of industrially-fabricated homogenous ceramic blocks (9,15).

Alternatively, another way to prevent veneer failure is to eliminate the veneer layer and use FDPs made entirely of lithium disilicate or zirconia (16,17). Lithium disilicate combines good aesthetic and mechanical properties; thus, it is preferred for monolithic restorations in posterior areas (18-20). Additionally, fabricating monolithic restorations from pure zirconia improves the mechanical stability of the prostheses and expands the range of indications. Monolithic zirconia crowns have greater fracture resistance than crowns that include lithium disilicates (21-23).

Until now, only limited data were available regarding the fracture resistance of four-unit FDPs fabricated with these new materials and techniques. Therefore, the purpose of this study was to compare the fracture resistance of CAD/CAM-generated monolithic and veneered all-ceramic four-unit posterior FDPs after artificial aging in a mastication simulator. The hypotheses were that FDPs veneered using different techniques and monolithic FDPs would survive during aging, and all tested groups of FDPs would show equal fracture loads before and after aging.

## Materials and Methods

### Preparation and manufacture of FDPs

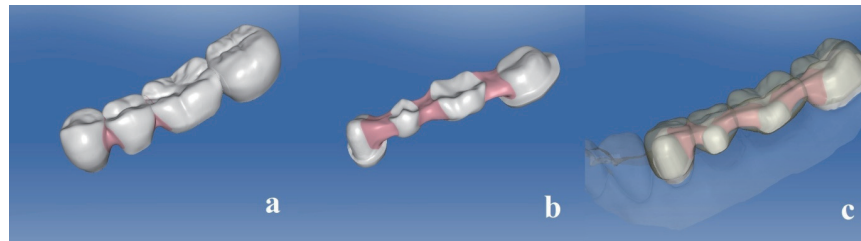
In this study, a mandibular typodont model was used (ANA-4V Advanced Standard Typodont; Frasco GmbH, Tettnang, Germany). A four-unit FDP was formed using an artificial mandibular first premolar tooth and a second molar tooth as abutment teeth in the absence of the first molar and second premolar. The tooth preparation yielded a 1.2-mm deep chamfer, a 2 mm occlusal reduction, and 6-degree angled axial walls. A round-end, tapered, diamond, regular grit bur was used to impart the 6-degree angulation by holding the bur parallel to the intended path of the insertion of the preparation. A silicone index (Optosil; Heraeus Kulzer, Hanau, Germany) was fabricated before the tooth preparation for standardizing the reduction. Sharp edges and undercuts were avoided. The prepared teeth were duplicated from a Co-Cr alloy

powder (Eos Cobalt Chrome SP2; Eos GmbH, Krailling, Germany) using direct-laser sintering technology (Eosint M 270; Eos GmbH). A total of 120 metal duplicate die pairs were fabricated for 12 groups ( $n = 10$  per group).

The prepared teeth were mounted on a Typodont model (ANA-4V Advanced Standard Typodont; Frasco GmbH). Impressions were taken with a polyvinyl siloxane (PVS) impression material, and custom trays were fabricated using visible light-polymerizing acrylic resin (Triad VLC; Dentsply Caulk, York, PA, USA). A one-step dual viscosity impression was made with light- and heavy-body PVS impression materials (Hydrosil light and heavy body; Zhermack Spa, Badia Polesine, Italy). The impressions were inspected with a magnifying glass (Loupe opt-on; Orange Dental, Biberach, Germany) at  $2.7\times$  magnification. Then, lubricated metal duplicates were placed in the impressions, and the casts were fabricated from a Type IV dental stone. After setting for 1 h, the casts were separated from the impressions, and the prepared teeth were visually inspected for irregularities (Loupe opt-on; Orange Dental) by a single operator at  $2.7\times$  magnification.

Digital impressions were obtained using an intraoral scanner (CEREC Omnicam; Sirona Dental Systems, Bensheim, Germany). To standardize the restoration dimensions, all restorations were designed in the “multi-layer” function of the CAD/CAM software (inLab SW 4.2; Sirona Dental Systems). This function was used to fabricate CAD-on restorations, and an anatomical full FDP was designed by the software. The connector dimensions of the anatomically-designed FDP were  $16\text{ mm}^2$ . Monolithic groups were milled from this design without any modifications. For the veneered zirconia groups, the anatomical design was split by the software into anatomical framework and veneer layers. After splitting, the connector dimensions of the zirconia framework were designed to be  $12\text{ mm}^2$ . Thereby, the standardized anatomical zirconia framework and veneer design could be provided for all the veneered zirconia groups. The cement thickness was  $80\text{ }\mu\text{m}$ . Manual adjustments were not made to the marginal aspects of the designs. Minor adjustments were made to the occlusal surface of the restorations to ensure similar material thicknesses for all the specimens, if necessary. The designs of the monolithic FDP, zirconia framework, and CAD-on are shown in Fig. 1. All ceramic FDPs were milled in a milling unit (CEREC MC XL; Sirona Dental Systems).

The following six groups of FDPs were processed according to the manufacturers’ recommendations: 1) monolithic lithium disilicate (M-E), 2) monolithic zirconia (M-TZI), 3) veneered zirconia by convention-



**Fig. 1** The design of the restorations: a) monolithic FDP, b) zirconia framework, and c) CAD-on.

**Table 1** Fabrication and cementation procedures

Material (Manufacturer)	Fabrication	Cementation
IPS e.max CAD (Ivoclar Vivadent, Schaan, Liechtenstein) Monolithic lithium disilicate	The milled FDPs in a crystalline intermediate phase were crystallized in a porcelain furnace (Programat P300; Ivoclar Vivadent).	Etched for 20 s with IPS Ceramic Etching Gel. Monobond applied for 60 s and dried.
Incoris TZI (Sirona Dental Systems) Monolithic zirconia	The milled FDPs in a partially sintered phase were dried in a porcelain furnace for 30 min at 80°C prior to sintering in a sintering furnace (InFire HTC; Sirona Dental Systems).	Airborne-particle abraded with 50 µm aluminum oxide particles at 200 Kpa.
IncorisZI (Sirona Dental Systems) Zirconia framework	Same as Incoris TZI.	Airborne-particle abraded with 50 µm aluminum oxide particles at 200 Kpa.
Cerec Blocks (Sirona Dental Systems) Veneer cementing on zirconia	Milled	Hydrofluoric acid applied for 60 s. Silanized for 60 s.
IPS e.max Press (Ivoclar Vivadent) Lithium disilicate pressing on zirconia	Lithium disilicate glass ceramic ingots pressed onto the zirconia frameworks.	
Vita VM9 (Vita Zahnfabrik, Bad Säckingen, Germany) Feldspar ceramic for conventionally layering	Fired over the zirconia frameworks as veneer porcelain.	
IPS e.max CAD Crystall/ Connect (Ivoclar Vivadent) Fusion glass ceramic	Fused over the zirconia frameworks.	
Panavia F 2.0 (Kuraray Noritake) Self-etching dual cure resin cement	Used for the cementation of the restorations onto the metal dies and for the cementation of frameworks and veneer layers. Light cured.	Light cured from each side for 20s. OXYGUARD II applied. Waited for 3 min and washed.
Ceramic etching gel (Ivoclar Vivadent) 5% hydrofluoric acid gel	Used for roughening the lithium disilicate surfaces.	Applied for 20 s.
Monobond Plus (Ivoclar Vivadent) Silanizing agent	Used for silanization of the lithium disilicate surfaces.	Applied for 60 s.

ally layering (V-L), 4) veneered zirconia by lithium disilicate pressing (V-P), 5) veneered zirconia by lithium disilicate fusing (CAD-F-E), and 6) veneered zirconia by feldspathic ceramic cementing (CAD-C-CB).

The FDPs were cemented onto the metal dies using adhesive resin cement (Panavia 2.0; Kuraray Noritake, Tokyo, Japan). The cementation surfaces of the restorations were prepared according to the manufacturer's

**Table 2** Mean fracture resistance values of the experimental groups

Group/ Material	Control mean ( $\pm$ SD)	Aging mean ( $\pm$ SD)	Total mean ( $\pm$ SD)
M-E	1,367.8 ( $\pm$ 265.8) ( <i>n</i> = 10)	1,259.7 ( $\pm$ 257.5) ( <i>n</i> = 10)	1,313.8 ( $\pm$ 260.6) A ( <i>n</i> = 20)
M-TZI	1,566.7 ( $\pm$ 368.4) ( <i>n</i> = 10)	1,393.1 ( $\pm$ 392.8) ( <i>n</i> = 10)	1,479.9 ( $\pm$ 381.2) A ( <i>n</i> = 20)
V-L	1,076.8 ( $\pm$ 228.2) ( <i>n</i> = 10)	978.7 ( $\pm$ 208.6) ( <i>n</i> = 10)	1,027.8 ( $\pm$ 218.7) B ( <i>n</i> = 20)
CAD-F-E	1,178.2 ( $\pm$ 215.6) ( <i>n</i> = 10)	1,021.6 ( $\pm$ 253.8) ( <i>n</i> = 6)	1,119.5 ( $\pm$ 235.6) B ( <i>n</i> = 16)
CAD-C-CB	1,004.1 ( $\pm$ 159.1) ( <i>n</i> = 10)	948.0 ( $\pm$ 184.9) ( <i>n</i> = 6)	983.1 ( $\pm$ 165.4) B ( <i>n</i> = 16)
Total mean ( $\pm$ SD)	1,238.7 ( $\pm$ 321.5) a ( <i>n</i> = 50)	1,146.0 ( $\pm$ 320.7) b ( <i>n</i> = 42)	

The different capital letters vertically show that the difference is significant among the groups ( $P < 0.05$ ). The different small letters horizontally show that the difference is significant between the groups ( $P < 0.05$ ).

SD, standard deviation; 1) monolithic lithium disilicate (M-E), 2) monolithic zirconia (M-TZI), 3) veneered zirconia by conventional layering (V-L), 4) veneered zirconia by lithium disilicate pressing (V-P), 5) veneered zirconia by lithium disilicate fusing (CAD-F-E), and 6) veneered zirconia by feldspathic ceramic cementing (CAD-C-CB).

recommendations. Finger pressure was applied during cementation. Table 1 summarized the materials used in the study, their fabrication, and the cementation procedures.

After cementation, the roots of the dies were covered apically 2 mm from the crown margin with a 0.25-mm-thick silicone layer to replicate the periodontal ligament. All restorations were embedded into acrylic resin molds. Twenty specimens for each group were fabricated and divided into one of two subgroups: control and artificial aging ( $n = 10$  per group). Specimens in the control groups were directly subjected to fracture resistance testing. Specimens in the aging groups were simultaneously subjected to both thermocycling and mechanical loading (TCML) in a mastication simulator (Chewing Simulator; Esetron Smart Robotechnologies, Ankara, Turkey). Thermocycling was performed for 2,000 cycles in 5°C and 55°C water for 1 min per cycle. Mechanical loading was performed with a 200-N load for  $5 \times 10^5$  cycles. The load was vertically applied on the central fossa of the molar pontic with a steel ball (6 mm in diameter) at an approximate frequency of 2 Hz. A 1-mm sliding movement was also applied during loading. During simulation, failed restorations were excluded from the simulation process and statistical analysis.

Fracture resistance testing was performed using a universal testing machine (Compression/Tension Device; Esetron Smart Robotechnologies). The load was vertically applied to the central fossa of the molar pontic's occlusal surface with a steel ball (6 mm in diameter) at a crosshead speed of 0.5 mm/min. The load at fracture was recorded in Newtons (N), and fractures were examined

and analyzed during the simulation process and after the fracture resistance test. Fracture types were also recorded.

### Statistical analysis

Specimens that fractured before 500,000 cycles and did not survive the aging process were considered failed specimens. Kaplan-Meier survival analysis was used to examine the success rates of the specimens during TCML. The distribution of the fracture load data was analyzed using the Shapiro-Wilk test, and the data were normally distributed. The Levene test was used to assess the homogeneity of the variances ( $P = 0.515$ ,  $P > 0.05$ ). The loads at fracture of the different groups were analyzed by two-way ANOVA, and the means were compared via the Tukey test ( $\alpha = 0.05$ ). All calculations were performed with statistical software (IBM SPSS Statistics for Windows, Version 20.0; IBM Corp, Armonk, NY, USA).

### Results

All specimens in the M-E, M-TZI, and V-L groups survived during artificial aging and showed a 100% success rate. All specimens in the V-P group, as well as four of ten specimens in the CAD-F-E and CAD-C-CB groups, were fractured during the artificial aging process. The mean fracture load of the control group of V-P was  $969.8 \pm 177.7$  N and showed the lowest fracture load value among the groups. All specimens in the aged V-P group fractured during the aging process; therefore, the V-P group was not included in the two-way ANOVA.

The results of the two-way ANOVA showed that there was no interaction between the material and aging factors ( $P = 0.970$ ,  $P > 0.05$ ); however, both the material type ( $P$

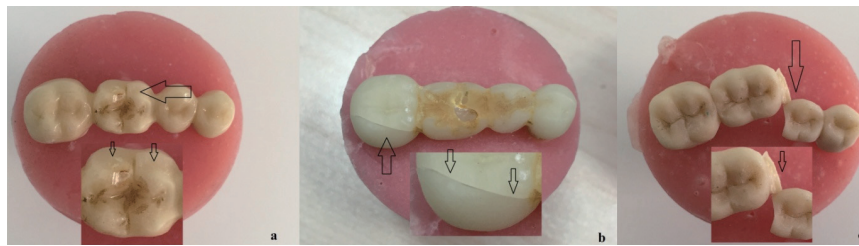
**Table 3** Fracture type of the aging groups during the aging process and after the fracture resistance test

Material	During aging		Success rate (%)	After fracture resistance test		
	Chipping	Connector fracture (restoration fracture)		Chipping	Crack	Connector fracture (restoration fracture)
M-TZI	0	0	100	0	7	3
M-E	0	0	100	0	2	8
V-L	0	0	100	6	0	4 (veneer fracture)
V-P	0	10	0	0	0	0
CAD-C-CB	0	4	60 (CPS:0.6, SE:0.155)	0	0	6 (veneer fracture)
CAD-F-E	1	3	60 (CPS:0.6, SE:0.155)	2	0	4

CPS: cumulative proportion surviving at the time, SE: standard error.

**Table 4** Fracture type of the control groups after the fracture resistance test

Group	Chipping	Crack	Connector fracture
M-E	0	4	6
M-TZI	0	5	5
V-P	0	0	10
V-L	5	1	4
CAD-F-E	0	1	9
CAD-C-CB	2	2	6

**Fig. 2** Fracture types: a) chipping, b) crack, and c) connector fracture.

$< 0.05$ ) and TCML process ( $P < 0.05$ ) affected fracture resistance. When the TCML was disregarded, significant differences were found among the groups ( $P < 0.05$ ). The mean fracture load values are shown in Table 2. The highest fracture load values were observed in the M-TZI group; however, the results were not significantly different from those of the M-E group. Also, no statistically significant differences were found in the values that were observed among the V-L, CAD-F-E, and CAD-C-CB groups.

According to the results, the TCML process decreased the mean fracture load values in all groups. Regardless of the material type, the mean fracture load values significantly decreased between the control and TCML groups, as shown in Table 2. The fracture types of the failed specimens during TCML and those after the fracture resistance test are shown in Tables 3 and 4, respectively. All specimens in the V-P group showed connector fractures during the aging process. The M-TZI, M-E, and V-L groups did not show any failures during TCML. The

fracture resistance test indicated that connector fractures were the most frequent type of failure in all the control groups. The aged M-TZI, M-E, and V-L groups generally showed connector fractures after the fracture strength test. The fracture types of the restorations are shown in Fig. 2.

## Discussion

In the present study, the mechanical performance was tested for the CAD/CAM-generated monolithic and veneered all-ceramic four-unit FDPs before and after TCML in a mastication simulator. The study tested and compared contemporary restorative options for an all-ceramic, four-unit posterior FDP, whereas all-ceramic, three-unit FDPs were generally tested in the literature (3,11,14,16,24). Regarding four-unit posterior all-ceramic FDPs, a few studies have been conducted that included veneered, high-strength ceramic frameworks (e.g., zirconia, lithium-disilicate, or zirconia-reinforced glass-infiltrated alumina) (25-27), but these studies did



not compare M-E and M-TZI, which were veneered with zirconia with various techniques. When there are limited clinical trials, *in vitro* studies simulating clinical conditions should be performed to represent the potential success and possible complications of a treatment modality (28). In the present study, current all-ceramic restorative options for a four-unit, all-ceramic posterior FDP were tested under standardized and, as much as possible, simulated clinical conditions.

With the aim of standardized specimen preparation, metal dies were manufactured by duplicating a prepared tooth from a Co-Cr alloy powder using direct-laser sintering technology. To fabricate standard restorations, all FDPs were designed using the “multilayer” function of the CAD/CAM software, which was used to fabricate CAD-on restorations. Monolithic restorations were milled from this design without any modifications. For the veneered zirconia groups, the anatomical design was split by the software into anatomical framework and veneer layers. Thus, a standardized anatomical zirconia framework could be provided for all the veneered zirconia groups.

Repetitive stress during the mastication cycle must be considered as well as the initial mechanical strength because subcritical crack growth is generally caused by these stresses in ceramics. Therefore, artificial aging is a crucial part of an *in vitro* study (29,30). The TCML parameters were chosen in accordance with previous *in vitro* studies that applied aging to specimens by simulating a 5-year period of intraoral use (29,31). A cyclic loading force of 200 N was applied to simulate clinical conditions in the posterior area as maximum masticatory forces varied greatly, from 190 to 290 N, in the anterior region and could reach up to 360 N in the molar region (20,32,33). Several *in vitro* studies applied 50 N (3,16) or 100 N (26,34) loads during cyclic loading; however, low loading forces may cause overestimation of the fatigue resistance of the tested restoration. Testing the restorations with high cyclic loading forces can more accurately estimate the materials’ survivability under challenging clinical situations (20). Therefore, cyclic preloading at loads between 30 and 300 N was also an approach in artificial aging (24,27,35). The loading direction was a combination of vertical impact and horizontal sliding (1 mm), which was restricted to the molar pontic’s central fossa. There were, however, no internationally accepted standard loading conditions for testing ceramic restorations in a chewing simulator (30,31,36).

After exposure to chewing simulation, all V-P specimens showed major veneer chipping failures or connector fractures, whereas no failure, including chip-

ping, was observed in the V-L group. The more frequent chipping failure in the pressed-over FDPs compared to the conventionally-layered restorations was also reported by Baldassarri et al. (2). Two things may be responsible for this outcome: one is differences in porcelain firing procedures (37,38) and the other is differences in fracture toughness values (2,39). An over-pressed porcelain layer, which is applied by one porcelain firing, must cool down as a complete layer. This may generate higher residual stress between the veneer and zirconia interfaces. In the conventional layering method, multiple firings of veneering porcelain in thinner layers may lead to small shrinkage volumes and lower residual stress (39). Furthermore, heating the specimens may allow for some stress relaxation during the repeated firings of the layers. Also, less porosity may exist between the layers. Conversely, the quality of the interface between the veneer porcelain and zirconia framework layers has been reported as superior for the over-pressing technique because of the presence of structural defects and air bubbles in the conventional veneering technique (40). In the present study, chewing simulations revealed no failure for monolithic FDPs made of zirconia or lithium disilicate. This finding led to a rejection of the first hypothesis, which was that FDPs veneered using different techniques and monolithic FDPs would survive during TCML. The superiority of fatigue resistance of monolithic ceramic restorations compared to bi-layered restorations was previously reported (13,16,23). CAD/CAM-designed veneered FDPs in the current study showed a 60% survival rate during chewing simulation, which consistent with a previous study investigating the fatigue reliability of CAD-on FDPs (14).

Comparing the groups’ fracture loads revealed that monolithic all-ceramic FDPs showed significantly higher fracture resistance than veneered zirconia FDPs. This finding led to the rejection of the second hypothesis that all tested FDP groups would show equal fracture loads before and after aging. Monolithic zirconia FDPs showed the highest fracture resistance, followed by monolithic lithium disilicate FDPs, but with no statistically significant difference. The enhanced performance of monolithic restorations compared to bi-layered restorations, including crowns and FPDs, has been shown in the dental literature (13,14,16,23). There may be various reasons for this difference. For example, low strength veneering ceramics are prone to fail at low loads, while monolithic FDPs composed of high-strength material do not fail. The interface between the core and veneer layers is the weakest link in a veneered system, but it is eliminated in a monolithic restoration (12). Alterna-

tively, the occluso-gingival connector dimensions of the high-strength framework material, which are critically important to the all-ceramic FDPs' mechanical stability, is naturally greater in full-anatomic FDPs (24). Monolithic restorations also provide the advantage of a more conservative preparation (13).

Lithium disilicate glass ceramic and translucent zirconia are the all-ceramic materials used for monolithic FDPs because of their high fracture resistance. The fracture resistance of three-unit monolithic zirconia FDPs has been reported to about 1,800 N (34); however, there is limited *in vitro* information on the mechanical performance of monolithic zirconia FDPs (34,41), and long-term clinical performance has not yet been assessed (42,43). Despite zirconia's high initial mechanical strength, its susceptibility to aging when exposed to the oral environment (cyclic loading, saliva, and thermal changes) is well known; therefore, further research is needed on monolithic zirconia FDPs. Although the lithium disilicate system shows high success rates in anterior and posterior crowns (19,22), very limited data are available on its long-term performance in posterior fixed dental prostheses. In the present study, monolithic lithium disilicate FDPs showed no failures during chewing simulation and showed a fracture resistance comparable to that of monolithic zirconia. Overall, monolithic lithium disilicate FDPs were superior to veneered zirconia frameworks. It should also be emphasized that this study included a four-unit posterior FDP with two pontics.

In addition to the improved veneering techniques for zirconia to overcome chipping or fracture of the veneer ceramic, the CAD-on technique has been introduced (2,12,44). However, very little research has been conducted using CAD-on veneered zirconia FDPs (3,14,35) despite the increasing interest in this technique for crown restorations (8,12,19,23). In the present study, two different CAD-on systems were used, and the fracture loads of both groups were similar and comparable to the conventionally-layered group. In contrast, crown studies revealed higher fracture resistance and more reliability for fused lithium disilicate CAD-on crowns compared to cementing feldspathic CAD-on crowns and conventionally-layered crowns (8,11,23). The divergent findings of this study, compared to the crown studies evaluating zirconia veneering techniques, may be attributed to the different mechanical behavior under loading of crowns and FDPs.

In the present study, the fracture resistance of specimens subjected to simultaneous TCML were compared to non-aged specimens. Considering all the groups,

mastication simulation significantly reduced the fracture loads, as reported by previous research (2,3,14,45); however, no significant effect of TCML on lithium disilicate FDPs was reported by Schultheis et al. (16). Chaar et al. (3) reported that aging significantly reduced the fracture resistance of zirconia FDPs veneered with layering techniques, whereas no significant effect was observed for zirconia FDPs veneered with the CAD-on and press-on techniques. The materials and techniques used in the other reports are quite different than those used in the present study, making a direct comparison rather difficult.

Another important factor for the resistance and longevity of FDPs is the connector dimensions (27). In the present study, the connector dimension of 16 mm<sup>2</sup> and 12 mm<sup>2</sup> for monolithic restorations and zirconia frameworks was established, respectively. The increased connector dimension of the high-strength materials (both zirconia and lithium disilicate) may have led to the superior fatigue and fracture resistance of monolithic FDPs observed in this study. Although there are no accepted standards for adequate connector dimensions, several *in vitro* studies reported that a connector area of 9 mm<sup>2</sup> is appropriate for zirconia-based FDPs (3,25). However, the location and the length of the span and material choice affect the FDP's dimensional requirements (24). Considering posterior localization and the span length (gap for second premolar and first molar teeth) of the FDPs tested in the present study, extended connector dimensions were preferred. However, it should be considered that anatomic conditions (such as crown length of abutment or opposing teeth) can limit increased connector dimensions. Moreover, large connectors may have negative effects on gingival health and aesthetics (16). In the present study, four-unit posterior all-ceramic FDPs were tested, and promising results were obtained. However, limitations should be considered. *In vitro* conditions do not simulate the clinical situation in several aspects, including that *in vitro* aging differs from intraoral aging in nature. During TCML, water was used instead of artificial saliva. As the abutment material affects the fracture load values of ceramic restorations (20), the use of metal abutments in this study may have led to an overestimation of the fracture load values. Also, the bond strength of resin cement differs when bonded to metal alloys versus dentin/resin die (34). To evaluate the survival and complication rates of these restorations, well-designed, long-term, randomized, controlled trials are required. In addition to fracture resistance, the wear, microleakage, marginal, and internal adaptation properties of FDPs after TCML need to be investigated.



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## Conflict of interest

The authors deny any conflict of interest.

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