Load to failure of high-density polymers for implant-supported fixed, cantilevered prostheses with titanium bases

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Purpose: To analyze the load to failure of different CAD/CAM high-density polymers (HDPs) and zirconia when titanium (Ti) bases were included in a cantilevered situation.

Materials and Methods: Five specimens were fabricated from five different CAD/CAM polymethyl methacrylate (PMMA) HDPs (Copratemp [CT]; Tempo-CAD [TC]; TD Dental [TD]; M-PM Disc-Pink [MPM]; M-PM Disc-White [MPMW]), and five specimens were prepared from a 3Y-TZP zirconia (FireZr [FZR]) (control). Ti bases (D Master Dental Implants) were cemented onto the specimens (8 mm [thickness] × 7 mm [width] × 30 mm [length]). Each specimen was fixated using a clamp for a cantilever loading distance of 10 mm. The load was applied on the cantilever until failure, and the maximum load to failure values (N) were analyzed by using analysis of variance (GLIMMIX procedure) with a lognormal error distribution in addition to the restricted maximum likelihood estimation method to eliminate the need for equality of variances and Tukey Honest Significant Difference ($\alpha = .05$). **Results:** Differences among load-to-failure values of HDPs were not significant ($P > .05$). However, zirconia had significantly higher load-to-failure values than HDPs ($P < .001$). The behavior of HDPs and zirconia under loading was different in terms of displacement. HDPs showed weaker but more ductile behavior than zirconia, which is stronger, but more brittle. **Conclusion:** Tested brands of HDPs performed similarly under loading. Zirconia with a Ti base showed higher strength compared to all tested HDPs with a Ti base. The loads that fractured the specimens with Ti bases were close to the maximum occlusal bite forces recorded in previous clinical studies. *Int J Prosthodont 2021. doi: 10.11607/ijp.7036*

INTRODUCTION

The goal of an interim implant-supported fixed complete-arch prostheses (ISFCAP) is to fulfill optimal esthetic and functional needs of the patients without biological and technical complications until the definitive prosthesis is placed.[1-4] An interim ISFCAP need to withstand masticatory forces during the osseointegration period.[1,5] Therefore, an interim

ISFCAP should have adequate strength especially when used in cantilevered situations.[1,6-8]

Conventionally, interim ISFCAP have been fabricated from autopolymerized or heat-polymerized poly(methyl methacrylate) (PMMA) acrylic resin.[1,9] Despite their common use, fractures have been reported in clinical studies with varying frequencies.[1,9] Therefore, reinforcing with fiber meshes, metallic bars or metal wires have been reported to improve fracture and impact strength of the conventional interim ISFCAP.[10,11] Conventional ISFCAP are also prone to inhomogeneities, porosities, and cracks due to fabrication techniques.[6,7]

ISFCAP can be fabricated by using computer-aided design and computer-aided manufacturing (CAD-CAM) technology.[12] High-density polymers (HDPs) are recently marketed materials, which are used for the fabrication of interim prostheses with CAD-CAM technology.[13,14] HDPs are highly cross-linked PMMA- or composite-based polymer materials.[14] CAD-CAM PMMA HDPs have gained popularity due to their advantages such as improved mechanical properties, biocompatibility, and decreased risk of porosities through manufacturing under standardized polymerizing conditions at high temperature and pressure.[6,13,15] They also have ease of machinability and reasonable cost.[12,14] No need for reinforcement has been claimed for CAD-CAM HDPs in a previous report.[12] CAD-CAM HDPs have been preferred especially in extended treatment phases and complex situations, and as a diagnostic prosthesis to verify impression accuracy before definitive prosthesis fabrication.[12,16,17] Wiegand et al[18] also reported that CAD-CAM HDPs had higher load-bearing capacity in 3-unit fixed partial prostheses than conventionally processed ones.

Although there are advancements in materials and fabrication techniques, several complications have been reported for interim ISFCAP, especially under high occlusal loads in clinical studies.[1,19] Biomechanical principles play a major role in the success of ISFCAP.[20] Cantilevers may adversely affect the longevity of ISFCAP and a cantilever can be a factor for prosthetic complications[20,21] like screw loosening, screw, denture tooth, and framework fractures.[22,23] Among them, fracture is one of the most common technical complications.[8] Fracture is related to material, prosthetic design errors (framework misfit, inadequate prosthetic space, extreme cantilevers), patient dependent factors (parafunctional activities) and fabrication errors.[22]

When load is applied to the ISFCAP, flexure is seen following the formation of deformation energy.[24] Additionally, if the stress in prosthesis and implant connection exceeds the biomechanical properties of the framework material, fracture may occur.[24] The determination of repair or re-fabrication of a new prosthesis depends on the fracture's location and severity.[10,20] A complete fracture generally starts after excessive load application, which can propagate through the cross-section of the framework and in a clinical scenario when complete separation of the fractured segments takes place, refabrication of the prosthesis is generally required.[8,10,22] Therefore, the evaluation of maximum load-to-failure performance has clinical importance.[10,13] Bevilacqua et al[25] also reported that maximum stress of the framework was seen at the level of the connection of the framework and the most distal implant. This area was a fulcrum and the bar had a tendency to bend.[25] In light of these, maximum stress can be expected at the junction of Ti base and implant, especially in cantilevered situations. When Ti base is used, the thickness of the framework around Ti base is less. Malo et al[26] reported cross-sectional material dimensions as occlusocervical height of minimum 5 mm and increasing the buccolingual width to a minimum of 6 mm in the areas of the titanium sleeves for complete-arch implant supported fixed polyetheretherketone (PEEK)-acrylic resin prostheses. However, there is no study that reported optimal material width and height to increase the strength of CAD-CAM HDP frameworks for ISFCAPs with Ti bases.

Although it has been reported that the fracture load of fixed prostheses may be affected by the elastic modulus of abutment[27] and increased with higher elastic modulus,[28] according to the authors' knowledge, no load-to-failure data are available for the cantilevered

CAD-CAM HDP interim prostheses frameworks with Ti bases. Additionally, although several CAD-CAM PMMA HDP blanks are available by different brands with different monomer and chemical compositions and mechanical properties,[13] there is limited information in the literature about the effect of varying HDPs on the load-to-failure performance.[13,14] The purpose of this in vitro study was to analyze the load-to-failure of different CAD-CAM HDPs and zirconia when used with Ti bases in a cantilevered situation. The null hypothesis was that the cantilevered CAD-CAM frameworks fabricated from different HDPs and zirconia would have similar load-to-failure performance when Ti base is used.

MATERIAL AND METHODS

Specimens (N=30) out of 5 different CAD-CAM PMMA HDPs (CT, CopraTemp; White Peaks Dental Systems GmbH&Co.KG.; TC, Tempo-CAD; GDS Dental GmbH; TD, Solid shade PMMA disc; TD Dental Supply; MPM, M-PM Disc-pink; Merz Dental GmbH; MPMW, M-PM Disc-white; Merz Dental GmbH) (Fig. 1) and one 3Y-TZP zirconia (FZR, FireZr; Glidewell Direct GmbH) (Table 1) were fabricated. The FZR specimens were milled and no surface alterations were made after milling. After milling, the FZR specimens were sintered and cooled down following the zirconia manufacturer's guidelines. The specimens were 8 mm-thick (occlusocervical), 7 mm-wide (buccolingual), and 30 mm-long. Similar previous studies with same design were considered when determining the sample size and specimen fabrication.[13,29] In previously published studies done by the same group, using the identical test set up in this study and 5 specimens per group allowed the detection of significant differences amongst groups. Specimens were fabricated integrating a Ti base standard tessellation language (STL) in the design for a Ti base (Dmaster Dental Implants) to be later cemented.

After the fabrication of specimens, cementation of Ti bases into their frameworks was executed by the same operator (B.B.). Bonding surfaces of Ti bases were air abraded with alumina with particle size of 110 µm for 10 s at a pressure of 2 bar and a distance of 10 mm. Air abrasion with 50 µm alumina particles was preferred for HDP and FZR specimens.[18] For chemical conditioning of bonding surfaces, a primer (MKZ; Bredent GmbH & Co.KG.) for Ti bases and FZR and a light-polymerizing bonding agent (Visiolink; Bredent GmbH & Co.KG.) for HDP specimens were applied according to manufacturer recommendations. For cementation, a resin luting agent (Panavia 21; Kuraray Co.) was applied according to manufacturer recommendations. To test the load-to-failure when a 10 mm-cantilever is loaded, each specimen was clamped at their first 20 mm (Fig. 2). Before the load application, the load frame was contacted to framework 2 mm before the end of the specimen. A universal testing machine (Instron Model 1321; Instron) and a biaxial servohydraulic load frame were used to set static loading at a 1 mm/min crosshead speed. The load in the vertical direction was applied to the cantilever part of the frameworks until a fracture was detected (Fig. 3) and the maximum load-to-failure values were recorded (N). Statistical analysis was performed by using analysis of variance (GLIMMIX procedure; SAS Proprietary Software v9.3, SAS Institute Inc) with a lognormal error distribution in addition to the restricted maximum likelihood estimation method to eliminate the need for equality of variances. Tukey HSD was also used to determine the significant differences (α=.05). To observe the materials' behavior under loads when they were monolithic, all materials were fabricated in identical size without Ti base space and Ti bases and loaded to failure with the same test set-up.

RESULTS

The mean maximum load-to-failure values for HDPs with Ti bases ranged from 483.10 to 563.54 N and the mean maximum load-to- failure for FZR was 3050.36 N (Fig. 4). The 3Y-TZP zirconia, FZR, had significantly higher load-to- failure values than all HDPs (P<.001).

However, no statistically significant differences were found among all HDP specimens with Ti bases (*P*>.05). Load-displacement curves of HDPs and FZR with Ti base are displayed in Figure 5 and a plot for the monolithic specimens' load-displacement curves is displayed in Figure 6.

DISCUSSION

The null hypothesis that different cantilevered CAD-CAM HDPs and zirconia would have similar load-to-failure performance when used with Ti bases was rejected because zirconia (FZR) had significantly higher load-to-failure values than HDP specimens (*P*<.001). Even though significant differences were not found amongst HDPs, the confidence intervals being fairly tight indicates the test design being stable with a high power.

For the specimen design in the current study, it was aimed to focus on the performance of tested materials under standardized test conditions rather than designing the specimens to replicate the shape of teeth. The specimens were fabricated in rectangular prism to facilitate the standardization of the test design, respecting manufacturer's recommendation and clinical simulation for specimen dimensions. The length of cantilever represented 1 molar width.[13,31] In a previous study, fracture behavior of CAD-CAM HDPs used for interim implant-supported fixed prostheses with cantilever was evaluated.[13] The authors reported the mean load-to-failure values of CAD-CAM HDP, injection-molded, and autopolymerized acrylic resin specimens and they ranged from 789 to 1380 N.[13] In the present study, the mean load-to-failure values for HDP specimens ranged between 483.10 and 563.54 N and they ranged below the results reported in Yilmaz et al's[13] study which tested specimens in similar dimensions without Ti bases.

The load-displacement curve for specimens with Ti bases (Figure 4) was generated to observe how the materials behaved under loading. The same tests were also performed using monolithic specimens for all groups without a Ti base space and Ti bases. The load-displacement curve was also generated for the monolithic group (Figure 5). These plots illustrate the maximum loads and displacements achieved by the materials from both Ti-base and monolithic groups. As expected, both average load and displacement from the monolithic group were higher than those from the Ti-base group. According to the plots in Figures 4 and 5 the HDPs showed weaker, yet more ductile behavior than FZR which is stronger but more brittle. FZR specimens sustained much higher loads, however, they failed with significantly less displacement than the HDPs. The addition of the Ti base (for both FZR and HDPs) reduces the maximum sustained load and the displacement at failure due to the addition of a stress concentration at the base.

FZR specimens with a Ti base (3050.36 N) failed under higher values than the maximum bite force values (Gibbs et al,[32] 1243 N, Ferrario et al,[33] 1221 N, Braun et al,[34] 1280 N) reported in the previous studies. The favorable load-to-fracture performance of tested FZR is in line with the results from previous in vivo studies, which evaluated the performance of zirconia frameworks when used in complete-arch situations.[35-40] These studies reported high survival rates for monolithic zirconia frameworks. Tischler[37] et al reported 99.4% cumulative survival rate for 49 complete-arch zirconia prostheses over a 4-year period. In a 2-7-year retrospective study, Vizcaya[36] reported no chipping or catastrophic failures even when zirconia complete-arch prostheses were used in both arches. Barootchi et al[35] reported higher prosthetic survival rates for zirconia frameworks (93.7%) when compared to metal-acrylic prostheses (83%) at 5 years in a retrospective analysis. Bidra et al[38] reported 1.4% failure rate for zirconia complete-arch prostheses in a systematic review. In another analysis, Bidra et al[40] evaluated 2039 zirconia complete-arch prostheses and reported 99.3% 5-year survival rate with 6 fractures. Sadwosky[39] in a review concluded that complete-arch monolithic zirconia prostheses offer advantages over metal-resin prostheses. The 3Y-TZP zirconia was used in the present study in order to give a perspective to the reader on how a definitive prosthesis material would perform in the used test settings. The comparisons with this definitive framework material may allow for interpretations whether tested HDPs may possibly be used in the long term as an interim prosthesis material.

According to a previous report, forces on all contacting teeth during mastication range between 190 and 260 N.[32] The mean maximum bite forces evaluated in some previous studies were approximately 700 N.[32-34] The results of the present study suggested that all HDP specimens with a Ti base had load-to-failure values (483.10-563.54 N) below mean maximum bite forces reported in 3 studies, however, above the range for forces on all teeth when they are in contact.[32-34] According to Hagberg et al, the range of maximum posterior force is 300 N to 600 N.[41] When this range for maximum forces is considered, all tested HDPs' average load-to-failure values were within this range and particularly on the higher end of this range. It may be speculated that tested HDPs with cantilever, especially the ones with load-to-failure values on the higher end of this range, would survive the chewing function even though the specimens are thinner at the Ti base connector.

Lopes et al[42] found that bruxers showed a high prevalence of mechanical complications in both the provisional and definitive restorations when used for all-on-4 treatment concept. Cosme et al[43] evaluated the mean maximum bite forces for the nonbruxer (859 N) and bruxer (806 N) young dentate adults. And, all HDPs with a Ti base in the current study showed lower load-to-failure strength than those patient groups' mean maximum bite forces. Clinicians may need to use caution when using HDPs with Ti bases in tested or smaller dimensions are used for young adults and patients with parafunctional habits. However, the reference averages were maximum bite forces, which are different than chewing forces.

Lopes et al[42] reported the occurrence of mechanical complications (81.9%) and fracture of the acrylic resin provisional prosthesis (59.4%) when they rehabilitated the patients with all-on-4 concept. Malo et al[26] reported debonding of acrylic resin from the PEEK around the Ti base when used for complete-arch fixed PEEK-acrylic resin prostheses. Drago[9] reported prosthetic complications (19%) primarily being denture base and denture tooth fractures (adhesive and cohesive) for the patients rehabilitated with complete-arch, screw-retained, interim acrylic resin prostheses. Although there are in vivo[26] and in vitro[10,30] studies that report fractures around cantilevered areas, it is not possible to directly compare present study results with previous studies as they differ in design and materials.

Drago[1] evaluated the type and frequency of prosthetic complications related with immediately loaded, complete-arch, interim acrylic resin prostheses. The author reported denture base fracture at the Ti base region. In the present study, the loads which fractured the specimens were lower than the mean maximum bite forces reported previously [32-34] when Ti bases were used, and this is in line with the denture base fracture at the Ti base region mentioned in a previous study.[1] Drago[1] also described the duration that patients used the complete-arch, interim, acrylic resin prostheses as 3 months for mandibular implants and 4 months for maxillary implants prior to proceeding with definitive impressions. Drago[9] examined the relationship between the prosthetic complications and cantilever length and anterior to posterior spread of complete-arch, interim, acrylic resin prostheses as well. The cantilever segments were designed not to exceed the mesio-distal width of a molar to decrease the prosthetic complication rates related to the cantilever segments of complete-arch acrylic resin interim prostheses. Also, conventional heat-polymerized acrylic resin (SR Ivocap injection system; Ivoclar Vivadent Inc.) was used for the fabrication of prostheses in that study. In line with the design in Drago[9] study, in the present study, the cantilever segment was also designed not to exceed the mesio-distal width of a molar. However, the fabrication technique of the prostheses differs in these studies because CAD-CAM was used in the present study.

Thermocycling process was not applied in the present study because the primary purpose was to investigate the load-to-failure of the HDPs when Ti bases were used and not to analyze the bond strength between the Ti bases and the HDP. Further investigations with thermocycling and the use of chewing simulator that better simulate the oral environment should be performed. Water absorption of HDPs may alter their mechanical properties and cause a decrease in their resistance. In the present study, one type of composite resin material was used for the cementation of Ti bases and different outcomes may be seen when different resin luting agents are used. The width and thickness of the specimens were selected to simulate a worst-case scenario. Because the specimen thickness and width in this study are on the smaller end when the reported averages for the framework thicknesses are considered,[36] the tested materials may resist higher occlusal loads when thicker and wider specimens are used. Future studies are needed to investigate the effect of thickness on the strength of CAD-CAM HDPs and zirconia with Ti bases.

CONCLUSIONS

Within the limitations of this in vitro study, there were no significant differences among the load-to-failure performance of high-density polymer specimens with a Ti base. Zirconia (3Y-TZP) showed significantly higher load-to-failure values than the HDPs. The loads fractured the high-density polymers with Ti bases were close to the maximum occlusal bite forces recorded in previous clinical studies.

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TABLES

Table 1. Tested materials

Material	Code	Composition	Manufacturer
CopraTemp	CT	Poly(methyl methacrylate)	White Peaks Dental Systems GmbH&Co.KG.
Tempo-CAD	TC	Poly(methyl methacrylate)/ Pigmet	GDS Dental GmbH
Solid shade PMMA disc	TD	Poly(methyl methacrylate)	TD Dental Supply
M-PM Disc-pink	MPM	Poly(methyl methacrylate)/ Highly cross-linked filler and fibre free organic modified polymer network	Merz Dental GmbH
M-PM Disc-white	MPMW	Poly(methyl methacrylate)/ Highly cross-linked filler and fibre free organic modified polymer network	Merz Dental GmbH
FireZr	FZR	3Y-TZP	Glidewell Direct GmbH

FIGURES

Figure 1. High density polymer specimens with titanium bases

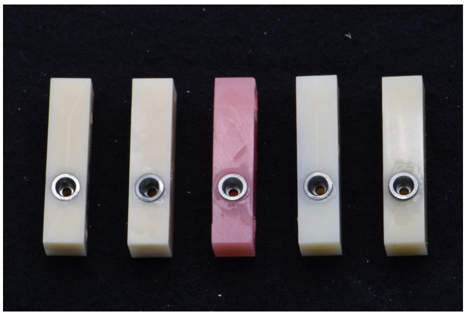

Figure 2. Schematic diagram of the test set-up. Rectangular prism-shaped specimen with a Ti base, illustrated in white, secured in clamp colored in gray. Load frame, colored in gray at the bottom, contacting the framework work 2 mm before the end of the specimen.

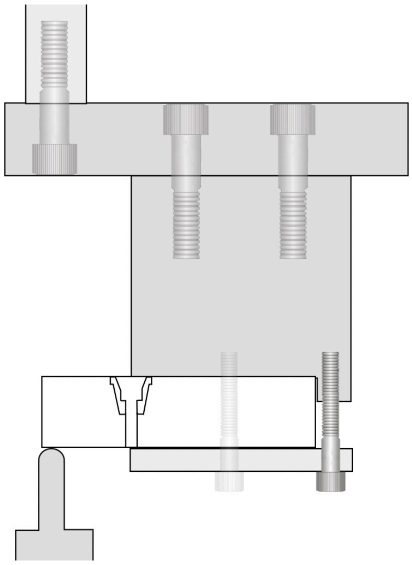

Figure 3. A, framework under static loading. B, framework at moment of fracture

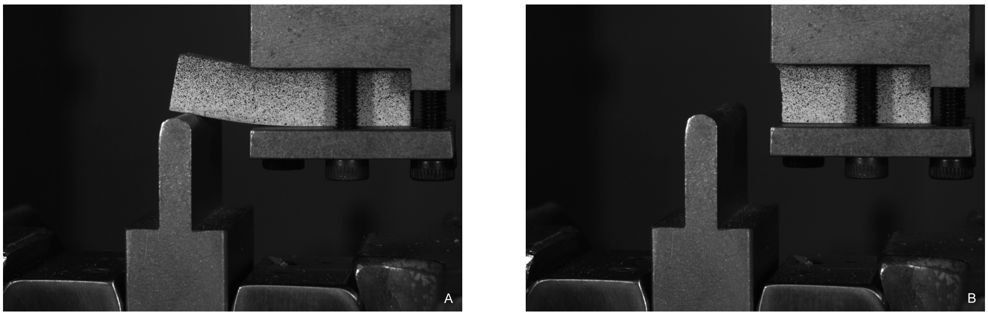

Figure 4. Mean maximum fracture values (N) and %95 confidence limits (N, Newton; CT, CopraTemp; MPM, M-PM Disc-pink; MPMW, M-PM Disc-white; TC, Tempo-CAD; TD, TD Solid shade PMMA disc; FZR, FireZr)

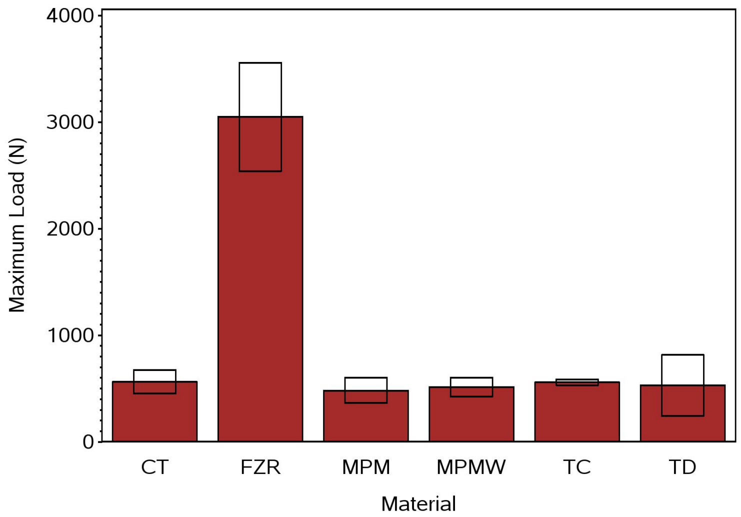

Figure 5. Load-displacement curves for HDPs (CT, MPM, MPMW, TC, TD) and zirconia (FZR)

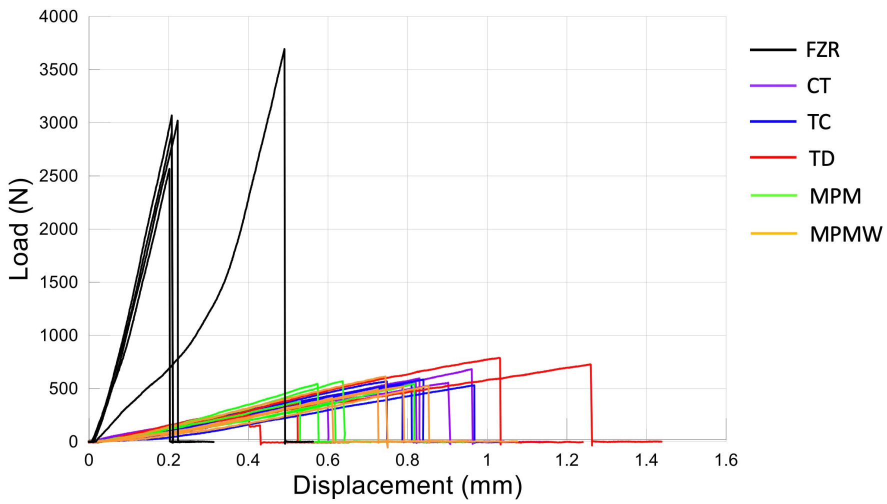

Figure 6. Load-displacement curves for monolithic HDPs (CT, MPM, MPMW, TC, TD) and zirconia (FZR) without Ti bases