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Effect of Thermomechanical and Static Loading on the Load to Fracture of Metal-Ceramic, Monolithic, and Veneered Zirconia Posterior Fixed Partial Dentures

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Fixed partial dentures; fracture load; monolithic zirconia; thermomechanical load; veneered zirconia.

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Abstract

Purpose: To evaluate the influence of static (not preloaded) and thermomechanical loading on the load to fracture of metal-ceramic, monolithic and veneered zirconia computer-aided design/computer-aided manufacturing (CAD/CAM) posterior fixed partial dentures (FPDs).
Materials and Methods: One hundred standardized specimens with 2 abutments screwed onto a platform were prepared from stainless steel to receive a posterior 3-unit FPD with an intermediate pontic. Specimens were randomly divided into 5 groups ($n = 20$): Metal-ceramic (control group), Lava Zirconia system, Vita In-Ceram YZ, IPS e.max ZirCAD, and Lava Plus. Half of the specimens of each group ($n = 10$) underwent no preloading, and the other half were subjected to thermomechanical loading in a masticatory simulator, and then all FPDs were loaded until fracture using a universal testing machine at a 1 mm/min crosshead speed. The load to fracture of the veneering ceramic and the load to fracture of framework (total fracture) were recorded for each specimen. Data were statistically analyzed using 2-way ANOVA, Tukey's HSD post-hoc test, Student's t test, and Weibull statistics, $\alpha = 0.05$.
Results: Significant differences were recorded between the metal-ceramic and veneered zirconia groups for the veneering ceramic load ($p < 0.001$; $f = 36.62$; $f = 57.76$) in no preloading and thermomechanical loading subgroups, respectively, but no differences were observed between the static and thermomechanical loading conditions. No differences were observed among the veneered zirconia groups. For the total load to fracture, significant differences were observed according to the material ($p < 0.001$; $f = 500.8$), between the metal-ceramic and Lava Plus group and the other zirconia groups in no preloading subgroup, and between metal-ceramic and the other groups ($p < 0.001$; $f = 303.33$) in thermomechanical loading subgroup. For the type of preloading, significant differences were observed ($p = 0.02$; $f = 5.24$) between the Lava Plus group and the other groups. Thermomechanical loading significantly decreased the fracture load of the Lava Plus group ($p = 0.005$). The Weibull statistics corroborated the results.
Conclusions: Monolithic zirconia restorations provided the highest load to fracture values among the zirconia groups tested; however, the results indicate that they must be used in the oral environment with caution, because their load to fracture was influenced by the aging simulation.

The introduction of computer-aided design/computer-aided manufacturing (CAD/CAM) technology in dentistry has enabled the application and development of new ceramic materials.[1] Despite the evolution of ceramics, metal-ceramic restorations, which have been used for many years, still represent the gold standard technique of choice for posterior crowns and fixed partial dentures (FPDs) given their high mechanical strength and predictability.[2,3]

Load to fracture is one of the most important factors for the long-term success of restorations.[3] Previous studies report that ceramic systems generally exhibit a lower load to fracture and fracture resistance than metal-ceramic restorations.[4,5]

However, currently, ceramic materials exhibit improved mechanical properties to achieve the same level of strength.[6] Yttrium oxide partially stabilized zirconia (Y-TZP) ceramic exhibits very good mechanical properties of flexural strength and fracture toughness[7] among other advantages, such as chemical inertness and biocompatibility.[8]

Zirconia possess low translucency, and the opaque white frameworks must be veneered with porcelain to improve esthetics.[1,9,10] The core/veneer interface is one of the weakest aspects of these restorations. Chipping of the veneering porcelain has been cited as the most frequent reason for failure of zirconia FPDs,[2,7,10-12] and several trials have attempted to reinforce the veneering porcelain.[13,14] Colored zirconia framework has been introduced, developed through staining or infiltrating chloride solution of rare earth elements before sintering, to mask the opaque appearance and whiteness of zirconia frameworks.[15] However, little information is available on optical aspects of colored zirconia, and it appears that using colored frameworks does not offer any direct advantages over the standard natural zirconia frameworks.[15,16] Furthermore, it has been reported that the veneering ceramic is what modifies the final color of ceramic restorations.[15] In addition, colored zirconia framework does not solve the chipping of the veneering porcelain, and to overcome this problem monolithic zirconia has been introduced to the market,[10,17] However, due to its recent introduction, studies regarding its behavior are limited.[10]

Ceramic restorations are subjected to masticatory forces in a wet environment, and several factors, including masticatory forces, temperature, and moisture, can affect their mechanical properties and load to fracture.[18] Therefore, these conditions should be considered during in vitro studies of these materials to extrapolate the results to the clinical situation.[19] However, given that numerous intraoral variables are difficult to reproduce in vitro, in addition to the very few studies[20] of validated laboratory tests that allow systematic research, it is very difficult to perform comparisons among different studies on the load to fracture of ceramics.[18] Recent studies have used the aging process or dynamic loading of specimens to simulate clinical conditions and better predict their long-term success.[7,10,19,21-24] However, the substantial heterogeneity of data makes comparison of results difficult.[25]

The aim of this in vitro study was to evaluate the influence of static (no preloading) and thermomechanical cyclic loading on the load to fracture of metal-ceramic, monolithic zirconia, and 3 different veneered zirconia CAD/CAM systems for posterior FPDs. The null hypothesis was that there would be no differences in load to fracture between all the studied systems.

Materials and methods

One hundred standardized specimens with 2 abutments screwed on a base (30 mm long, 17 mm wide, 4.5 mm thick) were fabricated from stainless steel (316L UNS S3 Alloy; Masteel, Birmingham, UK) in the Mechanical Workshop of the Physical Science Faculty (University Complutense of Madrid, Spain). The abutments were positioned on the base to receive posterior 3-unit FPDs with an intermediate pontic, such that one simulated a first mandibular premolar, and the other simulated a first mandibular molar. The abutments were designed with the following characteristics to simulate clinical conditions: 5 mm in height, 1-mm wide chamfer, a 6° angle of convergence of the axial walls, and rounded angles.[9,10,26,27] They were air-abraded with 100 μm alumina particles to eliminate the surface brightness and to create micromechanical retention.

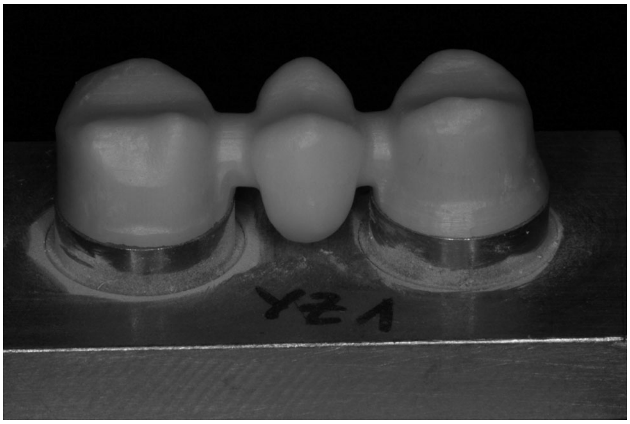

Figure 1 Anatomical design of the In-Ceram YZ framework.

The specimens were used as working dies, and were randomly assigned to 5 groups (n = 20 each) according to the material of the restoration: group 1, metal-ceramic (control group); group 2, Lava Zirconia system (3M ESPE, Seefeld, Germany); group 3, Vita In-Ceram YZ (Vita Zahnfabrik, Bad Säckingen, Germany); group 4, IPS e.max ZirCAD (Ivoclar Vivadent, Schaan, Liechtenstein) and group 5, Lava Plus (3M ESPE). Each group was randomly divided into 2 subgroups (n = 10 each) according to the preloading test: (1) one subgroup was subjected to preloading under thermomechanical loading and (2) the other subgroup underwent no preloading. Thereafter, both subgroups were subjected to static loading until fracture.

Each zirconia FPD was fabricated according to the manufacturers' specifications by an experienced dental laboratory technician. The fabrication of the zirconia frameworks consisted of scanning the steel dies with optical surface scanners (Lava Zirconia and Lava Plus groups: Lava Scan, 3M ESPE; In-Ceram YZ and IPS e.max ZirCAD groups: inEos, Sirona, Bensheim, Germany). The frameworks were designed to exhibit round connectors 9 mm^2 (3 × 3 mm) in size,[2,6,9-11,21,26-28] had an internal space of 50 μm for the cement, and had an axial wall thickness of 0.5 mm for the veneered zirconia groups and 1 mm for the monolithic zirconia group. The frameworks were milled from the specific presintered zirconia blocks in the respective milling units: Lava Form (3M ESPE) for the Lava Zirconia and Lava Plus groups and Sirona in Lab (Sirona) for the In-Ceram YZ and IPS e.max ZirCAD groups (Fig 1). The restorations were veneered with the corresponding hand-layered feldspathic porcelain (0.5 mm thick at the axial walls): Lava Ceram (3M ESPE) for the Lava Zirconia group; Vita VM9 (Vita Zahnfabrik) for the In-Ceram YZ group, and IPS e.max Ceram (Ivoclar Vivadent) for the IPS e.max ZirCAD group. The monolithic zirconia group was left unveneered, such that the FPDs with the final morphology were designed.

Metal-ceramic restorations were fabricated following the traditional lost-wax technique and were vacuum cast using a base-metal alloy of chromium-cobalt (Ugirex C; UginDentaire, Seyssinet-Pariset, France). The dies for copings were coated with 3 layers of die spacer (Space-It; TAUB Products, Jersey City, NJ), (total thickness, ∼50 μm). The patterns were waxed-up (Classic modeling wax-blue; Renfert GmbH, Hilzingen, Germany) onto the metallic dies with 0.5 mm thick axial walls and 9 mm^2 connectors[21,26] and invested with a graphite-free phosphate investment (Vestofix; DFS Diamon GmbH, Riedenburg, Germany). The casting was performed with an induction-heated and centrifugal vacuum/pressure casting machine (Jelrus Infinity L30; Whip Mix, Dortmund, Germany). After divesting, the castings were cleaned using an airborne particle abrasion device with aluminum-oxide powder (125 μm), and the veneering porcelain (Omega 900; Vita Zahnfabrik) was applied.

To ensure that all the specimens had the same porcelain thickness, one of the FPD frameworks was waxed, and the shape was duplicated using a soft and light-bodied silicone putty material (Express Penta Putty and Express Penta Ultra-Light Body; 3M ESPE) to be used as a guide. The veneering porcelain was then applied from the silicone index.[5,10] All restorations had the same final dimensions as verified by measuring the FPDs at different locations using a digital micrometer (Mitutoyo Co, Tokyo, Japan).[10,26]

The FPDs were luted to their corresponding master dies by the same operator with glass ionomer cement (Ketac Cem EasyMix; 3M ESPE) mixed following the manufacturer's specifications[10,26,27] at room temperature (18-24°C) and 50 ± 10% relative humidity.[18] The cement was applied with a brush on the axial surfaces of the abutments, and a constant seating load of 10 N was applied to the occlusal surface for 10 minutes using a dynamometric key (USAG 820/70; Utensilerie, Milan, Italy).[9,10,26,27]

After 48 hours of water storage of all groups, half of the specimens in each group were subjected to thermal and mechanical cycling loading simultaneously. Thermal cycling was performed in distilled water for 24 hours for 1032 thermal cycles at 5°C and at 55°C with a 30-second dwell time. The mechanical cycles were performed simultaneously by a masticatory simulator (Chewing Simulator CS-4.2 economy line, Thermo-cycling TC-3; SD Mechatronik GMBH, Feldkirchen-Westerham, Germany)[3,10,29] at a rate of 6550 mechanical cycles per hour (120,000 mechanical cycles). In the present study, a vertical load of 50 N with a vertical displacement of 2.5 mm at a 60 mm/sec crosshead speed was administered at the central fossa of the pontic.[7] After thermomechanical loading, all specimens were inspected under a stereomicroscope (Nikon SMZ-10; Nikon Corporation, Tokyo, Japan) at 40x magnification to ensure that no fracture occurred on the veneering ceramic. The other half of specimens in each group were not subjected to thermomechanical loading.

Subsequently, all experimental subgroups were loaded until fracture (National Centre of Metallurgical Research-CENIM; CSIC, Madrid, Spain) using a universal testing machine (UTM) (ME 405/10; SERVOSIS SA, Pinto, Spain),[9,10,27] with a 10 kN load-cell, at a 1 mm/min crosshead speed[10,18,27] at room temperature (23 ± 1°C). Axial compressive vertical load was applied at the central fossa of the FDP pontic by sliding a cone-shaped stainless steel bar (length: 12 mm) finished in a rounded tip adapted to the UTM until fracture of the veneering ceramic and/or fracture in the framework (total fracture) of the restorations (Figs 2–6). Clinically, the fracture of the veneering ceramic can be classified as reparable, nonreparable, and adhesive/cohesive chipping, but in the study the veneering fracture initiation point was defined as a sharp decrease in the stress plot of the loading curve, and was perceived as visible signs of chipping and loud cracking sounds.[9,10] It represented a major or nonrepairable chipping. The fracture of the framework (total fracture) was defined by a drastic drop in the loading curve together with a visible breakage of the prosthesis.[9,10]

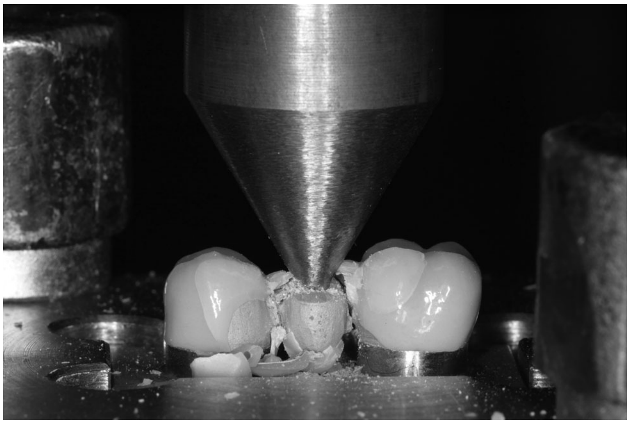

Figure 2 Fractured specimen of the metal-ceramic group, showing the total fracture and an adhesive fracture of the veneer ceramic with metal exposure.

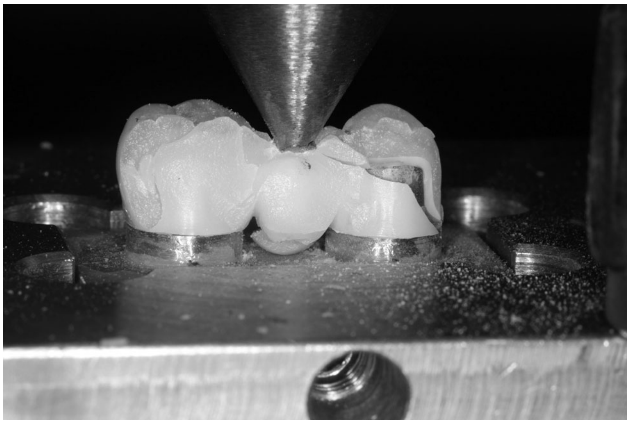

Figure 3 Fractured specimen of the Lava Zironia group.

Data on the veneering ceramic fracture and total loads to fracture of the FPDs were automatically recorded in Newtons (N) using built-in software (PCD2K; SERVOSIS, Madrid, Spain) that allowed force (N)-displacement (mm) curves to be created.[9] Means and standard deviations (SD) were calculated for each group. The normality of the variables was confirmed by the Shapiro-Wilk test. Data were statistically analyzed using 2-way ANOVA and Tukey's HSD post-hoc test for comparisons of the load to fracture values among the groups. Student's t test was used for comparisons between the static and thermomechanical conditions for each group. To facilitate accurate interpretation of data, the parameters of the Weibull

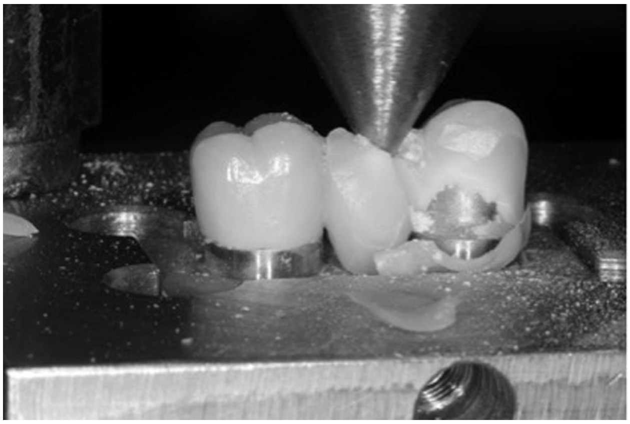

Figure 4 In-Ceram YZ specimen showing the fracture at the connector.

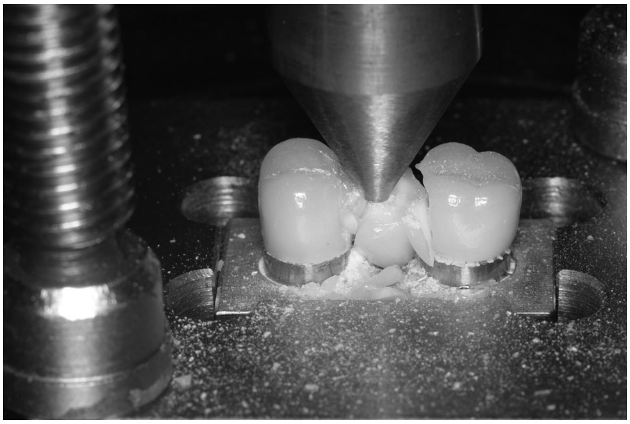

Figure 5 Fractured specimen of the IPS e.max ZirCAD group.

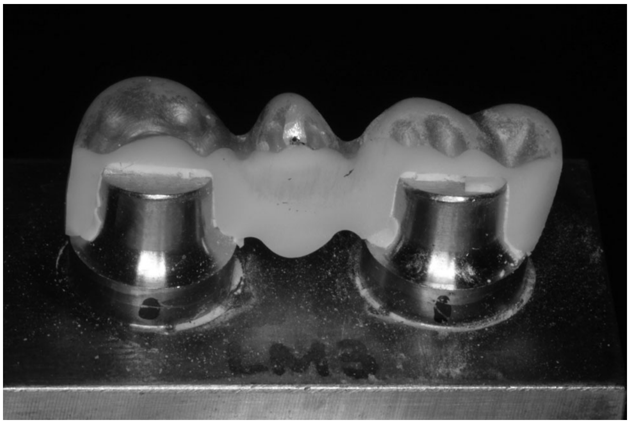

Figure 6 Fracture in half, lengthwise, of the Lava Plus group.

distribution, the Weibull modulus (m), and the characteristic fracture load ($\sigma 0$) were also analyzed by maximum likelihood with a 95% CI.[9,10,30] Statistical significance was set to 0.05. Statistical software (SPSS 22; SPSS Inc., Chicago, IL) was used for the analysis.

Table 1 Load to fracture for veneering ceramic in Newtons (N)

		Static loading		Thermomechanical loading	
Group	N	Mean	SD	Mean	SD
MC	10	3043.97^a	246.89	3008.69^a	193.65
LZ	10	1076.82^b	227.22	927.96^b	330.29
YZ	10	1035.27^b	611.10	969.90^b	663.39
ZZ	10	1145.91^b	750.35	928.35^b	391.04

MC = metal ceramic; LZ = Lava Zirconia; YZ = Vita In-Ceram YZ; ZZ = IPS e.max ZirCAD

Different superscript letters in the mean column indicate significant differences among groups ($p < 0.001$).

Table 2 Load to total fracture in Newtons (N)

		Static loading		Thermomechanical loading	
Group	N	Mean	SD	Mean	SD
MC	10	8313.83^a	624.05	7958.15^a	932.96
LZ	10	2017.42^b	344.01	1966.27^b	397.86
YZ	10	1967.25^b	366.58	1869.84^b	211.36
ZZ	10	1990.96^b	204.14	1908.74^b	118.90
LP	10	2605.33^c	288.15	2181.67^b	303.99

MC = metal ceramic; LZ = Lava Zirconia; YZ = Vita In-Ceram YZ; ZZ = IPS e.max ZirCAD; LP = Lava Plus.

Different superscript letters in the mean column indicate significant differences among groups ($p < 0.001$).

Results

All FPDs survived thermomechanical fatigue, and no cracks or fracture failures were observed. Comparisons of the 2 types of preloading tests for analysis of the load to fracture revealed that thermomechanical loading produces a slight decrease in the values of the veneering ceramic fracture load and the total fracture loads in all the groups analyzed (Tables 1 and 2), but only significant differences were observed for the total load to fracture ($p < 0.02$; f = 5.24).

Significant differences among the groups, regarding the material, were noted for the load to fracture of the veneering ceramic in both subgroups: no preloading ($p < 0.001$; f = 36.62), and thermomechanical loading ($p < 0.001$; f = 57.76) (Table 1). Tukey's HSD post-hoc test revealed that the values for the metal-ceramic group were significantly higher than those for the veneered zirconia groups ($p < 0.001$); however, no differences were observed between the preloading conditions ($p = 0.273$).

For the total load to fracture, ANOVA revealed significant differences regarding the material among the groups ($p < 0.001$; $F = 500.8$), including differences between the metal-ceramic and Lava Plus group and the other groups when analyzing the subgroup with no preloading. Likewise, in the thermomechanical loading subgroup, significant differences were observed among the groups ($p < 0.001$; f = 303.33). The load to fracture of the metal-ceramic group was significantly higher than the other groups ($p < 0.001$) (Table 2). No differences were noted

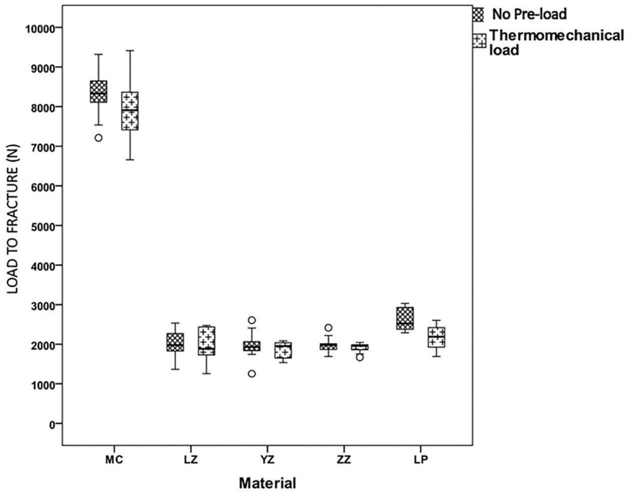

Figure 7 Box plots of load to fracture results of the different groups reported in Newtons (N).

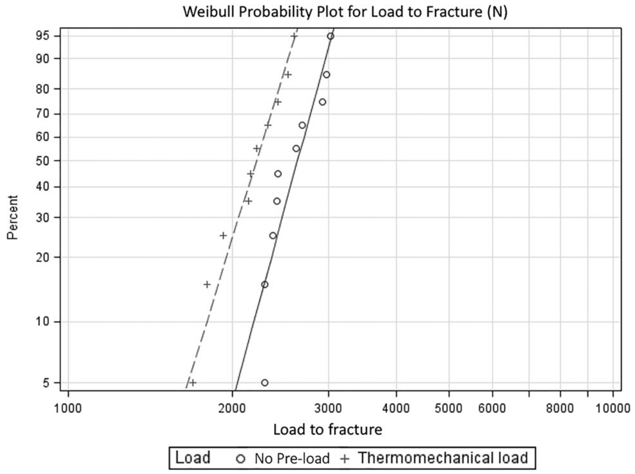

Figure 8 Weibull probability plot of the fracture load of the monolithic zirconia group. Cross line: thermomechanical load. Dotted line: No preload.

among the veneered zirconia groups. Significant differences were also observed regarding the total load to fracture for the type of preloading ($p = 0.02$; f = 5.24). The Lava Plus group exhibited significantly reduced load to fracture under thermomechanical loading compared to no preloading subgroup ($p = 0.005$) (Fig 7). These data were corroborated by the Weibull distribution parameters (Table 3). An overlap was noted for the metal-ceramic and veneered zirconia groups. Thus, no significant differences were observed between the preloading conditions; however, significant differences were detected between these conditions for the Lava Plus group (Fig 8).

Discussion

The data obtained support the rejection of the null hypothesis, as the load to fracture values of the metal-ceramic and monolithic zirconia FPDs exhibited significant differences from those of the veneered zirconia groups. In addition, the monolithic zirconia group was influenced by thermomechanical loading.

The present study compared the load to fracture of posterior FPDs under 2 conditions: no preloading and thermomechanical loading. Several authors have performed static load tests to study the load to fracture of ceramic materials and suggest that compressive testing is an adequate method.[1,9,27,28,31] However, other authors[3,10,12,19,22-24,29,32-39] recommend the use of artificial aging to reproduce oral environment conditions for in vitro studies of ceramic materials, as they may involve a decrease in fracture resistance. Nevertheless, fewer studies comparing static loading with some type of artificial aging have been reported,[7,19,23,24,34,35,38-40] and the results are controversial. One of the studies reported a reduction in the fracture resistance by approximately 40% with respect to the static load in veneered zirconia FPDs.[33] Furthermore, thermomechanical loading combinations are unusual in zirconia studies.[7]

In this study, metal-ceramic FPDs demonstrated significantly increased load to fracture values in both veneering ceramic and frameworks, with both types of preloading, compared to the zirconia groups. These findings are consistent with previous studies.[3,5,31,35] The veneering ceramic fractured at values lower

Table 3 Weibull statistics of fracture load for static (S) and thermomechanical loading (TML)

	m = Weibull shape				σ_0 = Weibull scale			
Group	Estimate	St Error	Lower	Upper	Estimate	St Error	Lower	Upper
MCS	15.79	3.80	9.84	25.32	8585.66	181.82	8236.58	8949.53
MCTML	9.63	2.30	6.02	15.41	8361.04	290.80	7810.06	8950.89
LZS	7.00	1.71	4.33	11.32	2155.41	102.78	1963.08	2366.57
LZTML	5.95	1.48	3.64	9.72	2122.06	118.98	1901.22	2368.55
YZS	6.22	1.48	3.90	9.93	2111.89	113.39	1900.94	2346.24
YZTML	12.92	3.56	7.52	22.19	1954.73	49.95	1859.22	2055.14
ZZS	9.86	2.22	6.34	15.35	2082.91	71.03	1948.24	2226.89
ZZTML	23.86	6.38	14.12	40.31	1957.08	27.11	1904.65	2010.95
LPS	10.23	2.50	6.33	16.54	2732.32	89.58	2562.26	2913.67
LPTML	8.95	2.26	5.45	14.71	2306.81	85.88	2144.48	2481.44

MC = metal ceramic; LZ = Lava Zirconia; YZ = Vita In-Ceram YZ; ZZ = IPS e.max ZirCAD; LP = Lava Plus.

than the metal plastic deformation, and was predominantly adhesive, as previously reported.[3,31,35] The metal fracture values were similar to a previous study for chromium-cobalt FPDs' unveneered structures.[41]

Few studies are available concerning the load to fracture of frameworks and porcelain veneering on zirconia posterior FPDs.[5,10,27] In this study, the veneer porcelain fractured at a lower load than the framework in all the veneered zirconia specimens with both types of preloading, which is consistent with previous studies.[1,27,31,42] The failure pattern observed was predominantly cohesive,[11,18,27,39,43] and this result is probably due to the superior mechanical properties of zirconia framework compared to veneer porcelain.[1] Clinical studies have also demonstrated that the main complication identified in posterior zirconia FPDs is the chipping of the veneer porcelain.[2,11,12,44] Previous studies have reported that several factors are involved in the fracture of the veneering porcelain.[27,39,40,42,45,46] To address this problem, new ceramics and techniques have been developed for veneering zirconia frameworks,[13,14,46-48] and recently, monolithic zirconia has also been introduced.[10,17,38,49,50] However, no difference in the load to fracture of the veneering ceramic between the static and thermomechanical loading conditions was observed in this study, and this result is inconsistent with other reports, which have shown that thermomechanical loading significantly reduced the load to fracture of the veneer ceramic.[34,35] This difference could be due to the different methodologies employed, that include the type of the restoration analyzed (crown or FPD), the type of die employed, or the number of cycles and the force applied during the thermomechanical loading.

No differences were observed between the two types of preloading for total load to fracture of veneered zirconia groups. These findings are consistent with previous studies[7,19,24,37,39,40] but inconsistent with other studies where thermomechanical loading exhibited a significant influence on the load to fracture.[23,32-35,37,46] However, the monolithic zirconia group was clearly affected by thermomechanical loading. This finding could be due to the potential sensitivity of zirconia to aging given that Y-TZP is directly exposed to the oral environment in monolithic restorations, and the mechanical properties could be affected as previously reported.[38] However, limited data on monolithic zirconia are currently available. Only one recent study[38] in crowns was reported, and the results differ with those of the present study, because the thermomechanical loading did not significantly decrease the load to fracture. It is likely due to the different aging simulation tests employed.

The relevance of cementation is an important point for the load to fracture tests.[30] Zirconia-based restorations offer the possibility of conventional and adhesive resin cements,[30,43] and previous studies have reported that no differences were observed for the load to fracture of bilayered or monolithic zirconia restorations between adhesively bonded and conventionally cemented restorations.[30,43,51] Furthermore, clinical studies also revealed no increased incidence rate of fracture related to cementation of zirconia-based restorations with glass ionomer cement.[52] In light of the above, in the study a glass ionomer cement was used for all restorations.

There are several limitations to this study. The study was performed in vitro, and despite the limitations and drawbacks of this study, important facets of the clinical conditions were simulated. Although the thermomechanical loading condition employed met the requirements of the ISO TR 11450, several authors have recommend that additional cycles are necessary.[7,34,38] This feature could influence the results of the veneered zirconia groups analyzed.

The results have clinical implications. Monolithic zirconia posterior FPDs provide high load to fracture values, and may be an alternative to avoid chipping; however, they were influenced by thermomechanical loading, indicating that their use in the oral environment could potentially shorten their clinical life.

Conclusions

1. All tested groups demonstrated clinically acceptable load to fracture values.
2. The monolithic zirconia group exhibited the highest fracture resistance values among the zirconia groups independent of the type of preloading.
3. Thermomechanical loading did not affect the load to fracture of veneering ceramic or the total fracture of the metal-ceramic and veneered zirconia groups, but it reduced the load to fracture of the monolithic zirconia group.

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