Biomechanical behavior of CAD/CAM cobalt-chromium and zirconia full-arch fixed prostheses

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PURPOSE. To verify the influence of computer-aided design/computer-aided manufacturing (CAD/CAM) implant-supported prostheses manufactured with cobalt-chromium (Co-Cr) and zirconia (Zr), and whether ceramic application, spark erosion, and simulation of masticatory cycles modify biomechanical parameters (marginal fit, screw-loosening torque, and strain) on the implant-supported system. **MATERIALS AND METHODS.** Ten full-arch fixed frameworks were manufactured by a CAD/CAM milling system with Co-Cr and Zr (n=5/group). The marginal fit between the abutment and frameworks was measured as stated by single-screw test. Screw-loosening torque evaluated screw stability, and strain analysis was explored on the implant-supported system. All analyses were performed at 3 distinct times: after framework manufacturing; after ceramic application in both materials' frameworks; and after the spark erosion in Co-Cr frameworks. Afterward, stability analysis was re-evaluated after 10^6 mechanical cycles (2 Hz/150-N) for both materials. Statistical analyses were performed by Kruskal-Wallis and Dunn tests (α=.05). **RESULTS.** No difference between the two materials was found for marginal fit, screw-loosening torque, and strain after framework manufacturing (P>.05). Ceramic application did not affect the variables (P>.05). Spark erosion optimized marginal fit and strain medians for Co-Cr frameworks (P<.05). Screw-loosening torque was significantly reduced by masticatory simulation (P<.05) regardless of the framework materials. **CONCLUSION.** Co-Cr and Zr frameworks presented similar biomechanical behavior. Ceramic application had no effect on the biomechanical behavior of either material. Spark erosion was an effective technique to improve Co-Cr biomechanical behavior on the implant-supported system. Screw-loosening torque was reduced for both materials after masticatory simulation. ***[J Adv Prosthodont 2020;12:329-37]***

KEYWORDS: Computer-aided design/computer-aided manufacturing (CAD/CAM); Dental marginal adaptation; Dental stress analysis; Implant-supported dental prostheses; Spark erosion

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INTRODUCTION

The choice of implant-supported fixed dental prostheses for complete edentulous arches has increased over the years.[1-3] Anatomic curved arches can result in more manufacturing distortions,[4,5] and a strict control of the prosthesis manufacturing process, including the attempt to achieve passive fit and greater stress distribution, has been encouraged in dental rehabilitation treatments. Therefore, biomechanical evaluation should be investigated, initially from the tightening of the prosthetic screw, in which the distribution of stress occurs in the implant supported component system.[6,7]

In attempts to achieve improved fit between framework and implant platforms, different manufacturing techniques have been developed, such as soldering,[4] spark erosion,[8] and elimination of laboratorial phases of the lost-wax technique through CAD/CAM milling technology.[9] This technology, also classified as subtractive manufacturing, uses computer numerical control (CNC) machining with diamond burs that cut parts from a prefabricated block.[10] This method is reported to promote results superior to those achieved by the lost-wax technique,[10-12] by reducing some clinical and laboratory steps.[13] Also, the technique can be conducted in materials such as titanium and its alloys, zirconia (Zr) and cobalt-chromium alloys (Co-Cr).[12,14] Co-Cr alloys have been recognized for their low cost, high elastic modulus, easy ceramic application, ease of casting, and the possibility of over-cast components,[15] while Zr has joined this field due to the esthetic demands of patients and clinicians along with its superior strength and rigidity.[2,16] In the short term, the failure rate associated with the use of Zr is very low.[16] However, results regarding the marginal fit and stress distribution of Zr full-arch fixed dental prostheses (FAFDPs) are widely discussed,[6,10,12,17,18] since corrective techniques cannot be applied in Zr frameworks.

Irrespective of the framework material, ceramic application has been proven to increase distortion in implant-supported prostheses.[17,19,20] Some possible explanations are regarding material type, prostheses' size, ceramic firing cycles, and their temperature.[17,20] In this context, laser welding and spark erosion are possible alternatives to improve the adaptation of the implant-supported system.[8,20,21] Spark erosion preserves the original resistance of the prosthesis and its veneering porcelain coverage as it incorporates the refinement of the margin framework by electrical discharges, correcting adaptation without framework sectioning.[8,20] However, the authors are unaware of studies in which spark erosion application has been used in CAD/CAM milling prostheses.

Beyond the attempt to achieve adequate marginal fit values, prosthetic screw-loosening is a current drawback in implant rehabilitation.[22,23] In addition, external forces such as chewing accelerate the process of loosening torque, due to the movement of the sliding screw threads. The consequence is the presence of an instability that decreases the preload to a critical level.[7,24] The effects of these external forces can be observed by a mechanical cycle device, which is used to simulate masticatory function.[15,25] A previous study[25] reported that screw-retained zirconia FAFDPs supported by six implants presented higher torque loss when compared with titanium FAFDPs. However, the comparison with Co-Cr frameworks with four implants has not been previously evaluated.

Hence, this *in vitro* study aimed to investigate: (1) the effect of framework materials (Co-Cr and Zr) on the biomechanical behavior of subtractive CAD/CAM FAFDPs (marginal fit, stability, and strain); (2) the influence of ceramic application on biomechanical behavior; (3) the effect of spark erosion on subtractive CAD/CAM FAFDP Co-Cr prostheses; and (4) screw-loosening stability after masticatory simulation. The null hypotheses were that (1) framework materials (Co-Cr and Zr) would have no effect on the biomechanical behavior of subtractive CAD/CAM FAFDPs; (2) no differences in biomechanical behavior would be found after ceramic application for the two material; (3) spark erosion would not improve marginal fit, stability, and strain in Co-Cr prostheses; and (4) no differences would be found in the stability of either material after masticatory simulation.

MATERIALS AND METHODS

A fully edentulous maxillary replica (master model) was prototyped from the database of the Renato Archer Information Technology Center (Campinas, Sao Paulo, Brazil).[26] All-on-four implant concept was used in the master model: 2 implants (Easy Grip Porous, 4.1 × 11.5 mm; Conexão - Sistemas de Prótese Ltd., Aruja, Sao Paulo, Brazil) placed in the lateral incisor regions and 2 implants (Easy Grip Porous, 4.1 × 13 mm; Conexão - Sistemas de Prótese Ltd.) tilted distally at 30° placed in the second premolar area.[26] Micro-unit abutments (mini-abutment 4.1 × 4.0 mm; Conexão - Sistemas de Prótese Ltd.) were screwed onto the anterior implants, and 30-degree-inclined mini-abutments (angled mini-abutment 4.1 × 4.0 mm; Conexão - Sistemas de Prótese Ltd.) were screwed onto the tilted implants. Twenty Ncm torque was applied in each of the four abutments (Torque Meter TQ-8800; Lutron, Taipei, Taiwan) and designated as abutments A, B, C, and D (Fig. 1).

The study was divided into four phases: initial (1st phase); ceramic application (2nd phase); spark erosion (3rd phase); and masticatory simulation (4th phase). Ten frameworks were manufactured by CAD/CAM milling system in Co-Cr (n = 5) and Zr (n = 5), simulating maxillary FAFDPs. The initial marginal fit, screw-loosening torque, and the strain induced on the implant analogs were analyzed (1st phase). Ceramic application was conducted in all frameworks, and

Fig. 1. Master model.

the tests were repeated (2nd phase). Spark erosion was performed in Co-Cr frameworks, followed by the repetition of the tests (3rd phase). Masticatory simulation was conducted to evaluate screw-loosening torque in all frameworks (4th phase) (Fig. 2).

A light scanner (Ceramill map 400+, Amann Girrbach, Koblach, Germany) scanned the master model with scan bodies (Scan Connect Micro-unit, Conexão - Sistemas de Prótese Ltd.) tightened on the abutments, and the framework's digital file was designed (Ceramill Mind software, Amann Girrbach). The specimens were obtained from Co-Cr blocks (Starbond Easy Disc; Scheftner, Mainz, Germany) (61% Co, 27.5% Cr, 8.5% W, 1.6% Si, < 1% C, < 1% Mn, < 1% Fe, and elastic modulus of 191 GPa) (n = 5) by a milling machine (CNC D15W; Yenadent, Istanbul, Turkey), as were the pre-sintered yttrium-stabilized tetragonal zirconia polycrystal blocks (ZirkOM SHT; Aidite (Qinhuangdao) High-technical Ceramics, Qinhuangdao, China) (94.39% ZrO_2, 5.3% Y_2O_3, 0.31% other oxides, and elastic modulus of 200 GPa) (Ceramill Motion 2; Amann Girrbach) (n = 5). Afterward, the Zr frameworks were finished in a sintering furnace (Ceramill Therm S; Amann Girrbach), at 1530°C for 12 hours, following the manufacturer's instructions. Thus, Co-Cr and Zr frameworks were obtained. The height of both frameworks was approximately 10 mm (Figs. 3A, 3B).

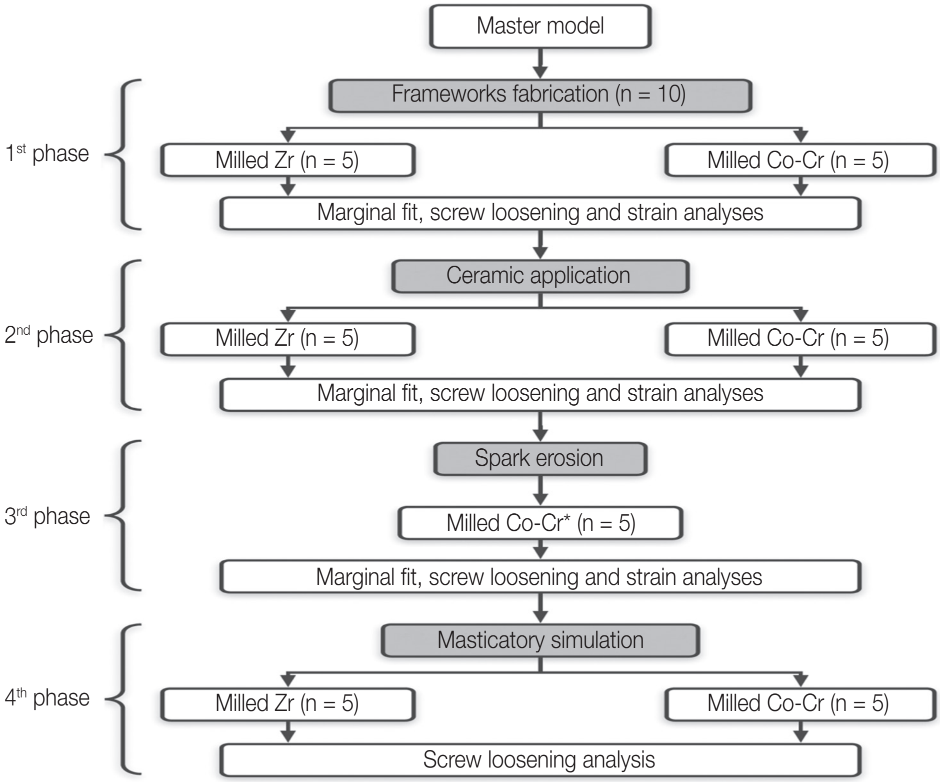

Fig. 2. Flowchart of study methodology design. *Zr frameworks did not receive spark erosion, since it can only be conducted on metallic frameworks.

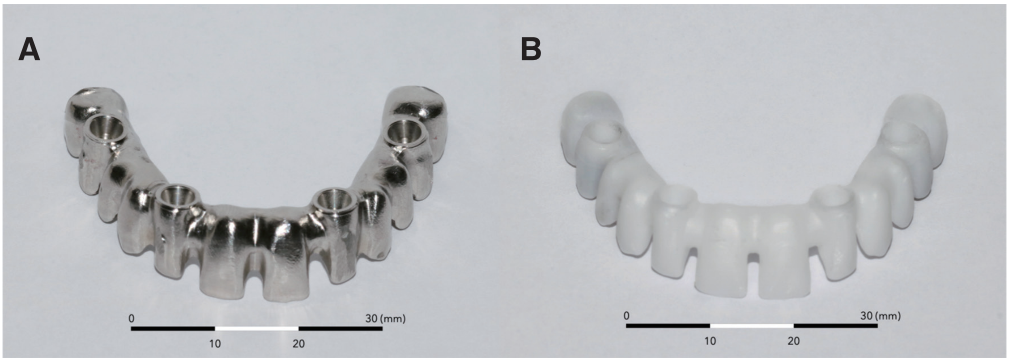

Fig. 3. Full-arch fixed dental framework from (A) Cobalt-Chromium, (B) Zirconia groups. The scale describes the size of the frameworks in width.

The marginal fit values were obtained by the single-screw test protocol.[15,20,22,27] Each framework was settled on the master model, with the help of a digital torque meter (Torque Meter TQ-8800; Lutron), and the screw at one extremity (abutment A) received a 10-Ncm torque. Values of all abutments (A, B, C, and D) at the framework/abutment interface were assessed on both buccal and lingual sides (in the mini-abutment platform center) (Fig. 4). The measurement was repeated with the abutment D tightened, resulting in 48 measurements to obtain a mean fit value for each framework.[15,20,22,27] The examiner was previously calibrated (T.B.) (0.996 intra-class correlation coefficient, $P < 0.001$) to perform the fit measurements, using a microscope at 120× magnification (VMM-100-BT; Walter UHL, Aßlar, Germany) associated with a digital camera (KC-512NT; Kodo Eletronics, Seoul, Korea) and a Quadra-check (QC 220-HH; Metronics Inc., Cincinnati, OH, USA).

The stability of the system was measured by screw-loosening torque in specific working models. Transfers were screwed on the master model and splinted with drills and low shrinkage self-curing acrylic resin (Pattern Resin LS, GC, Tokyo, Japan). The set was positioned into the silicone impression with the help of a parallelometer, after which the self-curing acrylic resin (Class Mold - Classico Artigos Odontológicos, Campo Limpo Paulista, Sao Paulo, Brazil) was poured. The framework was positioned on the working model, and each screw received a 10-Ncm torque in A-D-B-C sequence.[15] Ten minutes later, the screws were retightened with 10-Ncm, and screw-loosening torque values were measured after 24 hours[23] (Torque Meter TQ-8800; Lutron). The average value was determined for each framework and each group. New screws were used for each framework in each study phase.

Strain-gauge analysis followed the methodology previously described.[15,28] The mesial side of each modified conical implant abutment analog received one strain gauge (PA-06-060-BG-350 L; Excel Sensores Ltd., Taboao da Serra, Sao Paulo, Brazil), which was bonded with cyanoacrylate-based glue (Loctite Super Bonder; Henkel, Düsseldorf, Germany).[15,28] This set was associated with a plaster type IV dental stone (Durone IV, Dentsply Sirona, York, PA, USA) manipulated following the manufacturer's recommendations. One-quarter Wheatstone bridge electric circuit was mounted with temperature control (Fig. 5).[15,28] Then, each framework was settled in the modified analogs, the screws received a torque of 10-Ncm, similar to the same sequence described in the screw-loosening torque analysis, and the mean µstrain value was measured.[15] After the torque of all screws, the µstrain was recorded for 10 minutes using a specific equipment (ADS 2000, Lynx Tecnologia Eletronica, Sao Paulo, Brazil), with data processing software (AqAnalysis 2000, Lynx Tecnologia Eletronica, Sao Paulo, Brazil). The strain gauges were reset between each new analysis to standardize the readings.[6,15,28] The µstrain average value was determined for each framework and each group.

Ceramic application and related firings were conducted for all frameworks by an experienced dental technician according to manufacturer's recommendations (InSync ZrO_2, Jensen Dental, North Haven, CT, USA for Zr frameworks; and feldspathic InSync MC, Jensen Dental, for Co-Cr frameworks) (2nd phase). Co-Cr and Zr frameworks received five firing cycles in total (Table 1). Initially, a liner was applied to the Zr frameworks, followed by two dentin layers, one enamel layer to finalize the tooth shape, and glaze (Fig. 6). Co-Cr frameworks were initially sandblasted with 150 µm Al_2O_3 particles (Wilson, Polidental, Cotia, Sao Paulo, Brazil) at pressure of 3 bars and distance of up to 5 cm. Residual Al_2O_3 particles were removed from the frameworks with water. Two opaque porcelain layers, followed by one dentin layer, one enamel layer, and glaze, were applied.

Spark erosion (3rd phase) followed the methodology previously described.[8,20] The procedure could be conducted

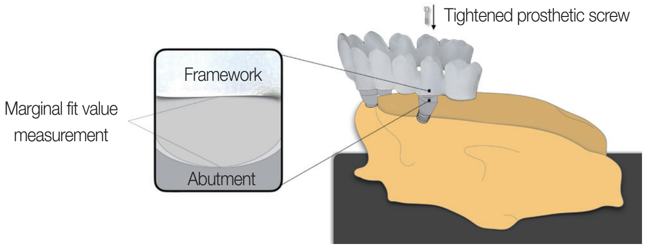

Fig. 4. Schematic representation of single-screw test. Prosthetic screw is tightened in the abutment located on the opposite side (abutment D). The marginal fit value measurement is obtained at the framework/abutment interface on buccal side (enlarged view on the left). Values of all abutments (A, B, C, and D) were assessed on both buccal and lingual sides.

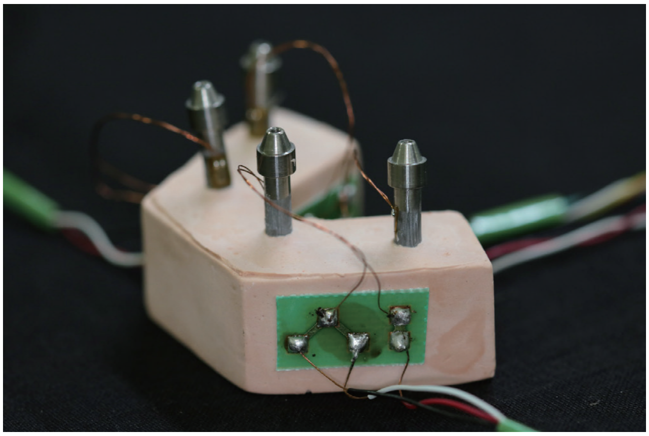

Fig. 5. Electric circuit in 1/4 Wheatstone bridge at strain-gauge model.

Table 1. Firing cycles and heat rate for cobalt-chromium (Co-Cr) and zirconia (Zr) frameworks according to the manufacturers' recommendations

Procedure	Co-Cr Firing temperature (Initial - End) (°C)	Co-Cr Heat rate (°C/min)	Zr Firing temperature (Initial - End) (°C)	Zr Heat rate (°C/min)
Liner	-	-	400 - 970	60
1st opaque layer	450 - 950	80	-	-
2nd opaque layer	450 - 950	80	-	-
1st dentin layer	450 - 880	60	400 - 765	40
2nd dentin layer	-	-	400 - 760	40
1st enamel layer	450 - 800	60	400 - 700	45
Glaze	450 - 780	50	400 - 730	45

Fig. 6. Zirconia framework with ceramic application.

only for Co-Cr frameworks because of their electrical conductivity. The framework was placed in the spark erosion model and low shrinkage self-curing acrylic resin increments (Pattern Resin LS, GC, Tokyo, Japan) were used to connect the framework to the machine (Form 2-LC ZNC, Charmilles Technologies, Geneva, Switzerland). The model and framework were surrounded by dielectric fluid (Eletron, Archem Quimica Ltda, Araras, Sao Paulo, Brazil). The amperage used started with 3 A, and finished with 1 A. The vertical movement of the equipment shaft was controlled and uniform to those of all Co-Cr frameworks.

For the 4th phase (masticatory simulation), Co-Cr and Zr frameworks were placed in their respective working models. New screws were used with 10-Ncm torque tightening, by means of a digital torque meter (Torque Meter TQ-8800; Lutron). These new screws received the same retightening protocol previously described for the stability analysis. One million (10^6) mechanical cycles[29] (Mechanical Fatigue Simulator ER11000 Plus, ERIOS, São Paulo, Brazil) were applied to all prostheses, with 2 Hz frequency,[30] a 150 N[23] oblique (30 degree angle)[30] load, directed on the right first molar occlusal surface. Artificial saliva at 37°C (1.5 mM Ca, 3.0 mM P, 20.0 mM $NaHCO_3$, pH 7.0) was used to immerse the prostheses.[15]

A priori sample size calculation was performed using data from Presotto *et al.* (2018).[8] Considering a minimum difference to be detected of 10 μm with a standard deviation of 2 μm (20%) in the variable "marginal fit", five frameworks per group would provide 95% power with a 5% significance level. All variables were analyzed using the D'Agostino and Pearson test to verify normality and the Brown-Forsythe and Bartlett's tests to determine the homoscedasticity of their variances. Kruskal-Wallis and Dunn tests evaluated the influence of the material (Co-Cr and Zr) and time (initial, ceramic application and spark erosion) on the marginal fit and strain values. On the screw-loosening torque values, evaluation time was analyzed at initial, ceramic application, spark erosion, and masticatory simulation times. All data were presented as the median and interquartile range (IQR). GraphPad Prism 8.0 software (GraphPad Software Inc., San Diego, CA, USA) was used for statistical analysis, and the significance level was set at 5%.

RESULTS

Marginal fit medians and IQR values are presented in Figure 7. Marginal fit median of Co-Cr frameworks after spark erosion (median 49.25, IQR 37.88 - 67.38 μm) was significantly lower than initial Zr (149.1, IQR 37.78 - 229.9 μm), ceramic veneered Zr (148.1, IQR 72.94 - 305.5 μm), initial Co-Cr (97.38, IQR 39.0 - 193.8 μm) and ceramic veneered Co-Cr (189.6, IQR 42.34 - 226.3 μm) medians (Kruskal-Wallis, H = 26.91; df = 4, $P < .0001$). Marginal fit medians between the Zr and Co-Cr at initial and ceramic veneered times were not significantly different ($P > .05$).

Screw loosening torque medians and IQR values are presented in Figure 8. Screw-loosening median of Co-Cr and Zr frameworks after mechanical cycling (Co-Cr median

7.1, IQR 6.4 - 7.8 and Zr median 6.5, IQR 6.1 - 6.9 Ncm) was significantly lower than those of ceramic veneered Zr (8.0, IQR 7.1 - 8.5 Ncm) and spark eroded Co-Cr (8.1, IQR 6.9-9.5 Ncm) medians (Kruskal-Wallis, H = 21.31; df = 4, *P* = .0016). There were no differences in screw-loosening torque between the Zr and Co-Cr at initial, ceramic veneered and mechanical cycled times (*P* > .05).

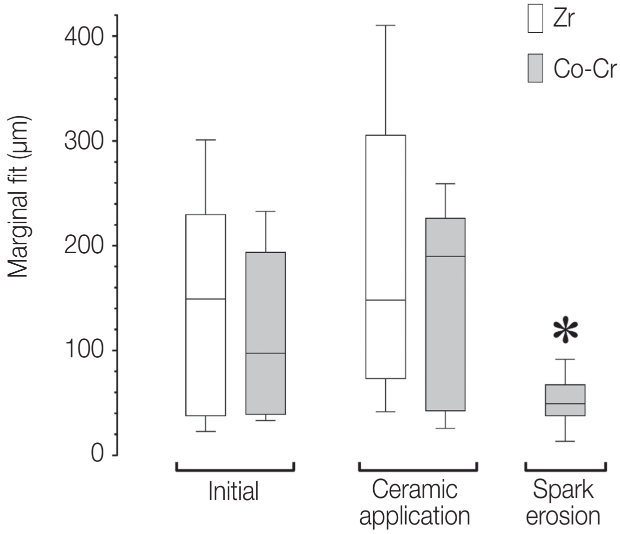

Fig. 7. Levels of marginal fit (µm) for Zr and Co-Cr groups at each evaluation time. Central bar = median value; boxes = first and third quartiles; whiskers = maximum and minimum values. *Significant difference between spark erosion Co-Cr group and the initial Zr, ceramic veneered Zr, initial Co-Cr and ceramic veneered Co-Cr (*P* < .05).

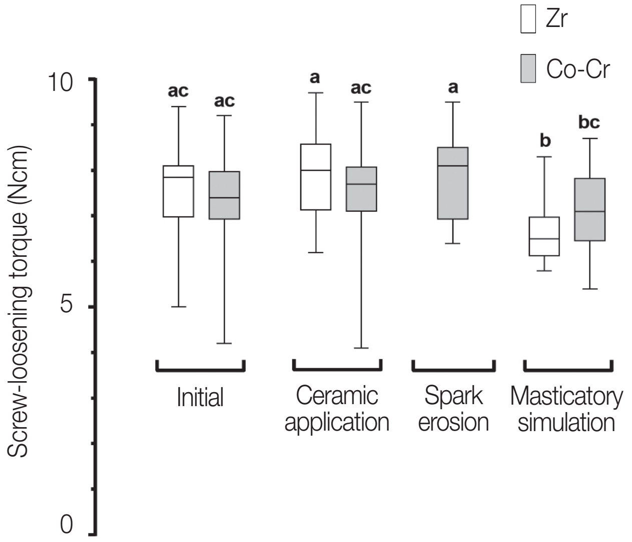

Fig. 8. Levels of screw-loosening torque (Ncm) for Zr and Co-Cr groups at each evaluation time. Central bar = median value; boxes = first and third quartiles; whiskers = maximum and minimum values. Different lowercase superscript letters indicate statistically significant differences between the materials and evaluation time (*P* < .05).

Strain medians and IQR values are presented in Figure 9. Strain median of Co-Cr frameworks after spark erosion (median 53.77, IQR 17.68 - 159.8 µstrain) was significantly lower than initial Zr (311.6, IQR 147.8 - 706.2 µtrain), ceramic veneered Zr (306.7, IQR 95.41 - 538.7 µstrain), initial Co-Cr (310.1, IQR 115.2 - 458.5 µstrain) and ceramic veneered Co-Cr (267.7, IQR 114.0 - 418.8 µstrain) medians (Kruskal-Wallis, H = 23.86; df = 4, *P* < .0001). There were no differences in strain between the initial Zr and ceramic veneered Zr (*P* = .43), between initial Co-Cr and ceramic veneered Co-Cr (*P* = .99), between initial Zr and initial Co-Cr (*P* = .32), and between ceramic veneered Zr and ceramic veneered Co-Cr (*P* = .84).

DISCUSSION

The comparison between materials for a consolidated technology such as subtractive CAD/CAM manufacturing should not be neglected. CNC technology has been recognized as a reliable technique for Co-Cr[11] and Zr[10] framework manufacturing, and once no statistically significant differences were found in aspects of the biomechanical behavior, the first null hypothesis - material frameworks would have no effect on the biomechanical behavior of subtractive CAD/CAM FAFDPs - was accepted.

Material influence was not observed for marginal fit, stability, and strain-gauge analyses. The discussion about passive fit centers around values up to 150 µm,[22] and up to 230 µm between the implants and framework;[3] therefore, at the initial time the values found can be considered within the

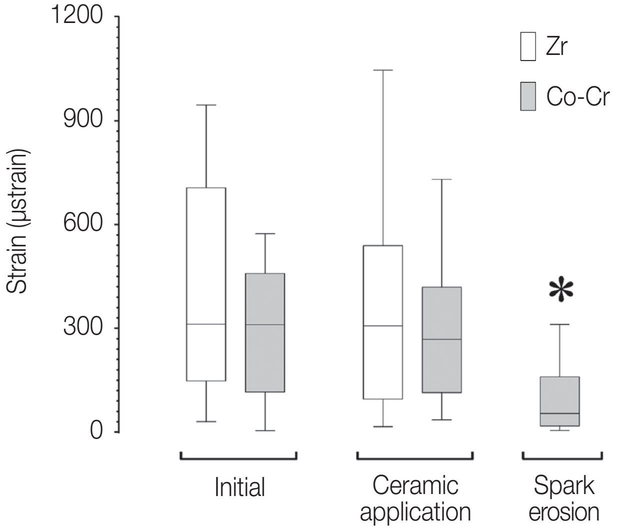

Fig. 9. Levels of strain (µstrain) for Zr and Co-Cr groups at each evaluation time. Central bar = median value; boxes = first and third quartiles; whiskers = maximum and minimum values. *Significant difference between spark erosion Co-Cr group and the initial Zr, ceramic veneered Zr, initial Co-Cr and ceramic veneered Co-Cr (*P* < .05).

clinically acceptable. A study with 3-unit, implant-supported, screw-retained frameworks also found no marginal fit statistical difference between materials.[12] Although elasticity modulus of Zr is higher than that of Co-Cr, the milling process was conducted in a soft block of Zr in which the wear of burs does not occur on a large scale.[5,31] The use of a high-speed 5-axis machine is reported to be able to reduce errors during milling, ensuring finishing and polishing procedures.[13,32] A wasted bur or a non-calibrated machine could contribute to less accurate milling.[32] Another factor that could influence marginal fit is the sintering process, mandatory when milling in soft blocks of Zr. This Zr post-treatment is associated with 20 - 25% of the material shrinkage,[14] and could induce uncontrolled distortions.[17] The use of Zr frameworks for long-term prostheses should still be indicated with caution, because neither soldering nor spark erosion can be performed to improve biomechanical behavior; however, promising results were found as no difference in marginal fit, stability, and strain was noted between the two materials. Studies[18,27] with smaller frameworks described a relationship in which lower fit could result in higher stress/strain in the implant-supported system rather than the material type, corroborating our findings. In a finite element analysis (FEA) study,[26] stiffer materials such as Co-Cr and Zr supported by four implants did not exceed the bone resistance limit; thus the median strain values found in this study might not interfere with osseointegration, since bone is reported to have a remodeling process.[15,33] Regarding stability, the protocol applied was described to be effective with casted Co-Cr,[23] and the results of this study confirmed it to be effective with milled CAD/CAM frameworks, irrespective of whether the material was Zr or Co-Cr.

The importance of ceramic application is related to the need for esthetics in Co-Cr frameworks and protection for Zr frameworks, since the literature describes low-temperature aging as a complication during function.[34] Positive results have been found after ceramic application for both materials, leading to acceptance of the second hypothesis, since no differences in biomechanical behavior were found after ceramic application for both materials. In a previous study,[35] milled Zr frameworks supported by six implants were porcelain-veneered through three firing cycles, and no influence on the fit was found. The findings of this study added that ceramic application did not alter the values of marginal fit, screw-loosening torque and strain from the initial phase, even when five firing cycles were conducted. Thus, CNC technology is suitable for manufacturing consistent frameworks, reducing the chances of distortions being induced after veneering coverage.

One alternative to reduce distortions in prostheses is spark erosion. To the best of the authors' understanding, no study was found evaluating this option in Co-Cr full-arch implant-supported prostheses manufactured by CAD/CAM milling technology. Since one of the principles of CAD/CAM is the manufacture of frameworks without the need for sectioning and soldering, the use of this fit-corrective technique for metallic frameworks would preserve the prosthesis integrity and its ceramic veneering.[8,20] The literature describes ceramic veneering as being responsible for inducing distortions in the frameworks.[8,17,19,20] In this study, fortunately, ceramic application did not influence the biomechanical behavior of the frameworks, but the importance of spark erosion evaluation was to clarify whether prostheses manufactured by subtractive CAD/CAM systems could be improved. Spark erosion has been described as effective for prostheses with poor fit values, such as those fabricated by conventional casting.[8,20] However, maintaining the clinical relevance, it is important to know that if a Co-Cr milled framework presents lower levels of fit, it can be improved by using spark erosion, without the need to manufacture a new framework, which would increase treatment costs and time. This thought can be extended to ceramic application. Different veneering materials, firing cycles, machines and technician ability might promote distortions that could be decreased without a new prostheses manufacturing. Therefore, the third hypothesis - spark erosion would not improve fit, stability, and stress distribution in Co-Cr subtractive CAD/CAM prostheses - was rejected, since higher levels of fit and lower values of strain were found. Although screw stability was not significantly increased after spark erosion, it was not reduced.

The evaluation of titanium prosthetic screw-loosening for each framework manufacturing material was conducted because lower marginal fit between framework and abutment and external forces could influence and hinder preload maintenance.[7] The screw-loosening process is a customary event in implant therapy,[23,25] albeit it has been reported as preceding other severe mechanical and biological drawbacks,[36] decreasing the longevity of the prosthesis. The tightening protocol aimed to prevent the partial loss of torque,[23] which occurs regardless of external forces, mainly in the first minutes after screw-tightening.[37] However, the fourth hypothesis - that no difference would be found between the stability of either material after masticatory simulation - was rejected. Adequate median values for screw-loosening torque were found at all times,[23] but the main difference found was between the last phase of the study (masticatory simulation) and previous evaluation times. One million mechanical cycles were conducted based on results from earlier studies,[15,29] and our results are in compliance with those of another study,[25] which also attributed the lower screw-loosening values of Zr frameworks to screw deformation after fatigue testing and to slight thread changes, exacerbated when in contact with material of higher elastic modulus, such as Zr, and joint vibration, which can reduce the preload. Nevertheless, it is important to state that even with significant screw-loosening values, regardless of the framework materials, none of the samples were visually apparently screw-loosened, which could indicate that the final preload in the screws may stand for a year of function.[25] Ceramic chipping, which has been reported as one of the main technical complications of Zr-based prostheses,[38] was also not shown. The effect of mechanical loading in milling CAD/CAM FAFDPs may be confirmed by further studies with different masticatory simulation times.

It is agreed that Co-Cr and Zr are considered stiffer materials when compared with other materials used to manufacture dental frameworks, such as titanium and its alloys. According to the results found and previous literature,[2,16,38] it is possible to suppose that both stiffer materials may be suitable for manufacturing extensive prostheses in clinical practice, as the CAD/CAM milling technology enables the fabrication of better fitting frameworks that are resistant to distortions. Stress evaluation on the frameworks and bone region was conducted in a previous FEA study,[26] which corroborates our findings. The authors tested Co-Cr, Zr, and titanium materials as implant-supported frameworks on a simulated maxilla bone region. They found higher stress levels in Co-Cr and Zr frameworks than in titanium frameworks, and also found that both stiffer materials transmitted less stress levels for implants, screws, abutments, and cortical bone.[26] Under acceptable fit conditions, materials such as Co-Cr and Zr are more resistant to the bending moment during masticatory forces, and the stress distribution to the implant-supported system is mitigated and better distributed. On the other hand, a stiffer material associated with an unsatisfactorily fitted prosthesis may cause complications such as screw-loosening, catastrophic chipping and/or bone reabsorption more quickly than expected. The FEA study[26] did not simulate ceramic coverage, which was evaluated in our study. The association of the results between our study, this previous FEA study, and literature would seem to support the reliable and adequate biomechanical behavior of stiffer materials for FAFDPs.

Each phase of this study was evidence-based dentistry. However, limitations can be related to the use of only one screw during fit measurements, which could decrease the fit values in the non-tightened abutments and mask the real situation; the use of work models for screw-loosening analysis, instead of the master model; and the measurement of strain only where the strain gauge was bonded, although it was to avoid a bias of bond in different positions in each framework.[15]

Overall, biomechanical studies play an important role in the literature. Through their initial *in vitro* view, the development and improvement of materials and technologies can be conducted. The esthetic characteristics of Co-Cr are in contrast to its low cost and reduced hands-on requirement for dental technicians. On the other hand, Zr has superior esthetic characteristics in contrast to its chipping and degradation at low temperatures. Future research comparing FAFDPs with other materials such as titanium alloys, 3D printing CAD/CAM systems, more implants supporting rehabilitation such as the all-on-six concept,[26] and longitudinal follow-ups of prostheses fabricated with these materials are necessary and could clarify the applicability of the findings of this study.

CONCLUSION

Milled Zr maxillary FAFDPs presented similar biomechanical behavior when compared with milled Co-Cr maxillary FAFDPs. Ceramic application did not affect the biomechanical behavior of Zr and Co-Cr FAFDPs supported by four implants. Spark erosion represented an effective technique to optimize the biomechanical behavior of milled Co-Cr FAFDPs. One year of masticatory simulation can reduce the prosthetic screw-loosening torque of milled Zr and Co-Cr prostheses supported by the all-on-four implant concept.

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