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Fracture load of metal-ceramic, monolithic, and bi-layered zirconia-based posterior fixed dental prostheses after thermo-mechanical cycling

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ABSTRACT

Objectives: This study aims to evaluate the fracture load of differently fabricated 3-unit posterior fixed dental prostheses (FDPs) with an intermediate pontic.
Methods: Fifty sets of two stainless-steel abutments were randomly assigned to five groups ($n = 10$ each) depending on the material and technique used for manufacturing the FDPs: (1) Metal-ceramic (MC, control); (2) Lava Zirconia (LZ, bi-layered); (3) Lava Plus (LM, monolithic); (4) VITA In-Ceram YZ (YZ, bi-layered); and (5) IPS e-max ZirCAD (ZZ, bi-layered). After being luted to the dies, all FDPs were submitted to thermo-mechanical cycling (120,000 masticatory cycles, 50 N; plus 774 thermal cycles of 5 °C/55 °C, dwell time: 30 s). Samples were then subjected to a three-point bending test until fracture in a universal testing machine (cross-head speed: 1 mm/min). Fracture load of the veneering ceramic (VF) and total fracture load (TF) were recorded. Microstructure and failure patterns were assessed. Data was analysed using one-way ANOVA and Tukey HSD *post-hoc* tests ($\alpha = 0.05$).
Results: MC restorations recorded higher VF and TF values than did zirconia FDPs ($p = 0.0001$), which showed no between-group differences. Within the bi-layered groups, TF was significantly higher than VF. LM pieces registered lower average grain size than did LZ specimens ($p = 0.001$). Overall, the connector was the weakest part.
Conclusions: All of the groups tested could withstand clinical chewing forces in terms of average fracture load. Zirconia-based samples performed similarly to each other, but showed lower mean fracture load values than did metal-ceramic ones.
Clinical significance: Monolithic zirconia may be recommendable for solving the chipping problem.

1. Introduction

Metal-ceramic fixed dental prostheses (FDPs) have widely been used since the early 1960s, showing good success rates [1]. However, given the request for more aesthetic and biocompatible treatments [2,3], from the late 1990s, the research has been focused on CAD/CAM (Computer Aided Design/Computer Aided Manufacturing) high-strength ceramic systems for achieving improved metal-free prosthetic solutions [3–6].

Yttrium-oxide partially-stabilised zirconia (Y-TZP) has become an attractive high-toughness core material for posterior FDPs, as it may withstand occlusal forces in the molar region [6–10]. Zirconia (also called 'ceramic-steel') has a completely crystalline microstructure, a

stress-induced toughening ('transformation toughening') mechanism [11] and physical features that are similar to those of stainless-steel [12]. However, it generally requires to be veneered with glass ceramic for obtaining proper aesthetics [3,13,14].

Regrettably, the 'chipping' of the veneering porcelain creates an uncertainly as regards the durability of zirconia-based restorations [6,15–19]. Minor chipping, consisting of little fractures, can be polished or repaired with composite resin. Conversely, major chipping of larger ruptures involving functional areas that would lead to a significant alteration of the anatomy after adjustments is considered 'failure', as the only option is the prosthesis' replacement [2,20]. Several efforts have been made to reinforce the fracture strength of the veneering porcelain,

such as the use of high-density CAD/CAM ceramic [21], press ceramic [22], or the 'Rapid Layer Manufacturing' (RLM) technique that individually fabricates the CAD/CAM core and the veneer layers and applies low-fusing glass ceramic or resin-cement to bond the interface [23].

Monolithic zirconia was introduced to circumvent the drawbacks of bi-layered all-ceramic prostheses [24], although its chemical stability and long-term clinical behaviour still remain insufficiently documented. Moreover, there is very little research on the comparison of the fracture load of different zirconia systems and the selection criteria are usually based on personal preferences and experience rather than on sound scientific rationale. FDPs are subjected to complex clinical conditions that induce temporary deformations and internal stresses within the materials and their interfaces [2]. For instance, water can act chemically at crack tips and decrease the ceramic strength, thus having influence on cyclic and static loading tests [25]. Simulated moisture, thermal and mechanical fatigue may therefore approximate the influence of the oral environment before compressive static forces are applied to record the fracture load of the tested materials by means of a three-point bending test [2,26–30].

Hence, the aim of this paper is to evaluate the fracture load of the veneering ceramic (VF), the fracture load of the complete restoration or 'total fracture load' (TF), and the fracture pattern of monolithic and bi-layered zirconia-based 3-unit posterior FDPs after thermo-mechanical cycling using a metal-ceramic group as control. Recently marketed CAD/CAM technologies were tested in order to identify the most resistant system. No previous study has compared such parameters between monolithic and veneered zirconia FDPs after thermal and mechanical fatigue followed by three-point bending flexural test.

The null hypotheses tested were that *(1)* the fracture load values (VF and TF) did not depend on the restorations' material and manufacturing technique, and that *(2)* VF and TF values coincided within each bi-layered method.

2. Materials and methods

2.1. Fabrication of the restorations

A hundred standardised stainless-steel master dies that simulated premolars prepared to be retainers of all-ceramic FDPs were fabricated. The abutments' configuration was: 5 mm in height; convergence of 6˚; chamfer of 120˚, 1 mm in width and 8 mm in diameter; axial reduction of 1 mm; vestibular and lingual cuspids separated by a central groove of 1 mm in depth and rounded angles [31–35]. The dies were randomly screwed in pairs onto fifty metallic bases customised as follows: 4.5 mm in height, 30 mm in length, 17 mm in width and two centred perforations separated by 7 mm.

The fifty sets of samples formed by the abutments fixed onto their respective bases were sandblasted with $100 \mu m$-alumina particles to eliminate the surface brightness and facilitate the scanning procedures and were randomly assigned to five groups (*n* = 10 each) depending on the material and manufacturing technique to be used for constructing 3-unit posterior FDPs with an intermediate pontic: MC: Cast metal-ceramic (control); LZ: Lava Zirconia (3 M ESPE, Seefeld, Germany); LM: Lava Plus (3 M ESPE); YZ: VITA In-Ceram YZ (Vita Zahnfabrik, Bad Säckingen, Germany) and ZZ: IPS e.max ZirCAD (Ivoclar Vivadent, Schäan, Liechtenstein).

The LM group consisted of monolithic zirconia (ZrO_2) pieces, while the remaining zirconia groups, *i.e.,* LZ, YZ, and ZZ were bi-layered restorations of yttrium-stabilised zirconium oxide frameworks veneered with glass ceramic. To date, the specific chemical identity and/or exact composition of the polycrystalline zirconia frames has frequently been withheld as a trade secret. Nonetheless, as a reference, the YZ cores include 91–94% of ZrO_2, 4–6% of Y_2O_3, 2–4% of HFO_2, and less than 0.1% of Al_2O_3, SiO_2, and Na_2O. The same occurs with the composition of the veneering porcelain. However, in all cases (MC, LZ, YZ, and ZZ),

the coating material were feldspathic ceramic (or 'glass ceramic') and consisted almost entirely of an amorphous glass phase with crystalline constituents such as leucite crystallites. All FDPs were obtained according to the manufacturers' instructions.

The restorations of the MC group (control) were fabricated using the traditional lost-wax casting technique. A base metal alloy of white Co-Cr for ceramics was selected (Ugirex C; Ugin Dentaire, Seyssinet-Paris, France). The alloy composition was: Co: 62%; Cr: 31%; Mo: 4%; and Si: 2.2%. The patterns of the structures were waxed-up onto the metallic abutments, which were varnished with three layers of die spacer beforehand to leave gap of $50 \mu m$ for the luting material. The frameworks were modelled with a wall-thickness of 0.5 mm and round-shaped connectors of $9 mm^2$, and were invested. The MC structures were cast in an induction heated vacuum pressure machine (Jelrus Infinity L30; Whip Mix, Dortmund, Germany), at 1470 ˚C. The casts were then retrieved and cleaned using airborne-particle abrasion with alumina (Al_2O_3) powders ($125 \mu m$) for 10 s at a working distance of 5 mm and a pressure of 2 bars.

For the oxidation of the divested structures, two thin layers of opaque porcelain (Omega 900; VITA Zahnfabrik, Bad Säckingen, Germany) were applied to the metal surfaces and were fired to 900 ˚C under vacuum in a calibrated ceramic oven (Programat P500/G2; Ivoclar-Vivadent AG, Schäan, Liechtenstein). The metallic cores were then veneered with dental and incisal compatible glass ceramic (Omega 900; VITA Zahnfabrik) at 910 ˚C. Finally, the glazing procedure was carried out at 915 ˚C.

The LZ, LM, YZ, and ZZ samples were prepared by an experienced technician who was accustomed to work with the CAD/CAM systems tested. The fabrication of the Y-TZP structures started by scanning and digitizing each pair of abutments with a specific non-contact, optical scanner depending on the study group: Lava Scan (band projection/triangulation, precision in the range of $25 \mu m$; 3 M ESPE, Seefeld, Germany) for LZ and LM groups; and InEos X5 Scanner (5-axis technology with automatic positioning combined with auto focus, precision < $12 \mu m$; Sirona Dental, Bensheim, Germany) for YZ and ZZ samples. A 3-D model was generated in each scanning cycle by using the next packages software for Windows: Lava System 3.01 CAD (3 M ESPE) for LZ and LM specimens and Sirona inLab 3D (Sirona Dental) for YZ and ZZ groups.

The following parameters were programmed with the abovementioned software. For the bi-layered groups (LZ, YZ, and ZZ): wall thickness of 0.5 mm in each plane. For the monolithic group (LM): total thickness of 1 mm at the axial walls and 2.5 mm at the occlusal surfaces. In all zirconia groups: an internal space of $50 \mu m$ for the luting agent and rounded connectors of $9 mm^2$. An enlargement of 20% was envisioned in all zirconia specimens to offset the shrinkage derived from the final sintering procedure that permitted zirconia pieces to achieve an increased density at their final size [32,33].

Pre-sintered CAD/CAM blocks of Y-TZP that were compatible with each system were selected to mechanise the samples in the respective milling units, *i.e.,* Lava Form (3 M ESPE) for the LZ group (obtaining milled structures) and LM group (obtaining milled FDPs with their final shape); and Sirona inLab (Sirona Dental) for the YZ and ZZ groups (obtaining milled structures). The sintering procedures of the enlarged pieces lasted for 8 h and were accomplished in specific ovens at high temperatures: Lava Therm (3 M ESPE) at 1500 ˚C for the LZ and LM groups; and VITA Zyrcomat (VITA Zahnfabrik) at 1530 ˚C for the YZ and ZZ groups.

The zirconia structures of the bi-layered groups were veneered in a specific oven (Programat P500/G2 M; Ivoclar Vivadent) with compatible hand-layered feldsphatic ceramic.

In the case of the LZ frameworks, a 0.1/0.2-mm film of liner (Zirconia Overlay Porcelain for Lava Frame; 3 M ESPE) was applied and submitted to a firing cycle at 820 ˚C. The structures were then covered with the corresponding ceramic (Lava Ceram; 3 M ESPE).[3] The first layer of dentine was submitted to 810 ˚C, while both the second layer of

dentine and the incisal coating were fired at 800 °C. The glazing procedure was made at 790 °C.

With respect to the YZ structures a 0.1/0.2-mm layer of liner (VITA VM9 Effect Bonder; VITA Zahnfabrik) was spread with a brush and fired at 980 °C. The frameworks were then veneered with two layers of dentine ceramic (VITA VM9 Base Dentine; VITA Zahnfabrik) that were heated to 910 °C, and 900 °C, respectively. Both the incisal veneering (VITA VM9 Enamel ENL; VITA Zahnfabrik) and the glazing processes were performed at 900 °C.

Concerning the ZZ frames, a 0.1/0.2-mm film of liner (IPS e.max Ceram ZirLiner; Ivoclar Vivadent) was fired at 960 °C. The structures were then coated with two layers of dentine and one layer of enamel of compatible veneering ceramic (IPS e.max Ceram; Ivoclar Vivadent) that were heated to 750 °C. The restorations were glazed at 725 °C.

Thus, monolithic zirconia (LM group) was left unveneered. These samples had a total thickness of 1 mm at the axial walls, and 2.5 mm at the occlusal surfaces. In the bi-layered groups, the thickness of the veneering coating was 0.5 mm at the axial walls, and 2 mm at the occlusal surfaces. Hence, all of the samples had identical final dimensions. During fabrication, the thickness of each structure and veneering material were verified by taking measures at different locations with a digital micrometer (Mitutoyo Co., Tokyo, Japan) with an accuracy of 0.01 mm. Finally, both retainers of every restoration were numbered to identify the study group and specimen [36].

2.2. Material characterization

Prior to the mechanical test, the microstructure was analysed on flat-parallel 1.8 mm-thick samples ($n = 10$ per zirconia group) that were cut from pre-sintered CAD/CAM blocks in a metallographic microcutting machine (Micromet; Remet, Bologna, Italy) with a diamond disk under irrigation. The specimens were sintered and veneered according to the manufacturers' specifications. The final pieces were embedded in epoxi resin and polished to 1 μm in an automatic device (Phoenix Beta; Buehler) by using different diamond suspensions. Finally, samples were thermal-etched at 1300 °C for 30 min. The specimens were then examined under Field emission gun scanning electron microscope (FEG-SEM, Nova NanoSEM 230; FEI, Eindhoven, Holland) at 40,000× magnifications. The average grain size was determined by applying the line intersection method to the representative FEG-SEM images of each group (ASTM E112-96(2004)e2 linear intercept count method) [37,38].

2.3. Luting procedure

The FDPs were randomly luted to the master dies with glass-ionomer cement (Ketac-Cem EasyMix; 3M-ESPE) in standard fashion [34,35]. The same powder-liquid proportion and mixing time (30 s) were used in all cases. All of the FDPs were handled by one operator at room temperature (RT: 23.0 ± 1.0 °C) and relative humidity (50 ± 5%) [27]. A special clamp was designed to maintain a constant seating pressure during cementation [3,39,40]. The axial surfaces of the abutments placed in the clamp were varnished with a thin layer of cement before inserting each FDP. The clamp press was unscrewed until its base made contact with the occlusal surface of the introduced restoration [41]. The upper screw that controlled the press was fitted with a dynamometric key (USAG 820/70; SWK Utensilerie SRL, Milan, Italy) [42,43] that permitted the application of a compressive force of 10 N for 10 min [3,39,40] (thus surpassing the total working time of the luting agent, which was 7 min). This uniform axial load was applied to ensure a correct cement distribution and to counteract the thyrotrophic behaviour of the luting material [43]. Finally, the excess cement was removed with a plastic scaler to avoid scratching or gouging the abutments and/or restorations [27,34,44].

2.4. Thermo-mechanical cycling

After 48 h of water storage, all cemented samples were subjected to thermo-mechanical cycling. Mechanical cycles of axial compressive loads were exerted by a masticatory simulator (Chewing Simulator CS-4.2 economy line; SD Mechatronik GmbH, Feldkirchen-Westerham, Germany) through a customised cone-shaped stainless-steel bar finished in a rounded tip that described a vertical displacement of 2.5 mm at a speed of 60 mm/s. The simulator applied forces of 50 N at the central fossa of each pontic, equidistantly from the abutments [2].

Thermal cycles were accomplished by using a specific machine (Thermo-cycling TC-3; SD Mechatronik GmbH) that was adapted to the chewing simulator. Therefore, both the mechanical and thermal cycling were simultaneously performed maintaining the functional independence of both equipments. The thermo-cycling was made in distilled water, with a dwell time of 30 s and temperatures ranging from 5 °C to 55 °C (41 °F to 98 °F) [14,30,45].

Each sample was thermo-mechanical cycled for 18 h at rates of 43 thermal cycles and 6550 mechanical cycles per hour. Hence, each restoration underwent 774 thermal and 120,000 mechanical cycles. These data simulated approximately 6 months of oral service [46]. Afterwards, all pieces were analysed under a stereomicroscope (Nikon SMZ-10; Nikon Corporation, Tokyo, Japan) at 40× magnifications to verify that no fracture had occurred on the veneering ceramic.

2.5. Three-point bending test

After thermo-mechanical fatigue simulation, all FDPs were subjected to a three-point bending test until fracture [47,48] according to the ISO 6872:2008 standard by means of a universal testing machine (UTM, ME 405/10; SERVOSIS SA, Pinto, Madrid, Spain) [3] operated with a 2000 kgf load cell at a cross-head speed of 1 mm/min [27,49] at RT [3].

A special metallic platen (height: 7 mm, length: 6 mm, width: 6 mm) was adapted to the UTM for accommodating the bases of the specimens. Axial compressive static loads were exerted by sliding a customised cone-shaped stainless-steel bar (length: 12 mm) finished in a rounded tip (diameter: 1 mm). The machined load piston was adapted to the UTM and was perpendicularly applied at the central fossa of each pontic, equidistantly from the abutments [34,48], until the fracture of the veneering material (VF) and the complete fracture of the restoration (TF) occurred. The fracture initiation point (VF) was determined by a sharp decrease in the stress plot of the loading curve and was perceived as visible signs of chipping and loud cracking sounds. Therefore, in this study, the VF variable represented the 'major chipping' that is clinically considered as 'failure'. The TF was defined by a drastic drop in the loading curve together with the visible breakage of the piece [3,36]. VF and TF values were registered for each specimen using inbuilt software for the testing machine (PCD2K; SERVOSIS SA) that allowed tensile-deformation or force (N)-displacement (mm) curves to be automatically created.[3] The load values were recorded in kilograms-force (kgf) and were converted into Newtons (N) before data analysis [48].

2.6. Statistical analysis

Ten observations were made per zirconia group for microstructure analysis, resulting in a total of 40 measures in the experiment. Ten VF values and ten TF values were recorded for each group tested except for the monolithic LM group, in which only TF values were taken. This yielded a total of 40 VF and 50 TF values.

Means and standard deviations (SD) of average grain size, VF and TF values were calculated for each study group. The normality of the variables was confirmed by the Kolmogorov-Smirnov and the Shapiro-Wilk tests [50,51] and the homogeneity of variances was verified according to the Levene's test [36].

The one-way analysis of variance (ANOVA) [30,51–56] was run to

assess the contribution of the material and manufacturing technique to: *(1)* the average grain size of each zirconia group, and *(2)* to the fracture load values (VF and TF) registered after thermo-mechanical fatigue and three-point bending test. The Tukey's honestly significant difference (HSD) test was used for multiple *post-hoc* comparisons when the overall ANOVA was significant [30,52–55].

Data were processed using the Statistical Package for the Social Sciences (software v.22) (SPSS/PC+, Inc.; Chicago, IL, USA) taking in advance the cut-off level for statistical significance at $\alpha = 0.05$ [53,56–58].

2.7. Evaluation of the failure modes

The broken samples were gently ultrasonicated for 1 h and air dried. The fracture patterns were visually inspected and evaluated under a stereomicroscope (Nikon SMZ-10; Nikon Co.) at 15× magnifications [34]. Failure types were classified as ductile (D: plastic deformation and tearing of the material) or brittle (B: no significant plastic deformation, showing flat surfaces that fit each other along the fracture line) [36]. Percentages of ductile and brittle fractures were calculated for each experimental group. The fracture location was also recorded [34].

3. Results

3.1. Material characterization

The average grain size values of YZ (0.72 μm ± 0.08 μm) and ZZ (0.75 μm ± 0.13 μm) restorations were statistically comparable to those of LZ (0.82 μm ± 0.11 μm) and LM (0.65 μm ± 0.06 μm) samples ($p > 0.05$). However, monolithic LM pieces (Fig. 1a) showed significantly lower grain size values than did LZ specimens (Fig. 1b) ($p = 0.001$).

3.2. Fracture load

All samples preserved their integrity after thermo-mechanical fatigue. Neither cracks nor fracture failures were detected within the restorations after cycling.

Mean (SD) VF values recorded in the tested groups are outlined in Table 1. The control group (MC) achieved the significantly highest VF values ($p = 0.0001$), while the three bi-layered zirconia groups evidenced no significant differences to each other concerning this parameter ($p > 0.05$).

Mean (SD) TF values are summarised in Table 2. The MC group attained the significantly highest TF values ($p = 0.0001$), while the zirconia groups demonstrated statistically comparable TF values ($p > 0.05$).

Within each bi-layered group, TF values were significantly higher than VF values ($p < 0.05$).

3.3. Failure modes

Representative images of the study groups after the three-point bending test are presented in Figs. 2–4.

In the bi-layered groups (MC, LZ, YZ, and ZZ), the veneering porcelain always failed before the total breakage of the specimens resulting in delamination of the ceramic coating except for the YZ group, in which three FDPs (30%) showed no previous veneering fracture. An irregular surface was discovered where the veneering material detached from the cores, with porcelain still bonded to the structures in all cases (Fig. 2).

A ductile failure starting at the occlusal aspect of the connectors was identified in all of the MC samples (Table 2) such that the higher the plastic deformation at the ultimate load, the rougher the fracture margins. Owing to the jagged topography of the broken frameworks, the fragments of MC samples could not be completely repositioned
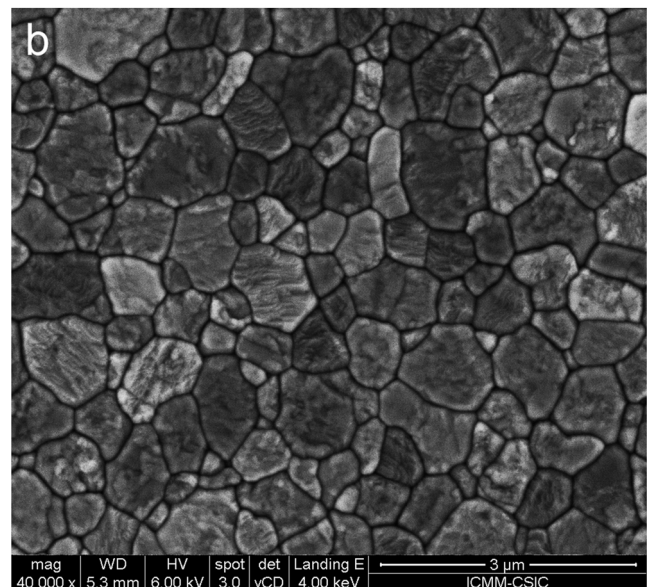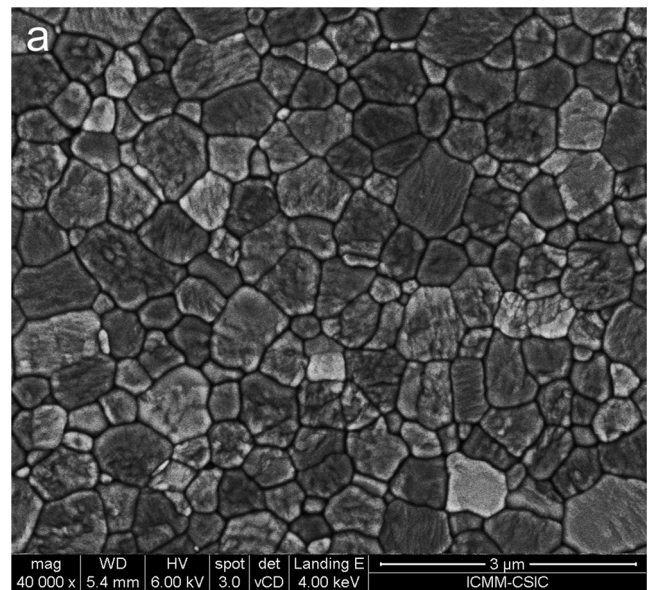

Fig. 1. FEG-SEM images of material characterization (40,000× magnifications; WD: 5.4 mm; HV: 6.00 kV; spot: 3.0; det: vCD; landing E: 4.00 keV; bar: 3 μm). (a) LM specimen showing small grains. (b) LZ sample containing bigger grains than did LM pieces.

(Fig. 3).

Conversely, all of the zirconia specimens showed a brittle failure (Table 2), in which the fragments perfectly fitted to each other along the fracture line (Fig. 4). The breakage of LZ, YZ, and ZZ pieces mainly (90%) started at the cervical aspect of the connectors and was diagonally propagated toward the occlusal surface of the pontic through the loading point. In the remaining 10% of cases, the fracture arose at the axial and occlusal surfaces of the retainers. Unlike what happened in the control group, the total failure of LZ, YZ and ZZ restorations occurred quickly after the initial fracture.

Finally, all of the LM monolithic restorations exhibited brittle vertical fractures (Table 2). In this group, the cracks originating within the occlusal surface of the FDPs instantaneously entered the complete thickness of the prostheses in a catastrophic manner. Hence, all of these specimens were broken in half, lengthwise (Fig. 4).

Table 1
Mean (SD) fracture load values of the veneering ceramic recorded in the tested groups (*Newton*).

Fracture load values of the veneering ceramic (VF)							
Group	N	Mean	SD	Min	Max	% exceeding 700 N	% exceeding 1000 N
MC	10	3008.69 **	193.65	2743.02	3315.34	100	100
LZ	10	927.96 *	330.29	487.06	1548.40	80	50
YZ	10	969.90 *	663.39	402.78	2086.42	90	60
ZZ	10	928.35 *	391.04	301.84	1451.38	80	40

MC: Metal-ceramic (Group 1, control). **LZ**: Lava Zirconia (Group 2, bi-layered all-ceramic restorations). **YZ**: VITA In-Ceram YZ (Group 4, bi-layered all-ceramic restorations). **ZZ**: IPS e-max ZirCAD (Group 5, bi-layered all-ceramic restorations).
SD: Standard deviation. **Min**: Minimum value recorded per group. **Max**: Maximum value recorded per group.
Different numbers of asterisks in the *Mean* column indicate significant differences between groups ($p = 0.0001$).

Table 2
Mean (SD) total fracture load values recorded in the tested groups (*Newton*).

Total fracture load values (TF)							
Group	N	Mean	SD	Min	Max	D (%)	B (%)
MC	10	7958.15 **	932.96	6658.12	9414.86	100	0
LZ	10	1966.27 *	397.86	1255.38	2473.52	0	100
LM	10	2181.67 *	303.99	1691.48	2602.88	0	100
YZ	10	1869.84 *	211.36	1535.66	2086.42	0	100
ZZ	10	1908.74 *	118.90	1668.94	2044.28	0	100

MC: Metal-ceramic (Group 1, control). **LZ**: Lava Zirconia (Group 2, bi-layered all-ceramic restorations). **LM**: Lava Plus (Group 3, monolithic all-ceramic restorations). **YZ**: VITA In-Ceram YZ (Group 4, bi-layered all-ceramic restorations). **ZZ**: IPS e-max ZirCAD (Group 5, bi-layered all-ceramic restorations).
SD: Standard deviation. **Min**: Minimum value recorded per group. **Max**: Maximum value recorded per group. **D (%)**: Percentages of ductile failures. **B (%)**: Percentages of brittle failures.
Different numbers of asterisks in the *Mean* column indicate significant differences between groups ($p = 0.0001$).

Fig. 2. ZZ restoration showing delamination of the veneering ceramic at the buccal surface of the intermediate pontic. A stepped and rough topography is discovered.

4. Discussion

In this experiment, all FDPs survived the fatigue after simulated clinical temperatures and chewing forces, suggesting a stable

Fig. 3. Ductile fracture of a MC sample (at the connector). The plastic deformation impedes the exact fitting of the broken parts along the fracture line.

Fig. 4. LM piece fractured in half, lengthwise. The flat broken surface reveals a brittle failure.

performance of all tested groups before three-point bending evaluation. Microstructure variations of natural teeth preparations and other subsequent variables related to impression and pouring techniques [33,59] were avoided by using standardised metallic models [39,40,60]. To enable the results' comparisons, the restorations were fabricated following the manufacturer's instructions and were luted in a standard fashion. The remaining steps of the study protocol were also strictly completed by the same technician under our supervision.

Our results require the rejection of the first null hypothesis, as MC specimens registered significantly higher VF values than did their bi-layered zirconia-based counterparts (Table 1) as concluded in previous investigations [54,61]. This may be due to the higher capacity of mechanical stress absorption of metallic structures that may limit the propagation of cracks [62]. On the contrary, zirconia cores are brittle and the risk of tension in the veneering porcelain is higher [63]. This may explain that, regardless of being coated with different brands of aesthetic glass ceramic, LZ, YZ, and ZZ groups achieved comparable VF values (Table 1). It is worth remarking at this point that the VF values represent the failure of the restorations because of major chipping. Likewise in a former research [61], the MC group also demonstrated significantly higher TF values than did the zirconia groups (Table 2). Our MC values were similar to those reported by Castillo de Oyagüe et al. [36] for 3-unit Co-Cr un-veneered structures. Further work is necessary to ascertain whether such differences will translate into significant clinical outcomes.

Very few investigations rely on the comparison of the mechanical behaviour of different zirconia systems, mainly in the case of posterior FDPs [3,34,52,61] and the inconsistency of their results may be related to their methodological variability. In our study, the monolithic LM pieces achieved the highest TF values of the four zirconia groups. This could be expected, as zirconia has higher strength and toughness compared to other ceramics because of the tetragonal-to-monoclinic phase transformation that occurs in this material [10–12]. However, such differences were not significant (Table 2). Although an investigation in crowns [54] recorded significantly higher fracture strength for the monolithic pieces with respect to the bi-layer configuration, the absence of related studies on multiunit prostheses impede precise comparisons.

On the one hand, even though the intensity of the masticatory forces greatly varies depending on gender, age, and location in the dental arch, the physiologic loads for healthy young adults are estimated to be around 700 N [61,64]. Our mean VF and TF values were greater (Tables 1 and 2) meaning that, in absence of parafunctions, all of the groups tested could withstand clinical chewing forces in terms of average fracture load. Nonetheless, given that parafunctional forces can be as high as 1000 N, dental restorations should ideally support this limit [7,16]. The TF data of all zirconia groups almost doubled this value (Table 2). However, in parafunctional patients, bi-layered zirconia restorations could entail an uncertain risk of chipping as the minimum VF values of these groups were situated around 400 N (Table 1). This should be further evaluated before recommending these systems for widespread use. Conversely, given that LM pieces attained TF values of 2181.67 $\pm$ 303.99 N (Table 2), monolithic zirconia could be suggested as an alternative material based on our study findings. Nevertheless, although more investigations are needed, the MC group may be still considered as the most resistant in light of its TF ad VF values (Tables 1 and 2).

On the other hand, there are other potential factors that may affect the in vitro fracture resistance of zirconia restorations, such as the ceramic microstructure, manufacturing system, surface treatment, veneering technique, cementation [48], storage, and fatigue test [8]. The grain size was assessed because of the known association between microstructure and mechanical properties of a substrate such that the lower the grain size, the higher the mechanical strength [65]. Our grain size measures were consistent with the manufacturers' specifications. LM samples showed significantly lower grain size than did LZ restorations (Fig. 1), and this was the only significant difference in microstructure detected among the zirconia groups. However, although the monolithic pieces registered the highest TF values of all zirconia FDPs, such differences were not significant (Table 2). Larger samples are thus required to better explore the correlation between grain size and fracture resistance.

Slightly different destruction styles were observed even within each tested group; which may be related to the high SD values recorded in this investigation (Tables 1 and 2). Nonetheless, due to the high elastic modulus of Co-Cr dental alloys [66], the cores of all MC specimens suffered an evident plastic deformation prior to fracture under static loading. According to previous research [36], this was translated into ductile failures of the frames that were always initiated at the occlusal aspect of the connectors (Table 2, Fig. 3). All of the LZ, YZ, and ZZ samples experienced brittle failures (Table 2) that usually (90%) started at the gingival embrasure of the connectors; which is consistent with the literature [8,19,34,47,48,67]. Such smooth fracture line progressed transversely to the occlusal surface of the pontic, as described [8,13,19,34,67]. The connector size and design may therefore have an important effect on the fracture resistance and longevity of zirconia-based FDPs [10,13,26,34,67,68]. For this reason, most authors recommend connector's dimensions of 6–9 mm^2, such as those used in our investigation [8,19,26,34,69]. In few exceptions (10%), the fracture started at the axial and occlusal surfaces of the retainers. Undoubtedly, a sufficient thickness of the ceramic material in all of these areas must

be provided in order to ensure an adequate functional life of the restoration. In contrast, all of the LM restorations exhibited brittle abrupt fractures (Table 2) that were initiated at the occlusal surface of the specimens and progressed vertically across the ceramic width (Fig. 4). The absence of veneering/core interphase together with the lower grain size (Fig. 1) could have influenced the failure pattern of these restorations, which should be further investigated.

Finally, the second null hypothesis was refuted because the average VF values of all bi-layered systems tested were significantly lower than their respective TF values. In this cases, in which a subjacent chipping problem is disclosed, the superior mechanical properties of the zirconia frames [70] and the thermal behaviour at the core/veneer adhesive interface should be taken into account [2,71]. A mismatch in the coefficients of thermal expansion (CTEs) of the veneering and core materials may lead to a decrease in bond strength or even to spontaneous debonding during firing [2,72]. In our experiment, the CTEs of the veneering ceramics were lower than those of the metallic (MC) and zirconia (LZ, YZ, and ZZ) cores as deduced from the data sheets. This might have produced compression loads on the ceramic coating and compensatory tensional forces on the structure surfaces [71] such that, overall, the veneering material fractured before the core in bi-layered samples (Fig. 2). This is consistent with a study made in crowns [61]. Exceptionally, no previous fracture of the veneering porcelain was registered in 30% of the YZ specimens. This might point to a slightly minor risk of chipping, as such FDPs also exhibited the highest VF values of the bi-layered zirconia groups (although no significant differences were reached) (Table 1).

Anatomically designed structures, such as ours, have been reported to reduce chipping, as they ensure an uniform thickness of the veneering material [3,48,70]. Actually, while the failure of metal-ceramic FDPs has been rated to be independent of the core shape, this remains uncertain in the case of all-ceramic prostheses, which requires further analysis [73]. The chipping also depends on the uneven masticatory loads presented *in vivo*, which in turn vary according to the type of food to be triturated by the posterior teeth [54]. This represents an unavoidable limitation in any *in vitro* study. Hence, future research on the prevention of chipping should consider, among others, the frame elastic modulus, the disparities of the veneering and core CTEs, the structure design, and the effect of functional and parafunctional forces. A deeper analysis of the destruction styles and their correlation with the fracture load should be conducted in future investigations. Concerning the high SD values recorded in our experiment, as previously stated, *'in spite of great care being taken to standardize procedures, any process involving manipulation of diverse materials is technique sensitive and individual aberrations can occur despite the technician experience'* [36]. Actually, the ceramic veneering of all bi-layered structures was manually performed. The classic lost-wax technique used for casting the MC frameworks is quite imprecise for being a handmade work that requires a number of materials [36]. The CAD/CAM tools utilised for fabricating the all-ceramic structures (LZ, YZ, and ZZ) and the LM monolithic restorations were driven by a technician, which may increase the risk of dimensional variations. Moreover, intraoral scanning still has technology-related errors [8,18,19,60]. The relatively small sample size may have also contributed to the high standard deviations. In the light of all this, despite the existence of many related studies with comparable sample sizes and SD [74–77], future research should try to overcome such study limitations.

There conclusions emerged under the tested conditions: *(1)* all groups demonstrated clinically acceptable fracture load values. Zirconia restorations performed equally, achieving lower fracture resistance than did metal-ceramic FDPs; *(2)* the weakest area was mainly located at the connectors, which should be reinforced; and *(3)* monolithic zirconia may be recommended for solving the chipping problem, as it showed fracture resistance values that were comparable to those of the bi-layered zirconia systems.

Most patients demand treatment if major chipping (clinically

considered as 'failure' in our experiment), happens on their veneering ceramic without waiting for the framework to break. Hence, in future clinical investigations, the TF value of monolithic zirconia pieces might be conceptually compared with the VF value of the bi-layered restorations, as 'failure load' of aesthetic FDPs. In such case, an obvious difference might appear between monolithic zirconia and the others. However, even though our results could be further interpreted for their clinical significance, this *in vitro* research aimed to evaluate and compare the original characteristics of the materials, and, thus, the LM group was not analysed for VF, but only for TF. Moreover, in the LM group, the cracks originating within the occlusal surface of the FDPs instantaneously entered the complete thickness of the prostheses so that all of these specimens were quickly broken in half, lengthwise. Notwithstanding that the relevance of laboratory investigations consists of preventing from undesirable outcomes [78], our findings should be reasonably extrapolated to the clinical environment, as dental restorations are subjected to complex thermo-biomechanical interactions and personal subjective interpretations that far exceed those incorporated in *in vitro* tests [40,43,79]. Finally, newly introduced indices on the oral health-related quality of life (OHRQoL) [80,81] should be validated and systematically applied for candidates of FDPs, which may add potential benefits for decision-making.

Declarations of interest

Acknowledgements

References

[1] J.A. Skupien, M.S. Cenci, N.J. Opdam, C.M. Kreulen, M.C. Huysmans, T. Pereira-Cenci, Crown *vs.* composite for post-retained restorations: a randomized clinical trial, J. Dent. 48 (2016) 34–39, http://dx.doi.org/10.1016/j.jdent.2016.03.007.

[2] H.A. Vidotti, J.R. Pereira, E. Insaurralde, A.L. de Almeida, A.L. do Valle, Thermo and mechanical cycling and veneering method do not influence Y-TZP core/veneer interface bond strength, J. Dent. 41 (2013) 307–312, http://dx.doi.org/10.1016/j.jdent.2012.12.001.

[3] V. Rodríguez, R. Castillo-Oyagüe, C. López-Suárez, E. Gonzalo, J. Peláez, M.J. Suárez-García, Fracture load before and after veneering zirconia posterior fixed dental prostheses, J. Prosthodont. 25 (2016) 550–556, http://dx.doi.org/10.1111/jopr.12357.

[4] E. Gonzalo, M.J. Suárez, B. Serrano, J.F. Lozano, Comparative analysis of two measurement methods for marginal fit in metal-ceramic and zirconia posterior FPDs, Int. J. Prosthodont. 22 (2009) 374–377.

[5] R.C. Oyagüe, F. Monticelli, M. Toledano, E. Osorio, M. Ferrari, R. Osorio, Effect of water aging on microtensile bond strength of dual-cured resin cements to pre-treated sintered zirconium-oxide ceramics, Dent. Mater. 25 (2009) 392–399, http://dx.doi.org/10.1016/j.dental.2008.09.002.

[6] J. Peláez, P.G. Cogolludo, B. Serrano, J.F. Lozano, M.J. Suárez, A prospective evaluation of zirconia posterior fixed dental prostheses: three-year clinical results, J. Prosthet. Dent. 107 (2012) 373–379, http://dx.doi.org/10.1016/S0022-3913(12)60094-8.

[7] J. Tinschert, D. Zwez, R. Marx, K.J. Anusavice, Structural reliability of alumina-feldspar-, leucite-, mica- and zirconia-based ceramics, J. Dent. 28 (2000) 529–535.

[8] A.J. Raigrodski, Contemporary materials and technologies for all-ceramic fixed partial dentures: a review of the literature, J. Prosthet. Dent. 92 (2004) 557–562.

[9] J.R. Kelly, I. Denry, Stabilized zirconia as a structural ceramic: an overview, Dent. Mater. 24 (2008) 289–298.

[10] F. Komine, M.B. Blatz, H. Matsumura, Current status of zirconia-based fixed restorations, J. Oral Sci. 52 (2010) 531–539.

[11] R.C. Garvie, R.H. Hannink, R.T. Pascoe, Ceramic steel? Nature 258 (1975) 703–704, http://dx.doi.org/10.1038/258703a0.

[12] C. Piconi, G. Maccauro, Zirconia as a ceramic biomaterial, Biomaterials 20 (1999) 1–25.

[13] C. Larsson, L. Holm, N. Lövgren, Y. Kokubo, P. Vult von Steyern, Fracture strength of four-unit Y-TZP FPD cores designed with varying connector diameter. An in-vitro study, J. Oral Rehabil. 34 (2007) 702–709.

[14] P.C. Guess, A. Kulis, S. Witkowski, M. Wolkewitz, Y. Zhang, J.R. Strub, Shear bond strengths between different zirconia cores and veneering ceramics and their susceptibility to thermocycling, Dent. Mater. 24 (2008) 1556–1567, http://dx.doi.org/10.1016/j.dental.2008.03.028.

[15] I. Sailer, J. Gottnerb, S. Kanelb, C.H. Hammerle, Randomized controlled clinical trial of zirconia-ceramic and metal-ceramic posterior fixed dental prostheses: a 3-year follow-up, Int. J. Prosthodont. 22 (2009) 553–560.

[16] J. Schmitt, S. Holst, M. Wichmann, S. Reich, M. Gollner, J. Hamel, Zirconia posterior fixed partial dentures: a prospective clinical 3-year follow-up, Int. J. Prosthodont. 22 (2009) 597–603.

[17] B. Al-Amleh, K. Lyons, M. Swain, Clinical trials in zirconia: a systematic review, J. Oral Rehabil. 37 (2010) 641–652, http://dx.doi.org/10.1111/j.1365-2842.2010.02094.

[18] J. Pelaez, P.G. Cogolludo, B. Serrano, J.F. Serrano, M.J. Suarez, A four-year prospective clinical evaluation of zirconia and metal-ceramic posterior fixed dental prostheses, Int. J. Prosthodont. 25 (2012) 451–458.

[19] R. Agustín-Panadero, J.L. Román-Rodríguez, A. Ferreiroa, M.F. Solá-Ruíz, A. Fons-Font, Zirconia in fixed prosthesis. A literature review, J. Clin. Exp. Dent. 6 (2014) 66–73, http://dx.doi.org/10.4317/jced.51304.

[20] K.J. Anusavice, Standardizing failure, success, and survival decisions in clinical studies of ceramic and metal-ceramic fixed dental prostheses, Dent. Mater. 28 (2012) 102–111, http://dx.doi.org/10.1016/j.dental.2011.09.012.

[21] M.N. Aboushelib, C.J. Kleverlaan, A.J. Feilzer, Microtensile bond strength of different components of core veneered all-ceramic restorations. Part II: Zirconia veneering ceramics, Dent. Mater. 22 (2006) 857–863.

[22] F. Beuer, J. Schweiger, M. Eichberger, H.F. Kappert, W. Gernet, D. Edelhoff, High-strength CAD/CAM-fabricated veneering material sintered to zirconia copings-a new fabrication mode for all-ceramic restorations, Dent. Mater. 25 (2009) 121–128, http://dx.doi.org/10.1016/j.dental.2008.04.019.

[23] A.K. Costa, A.L. Borges, G.J. Fleming, O. Addison, The strength of sintered and adhesively bonded zirconia/veneer-ceramic bilayers, J. Dent. 42 (2014) 1269–1276, http://dx.doi.org/10.1016/j.jdent.2014.08.001.

[24] H.K. Kim, S.H. Kim, J.B. Lee, J.S. Han, I.S. Yeo, Effect of polishing and glazing on the color and spectral distribution of monolithic zirconia, J. Adv. Prosthodont. 5 (2013) 296–304, http://dx.doi.org/10.4047/jap.2013.5.3.296.

[25] J.R. Kelly, Clinically relevant approach to failure testing of all-ceramic restorations, J. Prosthet. Dent. 81 (1999) 652–661.

[26] A.R. Studart, F. Filser, P. Kocher, H. Lüthy, L.J. Gauckler, Mechanical and fracture behavior of veneer-framework composites for all-ceramic dental bridges, Dent. Mater. 23 (2007) 115–123.

[27] E. Tsalouchou, M.J. Cattell, J.C. Knowles, P. Pittayachawan, A. McDonald, Fatigue and fracture properties of yttria partially stabilized zirconia crown systems, Dent. Mater. 24 (2008) 308–318.

[28] P.C. Guess, S. Schultheis, M. Wolkewitz, Y. Zhang, J.R. Strub, Influence of preparation design and ceramic thicknesses on fracture resistance and failure modes of premolar partial coverage restorations, J. Prosthet. Dent. 110 (2013) 264–273, http://dx.doi.org/10.1016/S0022-3913(13)60374-1.

[29] P.C. Guess, T. Vagkopoulou, Y. Zhang, M. Wolkewitz, J.R. Strub, Marginal and internal fit of heat pressed versus CAD/CAM fabricated all-ceramic onlays after exposure to thermo-mechanical fatigue, J. Dent. 42 (2014) 199–209, http://dx.doi.org/10.1016/j.jdent.2013.10.002.

[30] A.G. Turk, M. Ulusoy, M. Yuce, H. Akin, Effect of different veneering techniques on the fracture strength of metal and zirconia frameworks, J. Adv. Prosthodont. 7 (2015) 454–459, http://dx.doi.org/10.4047/jap.2015.7.6.454.

[31] E. Gonzalo, M.J. Suárez, B. Serrano, J.F. Lozano, Marginal fit of Zirconia posterior fixed partial dentures, Int. J. Prosthodont. 21 (2008) 398–399.

[32] R. Castillo de Oyagüe, M.I. Sánchez-Jorge, A. Sánchez Turrión, F. Monticelli, M. Toledano, R. Osorio, Influence of CAM vs. CAD/CAM scanning methods and finish line of tooth preparation in the vertical misfit of zirconia bridge structures, Am. J. Dent. 22 (2009) 79–83.

[33] R. Castillo Oyagüe, M.I. Sánchez-Jorge, A. Sánchez Turrión, Influence of CAD/CAM scanning method and tooth-preparation design on the vertical misfit of zirconia crown copings, Am. J. Dent. 23 (2010) 341–346.

[34] C. López-Suárez, E. Gonzalo, J. Peláez, V. Rodríguez, M.J. Suárez, Fracture resistance and failure mode of posterior fixed dental prostheses fabricated with two zirconia CAD/CAM systems, J. Clin. Exp. Dent. 7 (2015) 250–253, http://dx.doi.org/10.4317/jced.52067.

[35] C. Lopez-Suarez, E. Gonzalo, J. Pelaez, B. Serrano, M.J. Suarez, Marginal vertical discrepancies of monolithic and veneered zirconia and metal-ceramic three-unit posterior fixed dental prostheses, Int. J. Prosthodont. 29 (2016) 256–258, http://dx.doi.org/10.11607/ijp.4541.

[36] R. Castillo-de Oyagüe, R. Osorio, C. Lynch, A. Gilmour, M. Toledano, Effect of alloy type and casting technique on the fracture strength of implant-cemented structures, Med. Oral Patol. Oral Cir. Bucal 16 (2011) 619–625.

[37] M.J. Kim, J.S. Ahn, J.H. Kim, H.Y. Kim, W.C. Kim, Effects of the sintering conditions of dental zirconia ceramics on the grain size and translucency, J. Adv. Prosthodont.

C. López-Suárez et al.

5 (2013) 161–166, http://dx.doi.org/10.4047/jap.2013.5.2.161.

[38] S.S. Scherrer, M. Cattani-Lorente, E. Vittecoq, F. de Mestral, J.A. Griggs, H.W. Wiskott, Fatigue behavior in water of Y-TZP zirconia ceramics after abrasion with 30 μm silica-coated alumina particles, Dent. Mater. 27 (2011) 28–42, http://dx.doi.org/10.1016/j.dental.2010.10.003.

[39] M. Jesús Suárez, J.F. Lozano, M. Paz Salido, F. Martínez, Marginal fit of titanium metal-ceramic crowns, Int. J. Prosthodont. 18 (2005) 390–391.

[40] P. Gómez-Cogolludo, R. Castillo-Oyagüe, C.D. Lynch, M.J. Suárez-García, Effect of electric arc, gas oxygen torch and induction melting techniques on the marginal accuracy of cast base-metal and noble metal-ceramic crowns, J. Dent. 41 (2013) 826–831, http://dx.doi.org/10.1016/j.jdent.2013.06.007.

[41] R.C. Oyagüe, A.S. Turrión, M. Toledano, F. Monticelli, R. Osorio, In vitro vertical misfit evaluation of cast frameworks for cement-retained implant-supported partial prostheses, J. Dent. 37 (2009) 52–58, http://dx.doi.org/10.1016/j.jdent.2008.09.007.

[42] R.C. Oyagüe, A. Sánchez-Turrión, J.F. López-Lozano, J. Montero, A. Albaladejo, M.J. Suárez-García, Evaluation of fit of cement-retained implant-supported 3-unit structures fabricated with direct metal laser sintering and vacuum casting techniques, Odontology 100 (2012) 249–253, http://dx.doi.org/10.1007/s10266-011-0050-1.

[43] R. Castillo-Oyagüe, C.D. Lynch, A.S. Turrión, J.F. López-Lozano, D. Torres-Lagares, M.J. Suárez-García, Misfit and microleakage of implant-supported crown copings obtained by laser sintering and casting techniques, luted with glass-ionomer, resin cements and acrylic/urethane-based agents, J. Dent. 41 (2013) 90–96, http://dx.doi.org/10.1016/j.jdent.2012.09.014.

[44] R.C. Oyagüe, A. Sánchez-Turrión, J.F. López-Lozano, M.J. Suárez-García, Vertical discrepancy and microleakage of laser-sintered and vacuum-cast implant-supported structures luted with different cement types, J. Dent. 40 (2012) 123–130, http://dx.doi.org/10.1016/j.jdent.2011.11.007.

[45] S. Taufall, M. Eichberger, P.R. Schmidlin, B. Stawarczyk, Fracture load and failure types of different veneered polyetheretherketone fixed dental prostheses, Clin. Oral Investig. 20 (2016) 2493–2500, http://dx.doi.org/10.1007/s00784-016-1777-4.

[46] I. Krejci, F. Lutz, In-vitro test results of the evaluation of dental restoration systems. Correlation with in-vivo results, Schweiz Monatsschr. 100 (1990) 1445–1449.

[47] M. Guazzato, K. Proos, G. Sara, M.V. Swain, Strength reliability, and mode of fracture of bilayered porcelain/core ceramics, Int. J. Prosthodont. 17 (2004) 142–149.

[48] A. Sundh, M. Molin, G. Sjögren, Fracture resistance of yttrium oxide partially-stabilized zirconia all-ceramic bridges after veneering and mechanical fatigue testing, Dent. Mater. 21 (2005) 476–482.

[49] A. Attia, M. Kern, Influence of cyclic loading and luting agents on the fracture load of two all-ceramic crown systems, J. Prosthet. Dent. 92 (2004) 551–556.

[50] P. Gehrke, D. Johannson, C. Fischer, B. Stawarczyk, F. Beuer, In vitro fatigue and fracture resistance of one- and two-piece CAD/CAM zirconia implant abutments, Int. J. Oral Maxillofac. Implants 30 (2015) 546–554, http://dx.doi.org/10.11607/jomi.3942.

[51] P.R. Schmidlin, B. Stawarczyk, D. DeAbreu, A. Bindl, A. Ender, I.P. Ichim, Fracture resistance of endodontically treated teeth without ferrule using a novel H-shaped short post, Quintessence Int. 46 (2015) 97–108, http://dx.doi.org/10.3290/j.qi.a32634.

[52] C. Yildiz, B.A. Vanlioğlu, B. Evren, A. Uludamar, Y.K. Ozkan, Marginal-internal adaptation and fracture resistance of CAD/CAM crown restorations, Dent. Mater. J. 32 (2013) 42–47.

[53] F.A. Amir Rad, F.G. Succaria, S.M. Morgano, Fracture resistance of porcelain veneered zirconia crowns with exposed lingual zirconia for anterior teeth after thermal cycling: an in vitro study, Saudi Dent. J. 27 (2015) 63–69, http://dx.doi.org/10.1016/j.sdentj.2014.11.005.

[54] D.P. Lameira, W.A. Buarque e Silva, F. Andrade e Silva, G.M. De Souza, Fracture strength of aged monolithic and bilayer zirconia-based crowns, Biomed. Res. Int. 2015 (2015) 418641, http://dx.doi.org/10.1155/2015/418641.

[55] L.L. Seow, C.G. Toh, N.H. Wilson, Strain measurements and fracture resistance of endodontically treated premolars restored with all-ceramic restorations, J. Dent. 43 (2015) 126–132, http://dx.doi.org/10.1016/j.jdent.2014.10.001.

[56] V. Preis, A. Kammermeier, G. Handel, M. Rosentritt, In vitro performance of two-piece zirconia implant systems for anterior application, Dent. Mater. 32 (2016) 765–774, http://dx.doi.org/10.1016/j.dental.2016.03.028.

[57] M.H. Nicolaisen, G. Bahrami, S. Finlay, F. Isidor, Comparison of fatigue resistance and failure modes between metal-ceramic and all-ceramic crowns by cyclic loading in water, J. Dent. 42 (2014) 1613–1620, http://dx.doi.org/10.1016/j.jdent.2014.08.013.

[58] A. Hannigan, C.D. Lynch, Statistical methodology in oral and dental research: pitfalls and recommendations, J. Dent. 41 (2013) 385–392, http://dx.doi.org/10.1016/j.jdent.2013.02.013.

[59] C.D. Lynch, P.F. Allen, Quality of communication between dental practitioners and dental technicians for fixed prosthodontics in Ireland, J. Oral Rehabil. 32 (2005) 901–905.

[60] E. Gonzalo, M.J. Suárez, B. Serrano, J.F. Lozano, A comparison of the marginal vertical discrepancies of zirconium and metal ceramic posterior fixed dental prostheses before and after cementation, J. Prosthet. Dent. 102 (2009) 378–384, http://dx.doi.org/10.1016/S0022-3913(09)60198-0.

[61] R. Augstin-Panadero, A. Fons-Font, J.L. Roman-Rodriguez, M. Granell-Ruiz, J. del Rio-Highsmith, M.F. Sola-Ruiz, Zirconia versus metal: a preliminary comparative analysis of ceramic veneer behaviour, Int. J. Prosthodont. 25 (2012) 294–300.

[62] R. Morena, G.M. Beaudreau, P.E. Lockwood, A.L. Evans, C.W. Fairhurst, Fatigue of dental ceramics in a simulated oral environment, J. Dent. Res. 65 (1986) 993–997.

[63] M. de Kler, N. de Jager, M. Meegdes, J.M. van der Zel, Influence of thermal expansion mismatch and fatigue loading on phase changes in porcelain veneered Y-TZP zirconia discs, J. Oral Rehabil. 34 (2007) 841–847.

[64] V.F. Ferrario, C. Sforza, G. Zanotti, G.M. Tartaglia, Maximal bite forces in healthy young adults as predicted by surface electromyography, J. Dent. 32 (2004) 451–457.

[65] K. Nakamura, E. Adolfsson, P. Milleding, T. Kanno, U. Örtengren, Influence of grain size and veneer firing process on the flexural strength of zirconia ceramics, Eur. J. Oral Sci. 120 (2012) 249–254, http://dx.doi.org/10.1111/j.1600-0722.2012.00958.x.

[66] H.F. Morris, Properties of cobalt-chromium metal ceramic alloys after heat treatment, J. Prosthet. Dent. 63 (1990) 426–433.

[67] A. Sundh, G. Sjögren, Fracture resistance of all-ceramic zirconia bridges with differing phase stabilizers and quality of sintering, Dent. Mater. 22 (2006) 778–784.

[68] R.R. Wang, C.L. Lu, G. Wang, D.S. Zhang, Influence of cyclic loading on the fracture toughness and load bearing capacities of all-ceramic crowns, Int. J. Oral Sci. 6 (2014) 99–104, http://dx.doi.org/10.1038/ijos.2013.94.

[69] D. Suttor, K. Bunke, S. Hoescheler, H. Hauptmann, G. Hertlein, LAVA-the system for all-ceramic ZrO2 crown and bridge frameworks, Int. J. Comput. Dent. 4 (2001) 195–206.

[70] M. Borba, M.D. de Araújo, E. de Lima, H.N. Yoshimura, P.F. Cesar, J.A. Griggs, A. Della Bona, Flexural strength and failure modes of layered ceramic structures, Dent. Mater. 27 (2011) 1259–1266, http://dx.doi.org/10.1016/j.dental.2011.09.008.

[71] M.V. Swain, Unstable cracking (chipping) of veneering porcelain on all-ceramic dental crowns and fixed partial dentures, Acta Biomater. 5 (2009) 1668–1677, http://dx.doi.org/10.1016/j.actbio.2008.12.016.

[72] A. Saito, F. Komine, M.B. Blatz, H. Matsumura, A comparison of bond strength of layered veneering porcelains to zirconia and metal, J. Prosthet. Dent. 104 (2010) 247–257, http://dx.doi.org/10.1016/S0022-3913(10)60133-3.

[73] Y.R. Wei, X.D. Wang, Q. Zhang, X.X. Li, M.B. Blatz, Y.T. Jian, K. Zhao, Clinical performance of anterior resin-bonded fixed dental prostheses with different framework designs: a systematic review and meta-analysis, J. Dent. 47 (2016) 1–7, http://dx.doi.org/10.1016/j.jdent.2016.02.003.

[74] M. Ferrari, A. Giovannetti, M. Carrabba, G. Bonadeo, C. Rengo, F. Monticelli, A. Vichi, Fracture resistance of three porcelain-layered CAD/CAM zirconia frame designs, Dent. Mater. 30 (2014) e163–168, http://dx.doi.org/10.1016/j.dental.2014.02.004.

[75] B. Baladhandayutham, N.C. Lawson, J.O. Burgess, Fracture load of ceramic restorations after fatigue loading, J. Prosthet. Dent. 114 (2015) 266–271, http://dx.doi.org/10.1016/j.prosdent.2015.03.006.

[76] V. Nazari, S. Ghodsi, M. Alikhasi, M. Sahebi, A.R. Shamshiri, Fracture strength of three-unit implant supported fixed partial dentures with excessive crown height fabricated from different materials, J. Dent. (Tehran) 13 (2016) 400–406.

[77] L.M. Soares, C. Soares, M.E. Miranda, R.T. Basting, Influence of core-veneer thickness ratio on the fracture load and failure mode of zirconia crowns, J. Prosthodont. (2017), http://dx.doi.org/10.1111/jopr.12601 in press.

[78] C.D. Lynch, Excellence in oral & dental research, J. Dent. 40 (2012) 1–2, http://dx.doi.org/10.1016/j.jdent.2011.12.001.

[79] A.A. Xible, R.R. de Jesus Tavarez, R. de Araujo Cdos, W.C. Bonachela, Effect of silica coating and silanization on flexural and composite-resin bond strengths of zirconia posts: an in vitro study, J. Prosthet. Dent. 95 (2006) 224–229.

[80] R. Castillo-Oyagüe, M.J. Suárez-García, C. Perea, J.D. Río, C.D. Lynch, E. Gonzalo, D. Torres-Lagares, A. Preciado, Validation of a new, specific, complete, and short OHRQoL scale (QoLFAST-10) for wearers of implant overdentures and fixed-detachable hybrid prostheses, J. Dent. 49 (2016) 22–32, http://dx.doi.org/10.1016/j.jdent.2016.04.011.

[81] C. Perea, A. Preciado, J.D. Río, C.D. Lynch, A. Celemín, R. Castillo-Oyagüe, Oral aesthetic-related quality of life of muco-supported prosthesis and implant-retained overdenture wearers assessed by a new, short, specific scale (QoLDAS-9), J. Dent. 43 (2015) 1337–1345, http://dx.doi.org/10.1016/j.jdent.2015.08.010.