AMERICAN COLLEGE OF PROSTHODONTISTS
Your smile. Our specialty.

Influence of Digital Technologies and Framework Design on the Load to Fracture of Co-Cr Posterior Fixed Partial Denture Frameworks

Celia Tobar, DDS, MS, PhD, Veronica Rodriguez, DDS, MS, PhD, Carlos Lopez-Suarez, DDS, MS, PhD, Jesus Pelaez, DDS, MS, PhD, & Maria J. Suarez, MD, Dr Med, DDS, Dr Odont

Department of Conservative Dentistry and Buccofacial Prostheses, Faculty of Odontology, University Complutense of Madrid, Madrid, Spain

Keywords
CAD-CAM technologies; fixed prostheses; base metal alloys; cantilever; fracture.

Correspondence
Jesus Pelaez, Department of Conservative Dentistry and Buccofacial Prostheses. Faculty of Odontology, University Complutense of Madrid (UCM), Pza Ramón y Cajal s/n. 28040 Madrid, Spain.
E-mail: jpelaezr@ucm.es

Funding: This study was supported by a research grant between the University Complutense of Madrid (UCM) and Prótesis S.A. (No 381/2015) given to the last author.

Conflict of interest: The authors have no conflicts of interest to declare.

Accepted November 25, 2021

doi: 10.1111/jopr.13458

Abstract

Purpose: To compare the load to fracture of cobalt-chromium (Co-Cr) 3-unit posterior fixed partial denture (FPD) frameworks manufactured by conventional and digital techniques and to evaluate the influence of the framework design on the fracture load.
Material and methods: Eighty 3-unit Co-Cr posterior FPD frameworks were fabricated with two designs: intermediate pontic (n = 40) and cantilever (n = 40). Each design was randomly divided into four groups (n = 10): casting, direct metal laser sintering, soft metal milling, and hard metal milling. After thermal cycling, all specimens were subjected to a 3-point bending test until fracture. Data were statistically analyzed using one-way ANOVA, Welch and Brown-Forsythe test, Ryan-Einot-Gabriel-Welsch F and Tamhane T2 post hoc test, Student's t test, and Weibull statistics ($\alpha = 0.05$).
Results: Significant differences ($p < 0.001$; F = 39.59) were found among intermediate pontic frameworks (except between laser sintering and hard metal milling), and cantilevered frameworks (F = 36.75) (except between laser sintering and hard metal milling, and casting and soft metal milling). The cantilever groups showed load to fracture values significantly lower than those of the intermediate pontic ($p < 0.001$; F = 28.29). The Weibull statistics corroborated the results.
Conclusions: Hard metal milling and laser sintered frameworks exhibited the highest load to fracture values. However, all tested frameworks demonstrated clinically acceptable load to fracture values. The framework design directly affected the fracture load, with drastically lower values in cantilevered frameworks.

Fixed partial dentures (FPDs) can be made with different materials such as ceramics, polymers and metal alloys. Metal-ceramic prostheses are still widely used, especially for the rehabilitation of the posterior regions of the mouth, due to the good long-term survival and satisfactory mechanical properties.[1,2] The fracture resistance of a material is one of the main factors that directly influences the restoration success. Furthermore, the mechanical properties can be considered as the first criterion to select a restorative material that has to be able to withstand masticatory forces.[3] This is relevant in the posterior region and in presence of parafunctions where the forces can be as high as 1000 N.[4–6] Moreover, restorations are subjected to temperature variations and moisture in the oral environment that can affect their fracture load.[7] Therefore, several authors recommended that in vitro studies should include artificial aging to simulate the clinical conditions.[7–11]

Metal FPD frameworks have a high flexural strength and fracture toughness, due to the greater stress absorption capacity. The failure of metal-ceramic restorations usually happens from total or partial chipping of the veneering ceramic, leaving the metal framework intact.[2,5,7,12,13] However, metal frameworks properties can be influenced for various aspects such as the alloy type, design, and manufacturing technique.[14–16] Nowadays, cobalt-chromium (Co-Cr) alloys are the base metal alloys of choice due to their excellent mechanical properties such as fracture resistance and hardness, biocompatibility, corrosion resistance, and low cost.[14,17–21]

The introduction of computer-aided design and computer-aided manufacturing (CAD-CAM) technologies has helped overcome many of the disadvantages of traditional casting techniques.[22] The subtractive or milling CAD-CAM techniques allow Co-Cr alloys to be processed from prefabricated discs. Hard metal milling is the longest-running technique and enables accurate restorations, more efficient processing, and the availability of more materials.[23–26] Metal blanks can also be milled in a pre-sintered state.[25,27] This technique provides benefits compared to hard metal milling, allowing dry milling with less contamination, milling time, and wear of the milling unit.[25]

Alternatively, the additive CAD-CAM techniques for metals such as selective laser melting (SLM) or direct metal laser sintering (DMLS) can create structures from metal powder melting layer-by-layer by a laser power beam.[23,28–31] These processes achieve complicated geometries,[29] with less material waste, and with high productivity.[23]

The FPD design will directly affect biological and mechanical behavior.[32] The number and distribution of abutments, pontics, or the connector area can affect resistance to fracture.[33–36] Previous studies have reported that the connector area withstands the highest forces.[4,37,38] This is important in cantilever FPDs in which understanding biomechanics is decisive to correct design and use.[33,39–42] The minimum recommended connector dimensions used in conventional cast frameworks is 6.25 mm^2.[1] However, current studies on CAD-CAM metal frameworks used connectors of 9 mm^2,[4,6,7,27] and even slightly higher.[20] The information on mechanical properties, such as fracture features in metal base alloys FPDs manufactured with the current production methods is sparse, and even less in cantilever designs.

The aim of this in vitro study was to compare the load to fracture of posterior Co-Cr FPD frameworks manufactured by casting and 3 digital techniques, as well as the influence of the framework design (intermediate pontic or cantilever) on the load to fracture. The null hypotheses were that no differences would be found in the load to fracture among the different techniques, and between the framework designs.

Materials and methods

A total of 80 standardized machined stainless-steel specimens, with a platform and 2 abutments were fabricated (Mechanical Workshop of Physical Science, University Complutense of Madrid, Spain). Two platform designs were manufactured to receive posterior 3-unit frameworks: (1) intermediate pontic (30 mm in length, 17 mm in width, 4.5 mm in thickness, and 7 mm between abutments), and (2) cantilever (22 mm in length, 17 mm in width, 4.5 mm in thickness, and 0.2 mm between abutments). The abutments ($n = 160$) were designed simulating a prepared first mandibular premolar (5 mm in height, occlusal diameter of 5 mm, a 1-mm-wide chamfer, and a 6-degree angle of convergence), and screwed to the platforms.[4,6,7,43–45] The specimens were randomly assigned to 4 groups in each design, categorized according to the manufacturing technique: group 1, casting (CM); group 2, laser sintering (LS); group 3, soft metal milling (SM) and group 4, hard metal milling (HM). Ten specimens were fabricated for each study group according to the results of power analysis (the effect size was found to be 1 at 95% power when the total sample size was 32), conducted by using a software program (G*Program 3.1.9.4). The cantilever frameworks were identified with the letter c (Table 1).

To fabricate the CM and CMc frameworks, the specimens were scanned and digitized (Lava Scan ST; 3M ESPE, Seefeld, Germany), and the data were entered into the CAD software (DWOS Lava Edition; Dental Wings, Montreal, Canada). The wax patterns were printed (ProJet 1200 3D; 3D Systems, Rock Hill, SC), and the process continued with the lost-wax casting technique. For this purpose, a commercial phosphate graphite-free investment plaster (Vestofix; DFS Diamond GmbH, Riedenburg, Germany) was used. The casting was performed with an induction and centrifugal vacuum-casting machine (MIE-200C/R; Ordenta, Madrid, Spain) under vacuum pressure of 580 mmHg, at a melting temperature of 1480°C. The LS and LSc specimens were scanned and digitized as in the casted frameworks. The CAD design file was transferred to a DMLS device (PM 100 Dental; Phenix Systems, Clermont-Ferrand, France) and frameworks were made built by fusion of 20 mm layers of alloy powders from the occlusal surface to the margins, under an argon atmosphere, by Yb-fiber laser beam at 1650°C. To fabricate the SM and SMc frameworks, the specimens were scanned and digitized (Ceramill Map400; Amann Girrbach, Koblach, Austria). The frameworks were designed (Ceramill Mind; Amann Girrbach), and milled from presintered Co-Cr discs, with a magnification of 11% to offset post-sintering shrinkage, in the milling unit (Ceramill Motion 2; Amann Girrbach). The frameworks were placed in a sintering tray (Ceramill Argovent; Amann Girrbach) and sintered in a furnace (Ceramill Algotherm 2; Amann Girrbach) at 1.300°C under an argon atmosphere. The manufacture of the HM and HMc frameworks consisted of scanning the specimens (3Shape D750; 3Shape Dental System, Copenhagen, Denmark) and designing the frameworks with the design software (Molder Builder; 3Shape Dental System). The Co-Cr discs were inserted in the warehouse (PH 2/120 SAUER; DMG Mori, Stipshausen, Germany) of the milling unit (Ultrasonic 10 linear SAUER; DMG Mori), and machining was carried out.[4]

All frameworks were fabricated by experienced technicians, with the same dimensions (0.5 mm wall-thickness, cement space of 50 μm, a premolar shape pontic, and a connector area of 9 mm^2), which were verified with a digital micrometer (Mitutoyo Co, Tokyo, Japan). The frameworks were cleaned with water steam and sandblasted with aluminum-oxide particles (50 μm) under 50 N/cm^2 pressure (EXTRAmatic 9040; Kavo Dental GmbH, Biberach, Germany).

The frameworks were cemented with glass-ionomer cement (Ketac-Cem EasyMix; 3M ESPE), following the manufacturer´s specifications (room temperature: 18-24°C; relative humidity: 50 ± 10%). A constant seating load of 50 N was applied with a torque wrench (ZIACOM, Madrid, Spain) for 10 minutes.[4]

Each specimen was placed into a cylindrical polyethylene container with 30 ml of artificial saliva Fusayama-Meyer (LCTech, Obertaufkirchen, Germany),[4,46] and subjected to thermal cycling at 5°C and at 55°C for 6000 cycles in a climatic chamber (CCK0/81; Dycometal, Barcelona, Spain) controlled by iTools software (Eurotherm; Schneider Electric, Madrid, Spain).[4] Each framework was then further subjected to a

Table 1 Manufacturing technologies, alloys and their chemical composition and manufacturers of frameworks with intermediate and cantilever pontic

	Groups				
Manufacturing technologies	(1) Intermediate pontic	(2) Cantilever pontic	Dental alloy composition (weight %)	Alloy	Manufacturer
Casting	CM	CMc	Co 59.5, Cr 31.5, Mo 5, Si 2, Fe, C, Mn ≤1	Super 8	Dental Alloys Products
Laser sintering	LS	LSc	Co 65, Cr 28-30, Mo 5-6, Fe ≤0.5, C ≤0.02, Si, Mn, Ni ≤1	ST2724G	Sint-Tech
Soft metal milling	SM	SMc	Co 66, Cr 28, Mo 5, C ≤0.1, Si, Fe, Mn ≤1	Ceramill Sintron R 71 L	Amann Girrbach
Hard metal milling	HM	HMc	Co 59, Cr 25, Mo 3.5, W 9.5, Si 1, Fe, C, Mn, Ni ≤1.5	Starbond CoS DISC basic	Scheftner

Table 2 Mean, standard deviation (SD), minimum and maximum (N) of load to fracture values in frameworks with intermediate pontic

Intermediate pontic	Mean	SD	Minimum	Maximum
CM	9836.26	433.88	9167.90	10598.70
LS	10784.99	476.11	10101.84	11487.56
SM	9066.08	451.98	8336.86	9728.46
HM	11156.61	530.69	10430.14	11860.94

CM = casting; LS = laser sintering; SM = soft metal milling; HM = hard metal milling.

3-point bending test until fracture using a universal testing machine (UTM) (ME 405/10; SERVOSIS, Madrid, Spain) with a 10 Tm load-cell, at a 1 mm/min crosshead speed.[4,6,7,43,45,47,48] Tin foil (0.3-mm-thick) was interposed between the rounded tip (1.5 mm in diameter) of the cone-shape loading pusher adapted to the UTM and the frameworks for a more even load distribution and to reduce the stress concentration.[37,49–55] Axial compressive vertical load was applied at the central fossa of the pontic. Data on the load to fracture were automatically recorded in Newtons (N) by a specific software (PCD2K; SERVOSIS).[4,6,7,43–45] Fracture was defined as the moment in which a drastic decrease was recorded on the monitored curves, along with evidence of the visible cracks and acoustic events.[4,6,7,45]

Statistical analysis was performed with a statistical software (IBM SSPS Statistics, v22.0; IBM Corp, Chicago, IL) ($\alpha = 0.05$). Means and standard deviations (SD) were calculated for each group. The Shapiro-Wilk test was used to confirm the normality of the variables. One-way ANOVA was performed for comparisons among the groups. The Ryan-Einot-Gabriel-Welsch F and Welch and Brown-Forsythe test and TamhaneT2 test were used for post hoc comparisons among the intermediate pontic and the cantilever frameworks respectively. Student's t test was used for comparisons between frameworks designs. To facilitate accurate interpretation of data, the Weibull characteristic fracture load or scale ($\sigma 0$) and the Weibull modulus (m), were also analyzed.[4,7,45]

Results

The mean and standard deviation of the load to fracture values for intermediate pontic and cantilever groups are listed in Tables 2 and 3, respectively. LS and HM groups exhibited the highest load to fracture values. ANOVA revealed significant differences ($p < 0.001$; F = 39.59) among intermediate pontic frameworks groups. The post hoc test revealed differences among all frameworks except between LS and HM groups. Likewise, in the cantilever frameworks, significant differences ($p < 0.001$; F = 36.75) were found among the groups, except between CMc and SMc, and between LSc and HMc. When the design was compared, the cantilevered frameworks showed significantly lower load to fracture values than those with intermediate pontic ($p < 0.001$; F = 28.29).

The Weibull statistics revealed no significant differences in the m parameter, both in intermediate pontic frameworks and in cantilevered ones (Tables 4 and 5). All frameworks obtained high modulus values, which mean, and the sample distribution was lower. However, significant differences were found in the $\sigma 0$ parameter for the intermediate pontic frameworks (Table 4), except between the LS and HM groups, in which an overlap was noted. Likewise, significant differences were observed in the $\sigma 0$ parameter for the cantilever group (Table 5) except between the overlapped frameworks: CMc and SMc. Thus, the HM and HMc groups exhibited the highest characteristic fracture load values, achieving the least probability

Table 3 Mean, standard deviation (SD), minimum and maximum (N) of load to fracture values in cantilevered frameworks

Cantilever	Mean	SD	Minimum	Maximum
CMc	1858.77	108.88	1675.80	1983.52
LSc	2161.14	61.10	2061.92	2234.40
SMc	1831.69	97.43	1690.50	1997.04
HMc	2257.14	157.37	2042.32	2495.08

CMc = casting; LSc = laser sintering; SMc = soft metal milling; HMc = hard metal milling.

Table 4 Weibull statistics of load to fracture (N) for frameworks with intermediate pontic

	m = Weibull modulus						$\sigma 0$ = Weibull scale					
			CI (95%)		CI (99%)				CI (95%)		CI (99%)	
	Est	St Error	Lo	Up	Lo	Up	Est	St Error	Lo	Up	Lo	Up
CM	25.02	5.89	15.76	39.70	13.43	46.59	10036.28	134.57	9775.97	10303.53	9687.44	10397.68
LS	26.92	6.64	16.60	43.65	14.04	51.61	11000.67	136.71	10735.95	11271.91	10645.81	11367.35
SM	24.98	5.34	16.42	37.97	14.21	43.90	9240.56	108.51	9030.32	9455.70	8958.66	9531.32
HM	24.67	4.31	17.51	34.76	15.55	39.14	11398.64	109.40	11186.23	11615.08	11113.63	11690.95

CM = casting; LS = laser sintering; SM = soft metal milling; HM = hard metal milling. CI = confidence interval; Est = estimate; St Error = standard error; Lo = lower; Up = upper.

Table 5 Weibull statistics of load to fracture (N) for cantilever frameworks

	m = Weibull modulus						$\sigma 0$ = Weibull scale					
			CI (95%)		CI (99%)				CI (95%)		CI (99%)	
	Est	St Error	Lo	Up	Lo	Up	Est	St Error	Lo	Up	Lo	Up
CMc	23.34	4.36	16.19	33.66	14.26	38.21	1905.27	19.18	1868.05	1943.23	1855.33	1956.54
LSc	46.82	8.59	32.67	67.10	28.84	76.00	2188.14	11.00	2166.68	2209.81	2159.30	2217.36
SMc	20.27	3.41	14.58	28.17	13.01	31.58	1877.23	22.00	1834.60	1920.85	1820.07	1936.19
HMc	17.30	3.02	12.28	24.37	10.91	27.44	2326.51	31.79	2265.04	2389.66	2244.14	2411.91

CMc = casting; LSc = laser sintering; SMc = soft metal milling; HMc = hard metal milling. CI = confidence interval; Est = estimate; St Error = standard error; Lo = lower; Up = upper.

of failure (Fig 1). Regarding the frameworks design, significant differences were found in the $\sigma 0$ parameter, corroborating that a considerably lower load is necessary to fracture cantilevered frameworks. However, in m parameter no differences were found.

All frameworks first showed a plastic deformation and then, a crack appeared at the connector area with two different patterns according to frameworks design. The breakage of frameworks with intermediate pontic started at the cervical area of the connector and spread obliquely towards the occlusal surface (Fig 2). Conversely, in the cantilevered frameworks the breakage started at the occlusal surface of the connector (Fig 3).

Discussion

The data obtained support the rejection of the null hypotheses, as the load to fracture values among the groups and between the frameworks design exhibited significant differences.

Studies that compared the load to fracture of different metal frameworks with intermediate pontic are sparse. In the present study, the hard metal milling and laser sintering groups exhibited the highest load to fracture. The results were consistent with those of previous studies for laser sintered frameworks,[14,56–59] and hard metal milled frameworks.[60] Conversely, other studies have reported the lowest flexural strength for hard metal milled frameworks.[56,58] Regarding the soft metal milling group, the results were consistent with a previous study.[27] However, superior mechanical properties for soft metal milling group and comparable to laser sintering technique have been reported in other study.[56] In the study, the chemical composition of the alloys was not identical, and their microstructure was not analyzed. This factor could have influenced the differences among the analyzed groups as previously reported,[14,15,56,60] as well as the different manufacturing processes.[20,60] The fracture initiation of the tested frameworks occurred at values higher than 9000 N, suggesting that they can withstand clinical chewing forces even in the presence of parafunctions. The results were consistent with those of previous studies.[4,6,7]

The cantilever frameworks showed load to fracture values above those clinically relevant (1000 N). The hard metal

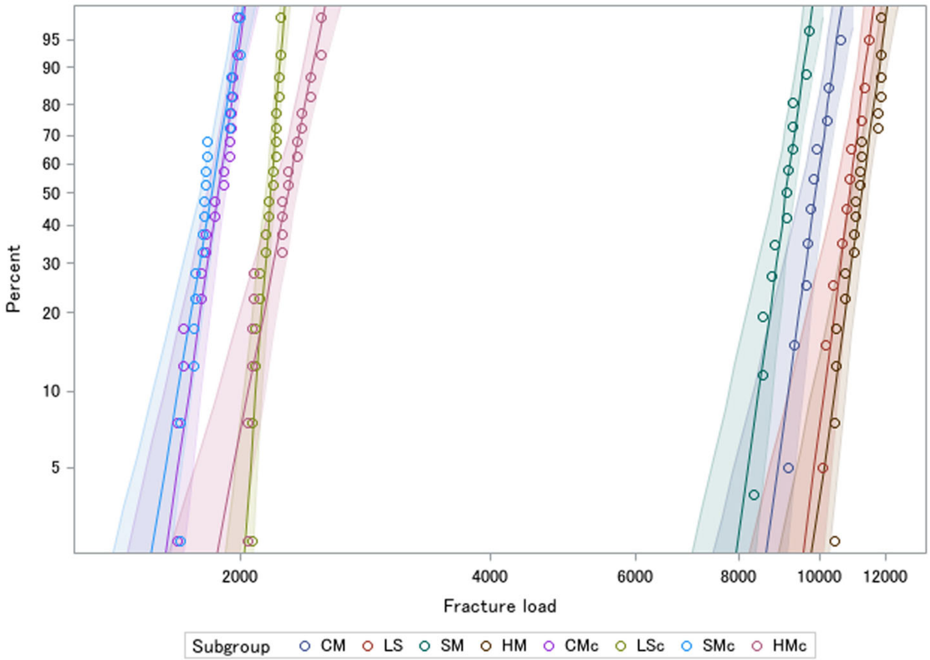

Figure 1 Weibull probability plot of the load to fracture for frameworks with intermediate and cantilever pontics. CM and CMc = casting; LS and LSc = laser sintering; SM and SMc = soft metal milling; HM and HMc = hard metal milling.

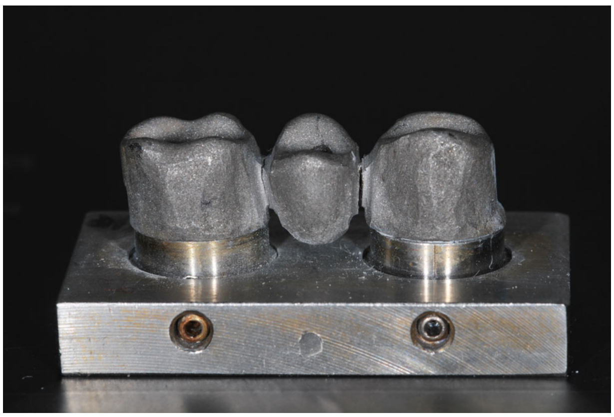

Figure 2 Fractured specimen of the soft metal milled group.

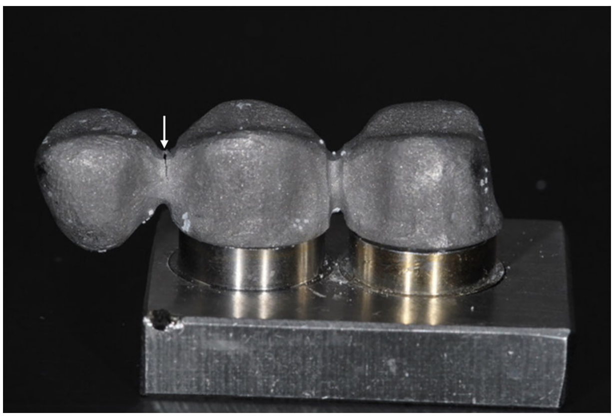

Figure 3 Fractured specimen of the soft metal milled group with cantilever.

milling and the laser sintering groups also showed the highest load to fracture values. The authors are unaware of previous studies evaluating the load to fracture of cantilevered frameworks manufactured with digital technologies, and it was thus not possible to compare the results of the study. The differences among the groups could also be due to differences in the microstructure of the alloys employed and the processing techniques.

When both designs manufactured with the same technology were compared, the cantilever frameworks showed a drastic decrease in their load to fracture values. The results indicate that the framework design influenced the fracture behavior. Furthermore, the highest values for principal stress in the cantilever frameworks were found in the connector between the pontic and the retainer.[40] All tested frameworks showed plastic deformation prior to fracture that occurred at the connector area. This pattern is consistent with previous studies in FPDs with intermediate pontic, which reported that the connector area withstands the highest tensile and shear forces.[6,7,37,38,43,45] In cantilevered frameworks, the breakage also occurred at the connector area but started in the upper surface of the connector. The results were consistent with those of previous studies.[39,40,61] The excessive load on the cantilever pontic is a risk factor due to the appearance of lateral forces, inclination and rotation of the abutment teeth.[33,62] This situation can cause the fracture or deformation with a gingival displacement of the pontic,[32,36,63] and it will depend on the dimensions of the extension.[33,39,64] Thus, although cantilever prostheses can be used in posterior areas[65] factors such as the dimensions of the prosthesis, the use of non-noble alloys, and the number of abutment teeth[33,39,64] should always be taken into account, and future research is required.

Artificial aging to simulate the clinical situation is controversial. Some authors have performed only static loading in their studies.[5,43,45] Other authors[7,8,11] applied artificial aging procedures with controversial results. However, aging process was recommended to avoid obtaining higher and unrealistic load to fracture values.[8,13,44,50,51] In this study, all the specimens were subjected to thermal cycling simulating a 5-year period in the mouth.[10,50,51] Cyclic loading seems to have a greater influence on the chipping of the veneering ceramic,[10,12,66] and was not performed in the study because unveneered frameworks were evaluated.

Finally, the results of the study were consistent with previous studies that reported that the mechanical properties and load to fracture of the frameworks depend on several factors such as the composition of the alloy and its fabrication technique,[14,20,56] which may lead to a certain microstructure and to a mechanical alteration when loaded.[31,56,60] All these factors make comparisons extremely difficult, and there is a need for a standardized methodology to compare the results among the different studies.

This study was performed in vitro, which allows for the evaluation of the mechanical properties of materials under standardized conditions but may not replicate the clinical conditions. In the study, metal abutments were used as in previous studies,[4,6,7,27,34,43–45,54] that avoid possible premature destruction when testing metal alloys.[4,27,48] Future studies are recommended with other alloys. In addition, clinical trials must be performed to evaluate the mechanical behavior of metal frameworks manufactured with digital technologies, especially with cantilever designs.

Conclusions

All the FPD frameworks evaluated exhibited load to fracture values above those clinically relevant and can therefore be indicated for posterior areas. Hard metal milled, and laser sintered frameworks obtained the highest load to fracture values, in both framework designs. The cantilevered frameworks showed load to fracture values drastically lower than those of the intermediate pontic.

Acknowledgments

The authors thank the dental laboratories of Prótesis SA., 3Dental, and Dental Creative for manufacturing the frameworks, as well as Dr. Carmen Bravo, Centre of Data Processing, Computing Service for Research Support, University Complutense of Madrid, for her assistance with the statistical analysis.

References

1. Ozcan M: Fracture reasons in ceramic-fused-to-metal restorations. J Oral Rehabil 2003;30:265-269
2. Eliasson A, Arnelund CF, Johansson A: A clinical evaluation of cobalt-chromium metal-ceramic fixed partial dentures and crowns: a three- to seven-year retrospective study. J Prosthet Dent 2007;98:6-16
3. Campos RE, Soares PV, Versluis A, et al: Crown fracture: failure load, stress distribution, and fractographic analysis. J Prosthet Dent 2015;114:447-455
4. Rodríguez V, Tobar C, López-Suárez C, et al: Fracture load of metal, zirconia and polyetheretherketone posterior CAD-CAM milled fixed partial denture frameworks. Materials (Basel) 2021;14:959. https://doi.org/10.3390/ma14040959
5. Agustin-Panadero R, Fons-Font A, Roman-Rodriguez JL, et al: Zirconia versus metal: a preliminary comparative analysis of ceramic veneer behavior. Int J Prosthodont 2012;25:294-300
6. Lopez-Suarez C, Castillo-Oyague R, Rodriguez-Alonso V, et al: Fracture load of metal-ceramic, monolithic, and bi-layered zirconia-based posterior fixed dental prostheses after thermo-mechanical cycling. J Dent 2018;73:97-104
7. Lopez-Suarez C, Tobar C, Sola-Ruiz MF, et al: Effect of thermomechanical and static loading on the load to fracture of metal-ceramic, monolithic, and veneered zirconia posterior fixed partial dentures. J Prosthodont 2019;28:171-178
8. Yang R, Arola D, Han Z, et al: A comparison of the fracture resistance of three machinable ceramics after thermal and mechanical fatigue. J Prosthet Dent 2014;112:878-885
9. Fischer J, Zbären C, Stawarczyk B, et al: The effect of thermal cycling on metal-ceramic bond strength. J Dent 2009;37:549-553
10. Antanasova M, Kocjan A, Kovac J, et al: Influence of thermo-mechanical cycling on porcelain bonding to cobalt-chromium and titanium dental alloys fabricated by casting, milling, and selective laser melting. J Prosthodont Res 2018;62:184-194
11. Baladhandayutham B, Lawson NC, Burgess JO: Fracture load of ceramic restorations after fatigue loading. J Prosthet Dent 2015;114:266-271
12. Nicolaisen MH, Bahrami G, Finlay S, et al: Comparison of fatigue resistance and failure modes between metal-ceramic and all-ceramic crowns by cyclic loading in water. J Dent 2014;42:1613-1620
13. Pjetursson BE, Sailer I, Makarov NA, et al: All-ceramic or metal-ceramic tooth-supported fixed dental prostheses (FDPs)? A systematic review of the survival and complication rates. Part II: multiple-unit FDPs. Dent Mater 2015;31:624-639
14. Zhou Y, Li N, Yan J, et al: Comparative analysis of the microstructures and mechanical properties of Co-Cr dental alloys fabricated by different methods. J Prosthet Dent 2018;120:617-623
15. Al Jabbari YS, Koutsoukis T, Barmpagadaki X, et al: Metallurgical and interfacial characterization of PFM Co-Cr dental alloys fabricated via casting, milling or selective laser melting. Dent Mater 2014;30:e79-e88
16. Al Jabbari YS, Barmpagadaki X, Psarris I, et al: Microstructural, mechanical, ionic release and tarnish resistance characterization of porcelain fused to metal Co-Cr alloys manufactured via casting and three different CAD/CAM techniques. J Prosthodont Res 2019;63:150-156
17. Li KC, Prior DJ, Waddell JN, et al: Comparison of the microstructure and phase stability of as-cast, CAD/CAM and

powder metallurgy manufactured Co-Cr dental alloys. Dent Mater 2015;31:e306-e315
18. Lucchetti MC, Fratto G, Valeriani F, et al: Cobalt-chromium alloys in dentistry: an evaluation of metal ion release. J Prosthet Dent 2015;114:602-608
19. Xin XZ, Chen J, Xiang N, et al: Surface characteristics and corrosion properties of selective laser melted Co-Cr dental alloy after porcelain firing. Dent Mater 2014;30:263-270
20. Oilo M, Nesse H, Lundberg OJ, et al: Mechanical properties of cobalt-chromium 3-unit fixed dental prostheses fabricated by casting, milling, and additive manufacturing. J Prosthet Dent 2018;120:156 e151-156 e157
21. Al Jabbari YS: Physico-mechanical properties and prosthodontic applications of Co-Cr dental alloys: a review of the literature. J Adv Prosthodont 2014;6:138-145
22. Kim MJ, Choi YJ, Kim SK, et al: Marginal accuracy and internal fit of 3-D printing laser-sintered Co-Cr alloy copings. Materials (Basel) 2017;10:93
23. van Noort R: The future of dental devices is digital. Dent Mater 2012;28:3-12
24. Alghazzawi TF: Advancements in CAD/CAM technology: options for practical implementation. J Prosthodont Res 2016;60:72-84
25. Kim KB, Kim JH, Kim WC: Three-dimensional evaluation of gaps associated with fixed dental prostheses fabricated with new technologies. J Prosthet Dent 2014;112:1432-1436
26. Miyazaki T, Hotta Y: CAD/CAM systems available for the fabrication of crown and bridge restorations. Aust Dent J 2011;56:97-106
27. Krug KP, Knauber AW, Nothdurft FP: Fracture behavior of metal-ceramic fixed dental prostheses with frameworks from cast or a newly developed sintered cobalt-chromium alloy. Clin Oral Investig 2015;19:401-411
28. Wu L, Zhu H, Gai X, et al: Evaluation of the mechanical properties and porcelain bond strength of cobalt-chromium dental alloy fabricated by selective laser melting. J Prosthet Dent 2014;111:51-55
29. Sundar MK, Chikmagalur SB, Pasha F: Marginal fit and microleakage of cast and metal laser sintered copings—an in vitro study. J Prosthodont Res 2014;58:252-258
30. Kocaağaoğlu H, Kılınç H, Albayrak H, et al: In vitro evaluation of marginal, axial, and occlusal discrepancies in metal ceramic restorations produced with new technologies. J Prosthet Dent 2016;116:368-374
31. Suleiman SH, Vult von Steyern P: Fracture strength of porcelain fused to metal crowns made of cast, milled or laser-sintered cobalt-chromium. Acta Odontol Scand 2013;71:1280-1289
32. Hämmerle CH, Ungerer MC, Fantoni PC, et al: Long-term analysis of biologic and technical aspects of fixed partial dentures with cantilevers. Int J Prosthodont 2000;13:409-415
33. Yang HS, Chung HJ, Park YJ: Stress analysis of a cantilevered fixed partial denture with normal and reduced bone support. J Prosthet Dent 1996;76:424-430
34. Bonfante EA, da Silva NR, Coelho PG, et al: Effect of framework design on crown failure. Eur J Oral Sci 2009;117:194-199
35. Rehmann P, Podhorsky A, Wöstmann B: Treatment outcomes of cantilever fixed partial dentures on vital abutment teeth: a retrospective analysis. Int J Prosthodont 2015;28:577-582
36. Brägger U, Hirt-Steiner S, Schnell N, et al: Complication and failure rates of fixed dental prostheses in patients treated for periodontal disease. Clin Oral Implants Res 2011;22:70-77
37. Oh WS, Anusavice KJ: Effect of connector design on the fracture resistance of all-ceramic fixed partial dentures. J Prosthet Dent 2002;87:536-542
38. Augereau D, Pierrisnard L, Barquins M: Relevance of the finite element method to optimize fixed partial denture design. Part I. Influence of the size of the connector on the magnitude of strain. Clin Oral Investig 1998;2:36-39
39. Eraslan O, Sevimay M, Usumez A, et al: Effects of cantilever design and material on stress distribution in fixed partial dentures—a finite element analysis. J Oral Rehabil 2005;32:273-278. https://doi.org/10.1111/j.1365-2842.2004.01429.x
40. Romeed SA, Fok SL, Wilson NH: Finite element analysis of fixed partial denture replacement. J Oral Rehabil 2004;31:1208-1217
41. Zenthöfer A, Ohlmann B, Rammelsberg P, et al: Performance of zirconia ceramic cantilever fixed dental prostheses: 3-year results from a prospective, randomized, controlled pilot study. J Prosthet Dent 2015;114:34-39
42. Pjetursson BE, Brägger U, Lang NP, et al: Comparison of survival and complication rates of tooth-supported fixed dental prostheses (FDPs) and implant-supported FDPs and single crowns (SCs). Clin Oral Implants Res 2007;18(Suppl 3):97-113
43. López-Suárez C, Gonzalo E, Peláez J, et al: Fracture resistance and failure mode of posterior fixed dental prostheses fabricated with two zirconia CAD/CAM systems. J Clin Exp Dent 2015;7:e250-e253
44. Lopez-Suarez C, Rodriguez V, Pelaez J, et al: Comparative fracture behavior of monolithic and veneered zirconia posterior fixed dental prostheses. Dent Mater J 2017;36:816-821
45. Rodríguez V, Castillo-Oyagüe R, López-Suárez C, et al: Fracture load before and after veneering zirconia posterior fixed dental prostheses. J Prosthodont 2016;25:550-556
46. Schiff N, Grosgogeat B, Lissac M, et al: Influence of fluoride content and pH on the corrosion resistance of titanium and its alloys. Biomaterials 2002;23:1995-2002
47. Ulusoy M, Toksavul S: Fracture resistance of five different metal framework designs for metal-ceramic restorations. Int J Prosthodont 2002;15:571-574
48. Eroğlu Z, Gurbulak AG: Fatigue behavior of zirconia-ceramic, galvano-ceramic, and porcelain-fused-to-metal fixed partial dentures. J Prosthodont 2013;22:516-522
49. Rosentritt M, Behr M, Thaller C, et al: Fracture performance of computer-aided manufactured zirconia and alloy crowns. Quintessence Int 2009;40:655-662
50. Rosentritt M, Siavikis G, Behr M, et al: Approach for valuating the significance of laboratory simulation. J Dent 2008;36:1048-1053
51. Rosentritt M, Behr M, van der Zel JM, et al: Approach for valuating the influence of laboratory simulation. Dent Mater 2009;25:348-352
52. Kellerhoff RK, Fischer J: In vitro fracture strength and thermal shock resistance of metal-ceramic crowns with cast and machined AuTi frameworks. J Prosthet Dent 2007;97:209-215
53. Gabbert O, Ohlmann B, Schmitter M, et al: Fracture behaviour of zirconia ceramic cantilever fixed dental prostheses in vitro. Acta Odontol Scand 2008;66:200-206
54. Wimmer T, Erdelt KJ, Eichberger M, et al: Influence of abutment model materials on the fracture loads of three-unit fixed dental prostheses. Dent Mater J 2014;33:717-724
55. Ohlmann B, Dittmar A, Rues S, et al: Comparison of fracture-load values of cantilevered FDPs. Acta Odontol Scand 2013;71:584-589

56. Kim HR, Jang SH, Kim YK, et al: Microstructures and mechanical properties of Co-Cr dental alloys fabricated by three CAD/CAM-based processing techniques. Materials (Basel) 2016;9:596
57. Han XT, Sawada T, Schille C, et al: Comparative analysis of mechanical properties and metal-ceramic bond strength of Co-Cr dental alloy fabricated by different manufacturing. Processes. Materials 2018;11:1801
58. Hong JK, Kim SK, Heo SJ, et al: Mechanical Ppoperties and metal-ceramic bond strength of Co-Cr alloy manufactured by selective laser melting. Materials (Basel) 2020;13:5745
59. Kaleli N, Uçar Y, Ekren O, et al: Effect of layer thickness on the flexural strength of multiple-unit laser-sintered metal frameworks. J Prosthet Dent 2021; Feb 24:S0022-3913(20)30551-5. https://doi.org/10.1016/j.prosdent.2020.10.006.
60. Padrós R, Punset M, Molmeneu M, et al: Mechanical properties of CoCr dental-prosthesis restorations made by three manufacturing processes. Influence of the microstructure and topography. Metals 2020;10:788. https://doi.org/10.3390/met10060788
61. Zhang ZP, Zhou SW, Li E, et al: Design for minimizing fracture risk of all-ceramic cantilever dental bridge. Biomed Mater Eng 2015;26:S19-S25
62. Laurell L, Lundgren D, Falk H, et al: Long-term prognosis of extensive polyunit cantilevered fixed partial dentures. J Prosthet Dent 1991;66:545-552
63. De Backer H, Van Maele G, De Moor N, et al: A 20-year retrospective survival study of fixed partial dentures. Int J Prosthodont 2006;19:143-153
64. Palmqvist S, Söderfeldt B: Multivariate analyses of factors influencing the longevity of fixed partial dentures, retainers, and abutments. J Prosthet Dent 1994;71:245-250
65. Pjetursson BE, Lang NP: Prosthetic treatment planning on the basis of scientific evidence. J Oral Rehabil 2008;35 (Suppl 1):72-79
66. Papanagiotou HP, Morgano SM, Giordano RA, et al: In vitro evaluation of low-temperature aging effects and finishing procedures on the flexural strength and structural stability of Y-TZP dental ceramics. J Prosthet Dent 2006;96: 154-164