Fracture resistance and marginal fit of the zirconia crowns with varied occlusal thickness

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PURPOSE. The present study aimed to evaluate the clinical applicability of monolithic zirconia (MZ) crowns of different thickness via determination of fracture resistance and marginal fit. **MATERIALS AND METHODS.** MZ crowns with 0.5, 0.8, 1.0, and 1.5 mm thickness and porcelain fused to metal (PFM) crowns were prepared, ten crowns in each group. Marginal gaps of the crowns were measured. All crowns were aged with thermal cycling (5 - 55^{o}C/10000 cycle) and chewing simulator (50 N/1 Hz/lateral movement: 2 mm, mouth opening: 2 mm/240000 cycles). After aging, fracture resistance of crowns was determined. Statistical analysis was performed with one-way ANOVA and Tukey's HDS post hoc test. **RESULTS.** Fracture loads were higher in the PFM and 1 mm MZ crowns compared to 0.5 mm and 0.8 mm crowns. 1.5 mm MZ crowns were not broken even with the highest force applied (10 kN). All marginal gap values were below 86 µm even in the PFM crowns, and PFM crowns had a higher marginal gap than the MZ crowns. **CONCLUSION.** The monolithic zirconia exhibited high fracture resistance and good marginal fit even with the 0.5 mm thickness, which might be used with reduced occlusal thickness and be beneficial in challengingly narrow interocclusal space. ***[J Adv Prosthodont 2020;12:283-90]***

KEYWORDS: Chewing simulator; Dental marginal adaptation; Fracture resistance; Porcelain; Zirconia

INTRODUCTION

Yttria tetragonal zirconia polycrystal (Y-TZP) is one of the most durable materials for a fixed dental prosthesis with high fracture resistance and improved aesthetic outcomes compared to porcelain fused to metal (PFM) restorations.[1] However, a major disadvantage of Y-TZP is the fracture of the veneer porcelain.[1-3] In veneered restorations using Y-TZP material, veneer chipping may occur in the long-term use compared to PFM restoration.[2] The innovative technological systems made it possible to fabricate zirconia restorations as monolithic zirconia (MZ) without porcelain fuse, which also eliminated the disadvantages of veneered zirconia.[4] In literature, studies regarding the strength of the full-contour zirconia of different thicknesses in clinical use was lacking.

Aesthetics,[5] mechanical strength,[6] and fit of the zirconia,[7] especially in monolithic restorations, depend on the thickness of the material. Recently, Ahmed *et al.*[7] revealed that the thickness of the zirconia, along with the sintering procedure and the finishing line width, affected marginal gap of the crowns even with the same zirconia type and manufacturing. Schriwer *et al.*[8] also demonstrated a decreased fracture resistance with an increased marginal gap, which was caused by different zirconia composition and manufacturing methods in MZ. The thickness of the zirconia might be considered as the increase in mechanical resistance. Sun *et al.* reported that MZ restorations had higher fracture load compared to the veneered zirconia and PFM restorations. The fracture load was reported to increase with MZ thickness; however, even the 0.6 mm thickness zirconia restorations had increased fracture load compared to the clinical occlusal biting forces.[6] In addition to mechanical resistance, Turkoglu and Sen suggested that thick zirconia structures (over 1 mm) resulted in lower Vicker hardness of the zirconia material and early cement resolution in the crown which might compromise the long-term use of restorations and

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cause complications.[9]

Preserving tooth structure is a significant principle of tooth preparation. PFM restorations require 1.5 - 2.00 mm tooth tissue reduction.[10] However, thin metals might not provide adequate resistance against occlusal forces and exhibit a poor fit and mechanical properties. MZ, with no requirement of veneering porcelain, eliminates further tissue loss.[11] Therefore, materials that could provide sufficient strength and mechanical properties even with the unusually thin structures might be beneficial. In this regard, the primary purpose of the present study was to evaluate the clinical applicability of MZ crowns with different thicknesses tested for the fracture load and marginal fit. In the present study, one type of MZ, whose minimal occlusal thickness was recommended as 0.5 mm in the manufacturer's instructions, was studied.[12] Thus, zirconia thicknesses were chosen as 0.5 mm, 0.8 mm, 1.0 mm, and 1.5 mm, with a minimum thickness of 0.5 mm and a maximum thickness of 1.5 mm. The hypothesis of the present study was that different zirconia thicknesses would not change the marginal compatibility of zirconia crowns, but the fracture resistance would increase with increasing thickness.

MATERIALS AND METHODS

Four MZ groups with a crown thickness of 0.5 mm, 0.8 mm, 1.0 mm, and 1.5 mm and a control group as PFM of 1.5 mm were created. In this study, a total of 50 samples were obtained, 10 samples in each group, with 80 strengths, 5% margin of error, and 0.56 effect size.[13] G * power 3.1.9.4 version was used for sample size calculation.

The mandibular right first molar tooth on an acrylic jaw (Ivoclar-Vivadent, Schaan, Liechtenstein) was prepared as the abutment tooth considering the tooth preservation and preparation rules.[10] Preparations were performed with diamond burs (229-014XC Torpedo, Romidan, Kiryat-Ono, Israel) using a rotary instrument. 1.5 mm reduction was made from the occlusal surface, and 1 mm reduction was made from all other surfaces. The final abutment had 1 mm 360° chamfer margin, 6° convergence angle, 4 mm height, 8 mm mesial-distal width, and 6 mm buccal-lingual width. Metal duplicates were obtained. For each study group, ten duplicates were prepared, and a total of 50 duplicates were prepared from Co-Cr alloy (Dentorium, New York, USA) via laser sintering.

BruxZir (BruxZir, Glidewell, Newport Beach, CA, USA) was used as MZ crown material. BruxZir include 0.14% Al_2O_3, 9.75% Y_2O_3, 87.81% ZrO_2 and 2.31% HfO_2 by weight.[14] All zirconia crowns were prepared by the same CAD/CAM system (Sirona in the lab, MC X5, Sirona, Bensheim, Germany). Firstly, a die spacer was programmed for a 20 µm cement gap before the manufacture of zirconia crowns. Then, a metal duplicate was scanned. To provide the same standard occlusal morphology, restorations were fabricated as occlusal coping. The cusp and central fossa were not considered (Fig. 1). Zirconia restorations were prepared with 0.5 mm, 0.8 mm, 1.0 mm, and 1.5 mm occlusal thickness, and zirconia restorations were sintered under 1580°C temperature for 2 hours. The glazing was performed with special appropriate spray glaze powder (BruxZir spray powder, Glidewell, CA, USA) under 830°C temperature following the manufacturer's protocol.

PFM crowns were fabricated via the lost-wax casting method. Before the fabrication of PFM crowns, two layers of die spacer (Stumpflack die spacer, S&S Scheftner GmbH, Mainz, Germany) were applied to obtain 20 µm cement gap. Then, the wax patterns of metal infrastructures were prepared to be 0.5 mm in thickness. The thickness of the wax patterns was checked using the dental caliper (Iwanson Decimal Caliper, Asa Dental, Bozzano, Italy) to ensure standardization. The wax patterns were invested and casted with a Ni-Cr alloy (Kera N®, Eisenbacher Dentalwaren GmbH, Woerth, Germany) with a melting degree of 1260°C - 1330°C. Ni-Cr alloy contained nickel 61.4%, chromium 25.9%, molenium 11%, cilisium 1.5%, magnesium 0.02%, cobalt 0.02 %. The thickness of the metal crowns was checked using the dental caliper. The crowns were sanded with 50 µm Al_2O_3 particles. Vita VMK Master porcelain (Vita Zahnfabrik, Bad Säckingen, Germany) was used to fabricate PFM restorations in accordance with the manufacturer's protocols. A thin layer of opaque porcelain (Vita OP3 opaque, Vita Zahnfabrik, Bad Säckingen, Germany) was fired onto the metal surface before veneering porcelain was applied. Afterward, porcelain powder and liquid were mixed, applied, fired, and glazed (Vita Akzent Plus, Vita Zahnfabrik, Bad Säckingen, Germany). Firing procedure of opaque, dentin porcelain, and glaze are presented in Table 1. PFM restorations had 0.5 mm metal thickness and 1.0 mm of porcelain thickness. All PFM restorations were leveled and glazed to have a final occlusal thickness of 1.5 mm, and then, crowns were completed. To provide the same standard occlusal morphology, restorations were fabricated as occlusal coping, similar to the production of MZ crowns. After

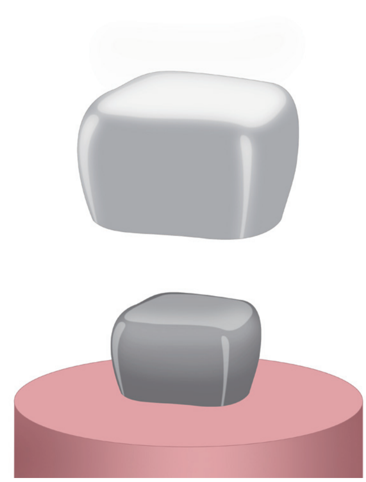

Fig. 1. Occlusal morphology of the crowns.

Table 1. Firing procedure of opaque and dentin porcelain

Porcelain	Firing	Preheating (°C)	Drying (min:s)	Heating Rate (°C/min)	Final Temp. (°C)	Holding (min)
VMK Master	Opaque	500	5:38	80	950	1
	Dentin	500	7:49	55	930	1
	Glaze	500	5:15	80	920	1

the porcelain was fired, the dimensions were measured with a digital caliper.[15] To minimize the measurement error, the porcelain thickness was checked by measuring from 5 different points as buccal, lingual, mesial, distal, and middle site of the occlusal surface. The same technician prepared all restorations in the study.

Marginal fit was evaluated before the cementation of the crowns. The vertical marginal gap evaluation was performed via a silicon replica method.[16] First, a light body was injected into the internal surface of the crowns and then placed on the prepared teeth with finger pressure for five minutes.[16] After the polymerization of the light body, silicone crowns were removed, and light body silicone was stabilized with an embedment of heavy body silicone. The replicas were cut with a surgical blade (#11) first in buccolingual and then mesiodistal directions. To obtain equal surfaces and perpendicular cuts to the surfaces, a careful approach was taken. Any distorted or ruptured replicas were repeated. Afterward, the marginal gap was measured. The evaluation was performed using a stereomicroscope under 10× magnification (ZEISS Stemi 2000-C Stereo Microscope and Axiocam ERc 5s, Göttingen, Germany). The sections were placed parallel to the microscope, and measurements of each crown's light body silicone layer were obtained at the reference points (Fig. 2). Measurements were taken from 4 points of each crown, namely mesial, distal, buccal, and lingual, and from where the metal die and crown margin is the closest in the cervical region (Fig. 2).[16] Each measurement was repeated 3 times, and a mean value of these measurements for each crown has been recorded. All crowns, 10 crowns in each group, were evaluated in terms of marginal gap evaluation. A total of 600 measurements were made for 50 samples. All measurements were carried out by the same experienced operator.

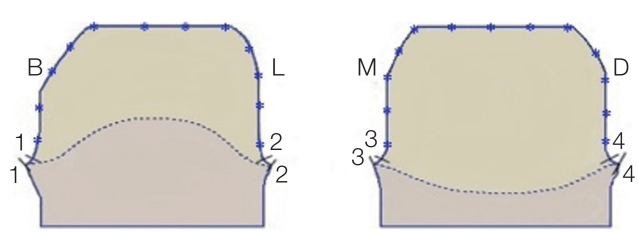

Fig. 2. Stars show the cutoff points for marginal gap measurement. Numbers show reference points for marginal gap measurement. B: Buccal, L: Lingual, M: Mesial, D: Distal.

After the marginal fit of crowns was evaluated, crowns were cemented. A dual-cured resin cement (Panavia F 2.0, Kuraray Noritake Dental Inc., Okayama, Japan) was used to cement of Bruxzir MZ crowns in line with the manufacturer's recommendations. In order to ensure the standardization of the study, the PFM crowns were cemented with the same resin cement. Internal surface of the crowns was sanded with 50 µm Al_2O_3 particles under 3 bar pressure at a distance of 20 mm for 10 seconds. All crowns were cleaned, dried, and primed before cementation. Alloy primer was applied into metal-supported crowns and metal duplicates and dried for 30 seconds. Panavia F Paste A and Paste B were mixed in a 1 : 1 ratio for 20 sec. The mixed resin cement was applied to the internal surfaces of the crowns, and the crowns were placed in metal duplicates. Then excess cement was removed, and the light was applied for 20 seconds. Oxyguard II was applied and left for 3 minutes. Oxyguard II on crowns was removed with a cotton roll and water spray. Overflowing cement residue was removed. The process was performed by the same operator, and finger pressure was applied during the cementing process.

After the cementation procedure, the crowns were aged with thermal cycling (TC) and chewing simulator (CS). Thermal aging was performed with 10000 cycles (with 664 cycles a day for 15 days). The bath temperature was 5°C - 55°C, with 60/10 seconds of bath/dwell cycles. The distilled water levels were monitored everyday, and any decrease was compensated without affecting the temperature of the water bath.

After the thermocycling procedure, the specimens were aged in a dual-axis chewing simulator (CS 4.2, SD Mechatronic GmbH, Feldkirchen-Westerham, Germany). As the antagonist abrader, a stainless steel metal stylus (diameter of 2.36 mm at a height of 0.6 mm from the tip) was used. The metal stylus was stabilized in the CS with a light-cured composite resin (GC Pattern Resin, GC Corp., Tokyo, Japan). In the CS device, the force was transferred to the middle site of the occlusal surface by opposing metal stylus. The chewing simulation parameters used are summarized in Table 2.[17] A one-year simulation as 240000 cycles was used.[18]

The fracture resistance of aging crowns was evaluated. A 10 kN loading device was used for fracture tests (Shimadzu AG-X 10, Kyoto, Japan). All crowns were stabilized on the

Table 2. The configuration of parameters set for dynamic aging[17]

Parameter	Data
Number of cycles	240000
Force	50 N
Height	3 mm
Lateral movement	2 mm
Descendent speed	55 mm/s
Lifting speed	55 mm/s
Feed speed	50 mm/s
Return speed	50 mm/s
Frequency	1 Hz

Table 3. Marginal gaps and fracture resistance of the samples

Parameters/Groups	Marginal gaps (µm)	Fracture resistance (N)
PFM	75.30 ± 10.70 ^{a}	3485.89 ± 195.68 ^{a}
MZ 0.5 mm	52.05 ± 8.67 ^{b}	1421.56 ± 302.91 ^{b}
MZ 0.8 mm	51.40 ± 6.89 ^{b}	2768.1 ± 923.35 ^{c}
MZ 1.0 mm	48.30 ± 10.39 ^{b}	3718.1 ± 624.45 ^{a}
MZ 1.5 mm	44.50 ± 7.89 ^{b}	Over 10000 N*

PFM: Porcelain fused to metal (1.5 mm thickness), MZ: Monolithic zirconia $P <$ .05, different letters indicate statistical significance. * not break with 10000 N (the highest force value of the machine) One way ANOVA and Tukey's HDS post hoc tests were used for multiple comparisons.

device and subjected to the force applied to the central fossa until the point of breakage. All crowns were subjected to a compressive force at a crosshead speed of 1 mm/min with a 4 mm^2 round steel end. Crowns were tightened between metal bars, and a force to break the crown was applied to the middle point of the occlusal surface of the crown. The applied force had 90 degrees in the vertical direction towards the central fossa with a stainless steel round application tip. The force at breakage was recorded as the fracture resistance point.

Statistical analysis was performed by software (IBM SPSS, IBM, Armonk, NY, USA). All data were tested for normality via One-Sample Kolmogorov-Smirnov (K-S) test and then subjected to further analysis for the differences among the groups. According to the One-Sample K-S test, one-way ANOVA and Tukey's HDS post hoc test were used. All data were shown as mean ± SD. $P < .05$ was considered statistically significant.

RESULTS

Marginal gaps and fracture resistance were presented in Table 3. PFM groups exhibited a significantly higher marginal gap than the MZ groups (p=0.005, $P < .05$). All marginal gaps were clinically acceptable; however, the highest marginal gap was detected in the PFM crowns. The marginal gap tended to decrease with the increasing thickness in the MZ crowns.

In terms of fracture resistance, PFM and MZ 1.0 mm had similar resistance (p=0.209, $P > .05$), which was higher than those of the 0.5 mm MZ crowns and 0.8 mm MZ crowns (p=0.025, $P < .05$). 0.5 mm crowns had the lowest fracture resistance compared to the other groups (p=0.01, $P < .05$). 0.8 mm crowns had higher strength than 0.5 mm crowns but lower resistance than PFM and 1.0 mm crowns (p=0.01, $P < .05$). The highest force of the device was 10000 N and could not break MZ 1.5 mm crowns, which had the highest fracture resistance (Table 3).

DISCUSSION

The present study evaluated the fracture resistance and marginal gaps of the MZ restorations with different thicknesses. The findings of this study supported the hypothesis of the study. The results revealed that zirconia crowns exhibited significant fracture resistance even with the thinner crowns, such as those of 0.5 mm, which had a higher fracture load than the biting force (800 - 1000 N).[19] The marginal gaps were higher in the PFM compared to the MZ crowns. Thinner MZ crowns exhibited acceptable fracture resistance.

Aesthetic is a significant factor in choosing the restoration material, and zirconia offers high aesthetic quality along with high mechanical resistance.[20] The major drawback of zirconia-framed porcelain veneered restorations is the porcelain cracking and breakage.[3,8] This flaw was attributed to cohesive failure, which is the result of the low conductivity of zirconia and high firing temperatures in veneered zirconia frames.[2,21] Also, the temperature changes in the manufacturing process of zirconia and veneer ceramics also differ, causing a thermal gradient and thus different contraction patterns and thermal damage.[22] The thickness of the zirconia material might also contribute to this phenomenon since the mechanical and physical features of material are directly related to its thickness.[21,22] Even though the precautions such as reduced anatomical design, homogenous veneering thickness, matched thermal coefficients or slow cooling was introduced to decrease heat damage and cohesive failure, the concept of MZ restorations was raised from these shortcomings of veneered zirconia restorations.[3,8,20]

MZ crowns and metal-ceramic crowns are not comparable in terms of elastic modulus, hardness, corrosion resistance, and cost.[23] Metal-ceramic restorations have been used for forty years and were considered as a gold standard for fixed restorations. For this reason, metal-ceramic crowns were included as a control group (a negative control group with lower physicochemical properties compared to the zirconia crowns) in this study, in which the mechanical properties and marginal compatibility of MZ crowns were investigated.[6,24]

Mechanical properties of MZ restorations are quite good. However, zirconia restorations need to be more translucent in aesthetic regions. For this reason, glazing is performed to enhance the aesthetic properties of zirconia restorations.[25] Considering this fact, all crowns in the present study were glazed to mimic clinical conditions.

Recently, Church *et al.*[5] evaluated the strength of four different zirconia materials and a full-ceramic material with varying thicknesses of 0.5 mm, 1 mm, 1.5 mm, and 2 mm. Their results revealed that all zirconia, regardless of the thickness, exhibited higher strength compared to the equivalent ceramics. Furthermore, MZ displayed better results overall in terms of strength and flexural modulus. Gierthmuehlen *et al.*[26] also demonstrated that fracture loads of ceramics with thicknesses of 1.5 mm and 1 mm were similar, suggesting that 1 mm ceramic thickness was reliable even in the posterior restorations. Dikicier *et al.*[27] demonstrated that the fracture strength of zirconia increased with the thickness (from 0.5 mm to 0.8 mm), and both were not affected by aging. The fracture strength of the zirconia increases with the occlusal thickness; however, even thinner zirconia restorations have adequate fracture strength, reported by Yin *et al.*[28] demonstrating a fracture strength greater than 1500 N and 2500 N in 0.5 mm and 1.00 mm zirconia, respectively. Kim *et al.*[29] also stated that 0.5 mm coping thickness provided fracture strength of 2300 N. Nakamura *et al.*[30] also reported that MZ restorations were applicable even in posterior regions, providing adequate fracture strength; however, they might not be resistant to thermal damage.

Furthermore, Sorrentino *et al.*[31] suggested that occlusal thickness could be reduced to 0.5 mm without affecting fracture strength and failure load, and 0.5 mm MZ could withstand occlusal forces. As for the present results, all crowns exhibited high fracture strength even with the 0.5 mm thickness, which had 1400 N fracture load, which is clinically acceptable.[19] The fracture load increased with the thickness, and 1 mm zirconia had a similar thickness with PFM restoration, which was also reported by previous studies.[26,28,32] However, thicker restorations require greater tooth preparation and tissue loss and might cause pulpal complications.[33] Minimal tooth preparation with lower tissue loss can be achieved with thinner MZ restorations even in the posterior region, which often require reduced occlusal thickness and thinner coping design due to the restricted interocclusal space.[28-30,32]

The internal and marginal gap is another factor decreasing the prolonged use of prosthetic restorations and was recently reported to influence the fracture resistance of the zirconia restorations.[7,8] Ahmed *et al.* showed that the fracture strength of the MZ was associated with the marginal discrepancies and that zirconia with 0.8 mm thickness had smaller marginal gaps compared to 1.5 mm and the fracture load was found to be associated with chamfer design and thickness of the zirconia.[7] The marginal fit is a significant factor for the longevity of restorations, and any failure might result in microleakage, caries, pulpal complications, or periodontal diseases. However, there is no consensus regarding the optimal marginal gap.[34] There were quite large differences between the values determined for ideal marginal fit in the previous studies, which ranged from 7.5 µm to 206.3 µm.[35,36] McLean and von Fraunhofer examined the marginal gaps of 1000 fixed prosthetic restorations for 5 years and stated that a marginal gap less than 80 µm was difficult to detect under clinical conditions.[37] Therefore, McLean and von Fraunhofer stated that the marginal gap should be less than 120 µm,[37] while Fransson *et al.* stated that the clinically acceptable marginal gap should be 150 µm.[38] However, many studies stated that a maximum of 120 µm should be left to ensure long-term use of restorations.[37,39,40] The present study evaluated the marginal gaps of the restorations with silicon replica method which was reported to be a reliable technique to observe marginal fit.[39,40] It was clinically acceptable as all marginal gaps were less than 120 µm. The results revealed that the highest marginal gap was found in PFM restorations, which were still below the suggested value (75 µm), and the difference among the groups was significant.

In a study investigating the effect of different impression techniques on the marginal gap, the marginal gap values of zirconia crowns were found between 26.6 µm and 81.4 µm.[41] In a study evaluating the effect of the finishing line configuration on the marginal gap, the marginal gap of Prettau zirconia crowns was measured as 109 µm, and the marginal gap of Zenostar zirconia crowns was measured as 84.7 µm.[42] Kale *et al.*[43] investigated the effect of different cement space on the marginal gap of zirconia crowns. The marginal gap of zirconia crowns with a cement space of 30 µm was measured as 85 µm, the marginal gap of crowns with a cement space of 40 µm was measured as 68 µm, and the marginal gap of crowns with a cement space of 50 µm was measured as 53 µm.[43] Kocaağaoğlu *et al.* evaluated the effect of 3 different digital scanning methods on the marginal gap in their study. The mean marginal gap of crowns produced using Trios-3 scanner was detected as 47.7 µm, and the mean marginal gap of crowns produced using CEREC scanner was measured as 58.7 µm.[44] Although the methods applied in the mentioned studies differed from the current study, the marginal gap values of all MZ crowns in the present study were in compliance with the results of aforementioned studies, being in the clinically acceptable marginal gap range (lower than 120 µm).

Zirconia is a widely used material in dental practice providing superior mechanical and aesthetic advantages, biocompatibility, and low bacterial adhesion over PFM.[45] However, the durability and sturdiness of the material were reported to decrease with constant wetting by exposure of saliva, low-high temperature shifts, and lateral forces.[46] To simulate the oral environment *in vitro* studies, dynamic aging procedures like thermal cycling and chewing stimulations can be used to mimic oral conditions.[17] To test the strength of the materials after the aging process can help reveal the exact behavior of the restorative materials as observed in the actual oral environment. The present study used a dynamic aging procedure with the two-step, firstly thermal aging and secondly mechanical aging, which was reported to be useful tools in oral simulation[17]; however, different aging models ranging from 5 hours to 1200000 cycles were reported.[17,47,48] Thermal aging can also be performed by steam autoclave procedure; nonetheless, thermal cycling was reported to mimic oral conditions better with low thermal damage to zirconia.[17,47] Stawarczyk *et al.*[49] found that the fracture strengths of various MZ ceramics after thermal aging with autoclave to be between 616 - 928 MPa. Munoz *et al.*[50] also reported that the flexural strength of various MZ varied between 721 - 1187 MPa after thermal aging with an autoclave. Considering the differences in crown thicknesses, thermal aging methods, and thermal cycling procedures used in other studies, the lowest fracture resistance was found to be 1421 N in crowns with the smallest crown thickness in the present study. This value is a clinically acceptable value above those specified in the literature.[19]

The aging procedure used in the present study, 10000 TC and 240000 CS corresponding to 1-year use, can be considered relatively a short duration compared to the previous studies reporting 5-year simulations of use;[17] however, Yang *et al.*[48] revealed that heat damage and related alterations in the zirconia was observed within six months of clinical use of zirconia. Relatively shorter durations such as six months and one year were shown to mimic oral conditions.[48]

However, the present results should be considered with certain limitations. The most significant limitation of the study was that the same depth of preparation was used for all experimental groups to evaluate restorations of 0.5 mm, 0.8 mm, 1.0 mm, or 1.5 mm thickness. Second is the lack of a zirconia group with a thickness lower than 0.5 mm. Thirdly, the present study was designed as an *in vitro* study involving thermal and mechanical aging. Clinical studies would give further information regarding the long-term use of zirconia restorations with reduced thickness. Fourthly, the study design included one type of zirconia with different occlusal thickness. And, the fifth is the lack of standardization of the luting procedure, which was performed with a finger pressure in the current study. Considering the effect of material type, fabrication method, and crown thickness on the mechanical strength of the material, different materials with different study conditions involving opposite tooth wear should also be evaluated. Lastly, further analysis, such as SEM imaging and mechanical testing other than fracture load, should be performed.

CONCLUSION

The present study evaluated the fracture resistance and marginal gaps of the MZ restorations with regular and reduced occlusal thickness. Within the limitations, the results showed that all groups had acceptable fracture load and marginal gaps and thus had clinically applicability. However, it was found that the fracture resistance increased with increasing ceramic thickness. MZ provided acceptable fracture resistance even with the lowest thickness as 0.5 mm. MZ with 1.00 mm thickness provided equal fracture resistance to PFM with 1.5 mm thickness. As for the marginal gaps in the zirconia groups, the values were similar among each other and lower than the PFM group. MZ crowns with a thickness of 0.5 mm showed a fracture resistance above the reasonable chewing force. Therefore, in later studies, the fracture resistance of MZ crowns with a thickness of less than 0.5 mm can be evaluated.

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