Effects of coping designs on fracture modes in zirconia crowns: Progressive load test

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Abstract

This study was designed to evaluate the effects of different coping designs on the fracture modes of posterior zirconia crowns. One hundred mandibular molar copings of ten different designs were fabricated from 3Y-TZP. All ten different groups (n = 10) had 1 mm wide shoulder, but they had varying heights of 1 mm, 2 mm, or 3 mm on the buccal, lingual and proximal sides. After being sintered and veneered, the zirconia crowns on titanium abutments were stored in distilled water at 37 °C for 24 h. Then they were loaded into a universal testing machine. The veneer fracture load and bulk fracture load values were measured. Moreover, the fracture surfaces were examined with a scanning electron microscope. One-way analysis of variance and the Scheffé posthoc test were carried out for statistical analyses (α = .05). Higher shoulders on the buccal, lingual and proximal sides resulted in higher fracture load values. As the heights of the shoulders increased, the occurrence of chipping decreased. Additionally, the bulk fractures on the buccal surfaces increased in the crowns with higher shoulders. We concluded that the shoulder coping design plays a critical role in the survival of posterior zirconia restorations by reducing veneering porcelain fractures.

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1. Introduction

Currently, porcelain-fused-to-metal crowns are generally used for esthetic replacements of posterior teeth. However, porcelain-fused-to-metal crowns have some drawbacks, such as dark shadows at margins, allergic reactions and a lack of biocompatibility [1,2]. As dentistry has evolved, the demand for metal-free crowns has arisen. This has led to the development of various all-ceramic restorations that are more similar to natural teeth [3]. All-ceramic restorations are widely used in dentistry because of their outstanding esthetics and biocompatibility. It has been reported that the estimated 5-year survival rate of all-ceramic crowns range between 90.7% and 96.6% (feldspathic/silica-based ceramics, 90.7%; leucite or lithium disilicate reinforced glass ceramics, 96.6%; glass-infiltrated alumina, 94.6%; densely sintered alumina, 96.0%; and zirconia-based crowns, 91.2%) [4]. All-ceramic single crowns suffer from technical complications, such as ceramic chipping, ceramic fracture, framework fracture, loss of retention, and marginal discoloration [4]. Veneering porcelain chipping is a common problem and has been occurring more often with zirconia-based crowns than all other ceramic crowns [4].

Porcelain chipping and zirconia restoration fractures are critical issues in restorative dentistry. Most posterior zirconia restoration failures were reported either as minor or major chipping, with the chipping rate as high as 25% [5-14]. Fracture or chipping of veneering porcelains can be either a fracture of the porcelain itself or a fracture originating from the interfaces between the coping and porcelain [15]. The following classification of the ceramic fractures has been suggested: type 1, crack of the veneering ceramic; type 2, chipping restricted to the veneering ceramic; type 3, chipping exposing the core; and type 4, fracture of the core [16]. Moreover, these classifications can be further subdivided into "non-critical" (i.e., can be repaired) or "critical" (i.e., must be replaced) [16]. The survival estimates for
“critical fracture” and “all fracture” in zirconia single crowns were 89.4% and 80.9%, respectively. In zirconia fixed dental prostheses (FDPs), the survival estimates were 68.6% for critical fracture and 24.6% for any fracture [16]. Porcelain cracks have been ascribed to tensile stresses arising from internal or external flaws. In a fractographic analysis of failed clinical zirconia crowns, heavy occlusal wear spots were observed at the failure origins [16]. Other possible causes for crack initiation and propagation are a mismatch of the coefficient of thermal expansion (CTE) between zirconia coping and veneering porcelain and residual stresses in veneering porcelain during the cooling process [17–20]. The CTE of veneering porcelain must be slightly lower than that of zirconia coping to place veneering porcelain under compression. Special attention should be given to match the CTE between zirconia coping and veneering porcelain under compression. More accurately higher reliability than the standard (0.5 mm uniform thickness) coping designs for molars. Several studies reported that modified coping designs (proximal and lingual supportive shoulder designs) enhanced the reliability of zirconia crowns [29–32]. However, the effects of various shoulder designs (height and position) on the fracture resistance has not been fully determined. There is little information available on zirconia coping designs [26,28,32,33]. Moreover, there is no universally accepted ideal coping design for posterior zirconia restorations. Tinschert et al. [34] demonstrated that zirconia-based FDPs showed a sufficient success rate in clinical use. However, they emphasized the importance of a sufficient veneering porcelain thickness with a range of 1–2 mm. Molin et al. [9] suggested that anatomically designed 3-unit FDPs were promising prosthetic alternatives. Broseghini et al. [35] reported the use of a functional area protection concept in a framework design to support veneering porcelain and showed satisfactory outcomes in terms of preventing veneer chipping for short periods.

The present study investigated the effects of coping designs on the fracture behaviors of veneering porcelains in mandibular zirconia molar crowns. The null hypothesis was that there is no difference in the fracture resistance of posterior mandibular zirconia crowns with various coping designs.

2. Materials and methods

2.1. Preparation of the specimens

A mandibular right first molar of the Nissin study model (D85DP-500B.1, Nissin Dental, Kyoto, Japan) was duplicated using a polyvinyl siloxane (Express™ light body, 3M ESPE, MN, USA) with a custom metal impression tray. A stone model was fabricated. The stone model was prepared (chamfer with a depth of 1.2 mm and 8° angle of convergence) using a carbide bur (Komet H 356 RGE 103.031, Gebr. Brasseler GmbH, Lemgo, Germany). To achieve an even thickness in the preparations, a putty index (Express™ STD Putty, 3M ESPE) was made prior to preparation. Moreover, the carbide bur was attached to the surveyor (Fl, DeguDent GmbH, Kanau, Germany) to ensure the standardization of the preparations. The prepared stone model was duplicated, and 100 identical titanium abutment teeth were fabricated using a CAD/CAM system (Myplant™, RaphaBio, Seoul, Korea). To keep the prepared abutment shape and the internal shape of the crown consistently identical, we needed 100 identical abutment teeth. It was prohibitive to attain 100 natural teeth with similar properties, shapes, and sizes. Titanium was chosen because it has superior fracture resistance than that of the zirconia coping. We determined that titanium block milling would be more accurate way than metal castings or resin packing to make consistently identical models.

The coping design of each group is presented in Fig. 1.

Group 1 (Control group): No shoulder (Fig. 1a).

Group 2: Unveneered 1-mm-high shoulder on the proximal/lingual side of the crown (Fig. 1b).

Group 3: Unveneered 1-mm-high shoulders on the proximal/lingual and the buccal side of the crown (Fig. 1c).
Group 4: Unveneered 2-mm-high shoulder on the proximal/lingual side of the crown (Fig. 1d).

Group 5: Unveneered 2-mm-high shoulder on the proximal/lingual side of the crown and a 1-mm-high shoulder on the buccal side of the crown (Fig. 1e).

Group 6: Unveneered 2-mm-high shoulders on the proximal/lingual and the buccal side of the crown (Fig. 1f).

Group 7: Unveneered 3-mm-high shoulder on the proximal/lingual side of the crown (Fig. 1g).

Group 8: Unveneered 3-mm-high shoulder on the proximal/lingual side of the crown and a 1-mm-high shoulder on the buccal side of the crown (Fig. 1h).

Group 9: Unveneered 3-mm-high shoulder on the proximal/lingual side of the crown and a 2-mm-high shoulder on the buccal side of the crown (Fig. 1i).

Group 10: Unveneered 3-mm-high shoulders on the proximal/lingual and the buccal side of the crown (Fig. 1j).

Except for Group 1, the copings in all other groups were reinforced with a shoulder (1 mm wide). The 3 mol% yttrium-stabilized presintered zirconia blocks (IPS e.max ZirCAD, Ivoclar Vivadent, Schaan, Liechtenstein; Table 1) were randomly divided into 10 groups (n = 10). The copings were manufactured using a Cerec scanning and milling machine (Cerec AC and Cerec inLab MC XL, Sirona, Bensheim, Germany) after scanning the titanium abutment tooth. The thickness of the coping was 0.5 mm and that of the cement space was 50 μm.

After the milling procedure, the copings were sintered at 1500 °C in a furnace (Cerec inFire Speed, Sirona). The copings were then examined on the titanium abutment teeth. The exclusion criteria were the following: visually unacceptable margin, under-contoured core, and evidence of rotation under finger pressure on the titanium abutment tooth. New copings were fabricated to replace the excluded specimens. One hundred copings were then adjusted by a professional dental technician under a stereo microscope (Stemi DV 4, Zeiss, Barrington, NJ, USA) with a magnification of 8 times until the best possible fit was achieved.

A layer of the ceramic liner (IPS e.max ZirLiner, Ivoclar Vivadent) was coated on the copings and fired at 960 °C. An unprepared tooth model was used for the external shape of the crowns. Wax-up of veneer porcelain on the copings was performed using a prefabricated silicone index to ensure an identical thickness of 1 mm. The wax (Nawax compact, Yeti Dental Products, Engen, Germany) surfaces were smoothed and finished. The wax-coping complexes were buried under the investing material (IPS PressVEST, Ivoclar Vivadent). A muffle-furnace was heated, and the waxes were then burnt out. The copings were overpressed by a fluorapatite glass ceramic (IPS e.max ZirPress, Ivoclar Vivadent; Table 1), which had an adequate CTE with respect to the zirconia copings. After slow cooling (overpressed copings were kept in the closed furnace until the temperature of furnace reached 450 °C at the cooling rate of 10 °C/min, then the furnace was opened), the investing materials were removed by air abrasion (50-μm glass beads at 2 bar pressure). The reaction layers formed in the crowns were removed by immersing the crowns into an ultrasonic cleaner containing HF solution (IPS e.max Press Invex Liquid, Lot 35611)
H31070, Ivoclar Vivadent) for 5 min. After the crowns were cleaned under tap water for 3 min, the extrusion flashes and sprues were removed with a water-cooled air-turbine without pressure. Finally, glaze paste (IPS e.max Ceram Glaze paste, Ivoclar Vivadent) was mixed with a special liquid (IPS e.max Glaze and Stain Liquid, Ivoclar Vivadent) to the appropriate consistency, applied uniformly on the crowns and fired at 750 °C. The internal surfaces of the crowns were cleaned with steam and degreased with 80% ethanol. Then, the surfaces of the titanium abutment teeth were air abraded (50-μm aluminum oxide at 0.5 bar pressure) and degreased before cementing. Each crown of all of the groups was then luted onto its corresponding titanium abutment tooth using self-adhesive resin cement (RelyX™ Unicem™, 3M ESPE). The resin cement was mixed according to the manufacturer's instructions. The crowns were filled with self-adhesive resin cement. The cement was spaced out by a disposable brush until the internal surfaces were coated. The crowns were then seated onto the titanium abutment teeth and held in place with finger pressure in accordance with the manufacturer's instructions. The excess cement was removed. An experienced dentist seated the crowns onto the titanium abutment teeth, while a dental assistant mixed the cement. The final test specimens received identification numbers and were stored in distilled water at 37 ± 1 °C for 24 h until they were loaded for the fracture test.

2.2. Single load to fracture test

From the storage container, all of the specimens were set directly onto a universal testing machine (Instron 5583, Instron, Canton, MA, USA). The specimens were adjusted to ensure that there were three-point occlusal contacts with the 7-mm diameter stainless steel ball. The stainless steel ball was positioned vertically at the central fossa of the occlusal surface, making contacts with the mesiobuccal, distobuccal, and mesiolingual cusps (Fig. 2). To avoid a high-stress concentration on the occlusal surface, a 1-mm piece of ethylene-vinyl acetate foil was placed between the occlusal surface and the stainless steel ball. The fracture load values were then recorded by loading the ball until ceramic fracture occurred in the universal testing machine. Axial compression was applied with a crosshead speed of 0.5 mm/min until ceramic fracture occurred. Fracture was defined as the occurrence of visible cracks in combination with load drops and acoustic events or by chipping, which made the crown clinically unusable. After mechanical loading, the fracture sites of the zirconia crowns were evaluated. All of the tests were carried out at 23 ± 1 °C.

The crowns were evaluated in terms of the following aspects:

1. Fracture load [veneer fracture load, bulk fracture load].
2. Fracture site [buccal area, lingual area or others].
3. Fracture mode [adhesive or cohesive].
   a. Cohesive failure within veneering porcelain [chipping].
   b. Delamination [coping exposure].
   c. Catastrophic failure [bulk fracture].

2.3. Fracture surface examination

The fracture surfaces of selected crowns were examined with a scanning electron microscope (SEM). Each specimen was gold-coated with a sputter coater (SC7620 Mini Sputter Coater, Poloron, Schwalbach, Germany) and then mounted onto the coded brass stubs and examined using SEM (FE-SEM, S-4700, Hitachi, Tokyo, Japan) at × 25, × 35, × 50, × 100, × 200, × 1000, and × 10,000 magnification.

2.4. Statistical analysis

Calculations and statistical analyses were performed using statistical software (SPSS 20.0, SPSS Inc., Chicago, IL, USA).
Table 2
Mean values and standard deviations in parenthesis of initial fracture load (N) and bulk fracture load (N) of zirconia crowns.

<table>
<thead>
<tr>
<th>Group</th>
<th>Initial fracture load (N)</th>
<th>Fractographic analysis</th>
<th>Bulk fracture load (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>4277 (1676)ab</td>
<td>Veneer fracture</td>
<td>4978 (1271)c</td>
</tr>
<tr>
<td>2</td>
<td>5186 (1754)abc</td>
<td>Veneer fracture</td>
<td>5638 (1906)cde</td>
</tr>
<tr>
<td>3</td>
<td>5260 (1634)abc</td>
<td>Veneer fracture</td>
<td>6532 (1793)cdefg</td>
</tr>
<tr>
<td>4</td>
<td>5962 (1656)abc</td>
<td>Veneer fracture</td>
<td>6467 (1404)cde</td>
</tr>
<tr>
<td>5</td>
<td>6130 (1649)abc</td>
<td>Veneer fracture</td>
<td>7672 (1856)cde</td>
</tr>
<tr>
<td>6</td>
<td>6259 (1691)abc</td>
<td>Veneer fracture</td>
<td>7680 (1449)cde</td>
</tr>
<tr>
<td>7</td>
<td>6452 (1352)abc</td>
<td>Veneer fracture</td>
<td>7852 (1574)cde</td>
</tr>
<tr>
<td>8</td>
<td>6719 (1487)abc</td>
<td>Veneer fracture</td>
<td>8451 (2023)cde</td>
</tr>
<tr>
<td>9</td>
<td>6668 (1697)abc</td>
<td>Veneer fracture</td>
<td>7744 (1486)cde</td>
</tr>
<tr>
<td>10</td>
<td>7933 (1588)abc</td>
<td>Veneer fracture</td>
<td>9104 (909)c</td>
</tr>
</tbody>
</table>

Initial fracture load and bulk fracture load data are analyzed separately. The same superscript lowercase letters in each fracture load indicate no significant differences respectively (Scheffé test: P > .05).

The veneer fracture loads and bulk fracture loads were registered, and the differences between the groups were calculated using one-way analysis of variance (ANOVA) after tests of normality (Shapiro-Wilk, P > 0.05). Additionally, a multiple comparison the Scheffé posthoc test was carried out to evaluate the differences between the experimental groups. The test was performed at a significance level of 0.05.

3. Results

The mean veneer fracture loads and mean bulk fracture loads with standard deviations of the tested zirconia crowns are listed in Table 2. In Group 1, the veneer fracture loads ranged from 2224 to 7373 N, with a mean value of 4277 ± 1676 N. For the crowns in Group 10 (7993 ± 1588 N), the mean veneer fracture loads were significantly higher (P < .001) than those in Group 1 (Tables 2 and 3). The mean veneer fracture load was 87% higher in Group 10 than in Group 1 (7993 ± 1588 N vs 4277 ± 1676 N; P < .001, ANOVA).

The mean bulk fracture load was 70% higher in Group 8 than in Group 1 (8451 ± 2023 vs 4978 ± 1271 N; P < .001, ANOVA). Moreover, the mean bulk fracture load in Group 10 was 83% higher than in Group 1 (9104 ± 909 vs 4978 ± 1271 N; P < .001, ANOVA). In addition, the mean bulk fracture load in Group 10 was significantly higher (P < .001) than that of Group 2 (5638 ± 1906 N; Tables 2 and 4).

Fig. 3 shows the mean veneer fracture loads and mean bulk fracture loads of each group. As the height of the shoulder increased, the fracture load increased as well.

All of the specimens eventually fractured during the load test (Fig. 4). They showed chipping, delamination, radial cracks or bulk fractures. Some specimens suffered small fractures of the lingual or buccal cusps reaching the proximal aspect, while the others had fractures across the buccal or lingual aspect in the mesiodistal direction. The fracture modes of the experimental groups are presented in Fig. 5. As the height of the shoulder increased, chipping decreased and bulk fractures at the buccal surface increased.

4. Discussion

The aim of the present study was to investigate the effects of coping designs on the fracture behavior of veneering porcelain for posterior mandibular zirconia crowns. The crowns that had approximate 3-mm shoulders showed the highest fracture loads. As revealed by one-way ANOVA on the testing results of each coping design, the fracture loads of posterior zirconia crowns were differed in various coping designs. Therefore, the null hypothesis that there is no difference in the fracture resistance of posterior mandibular zirconia crowns with various coping designs was rejected.

Conventional metal crowns are appropriate long-term replacement of posterior teeth because of their outstanding mechanical properties. However, the demand for metal-free restorations has led to the development of all-ceramic materials. Because of a lack of translucency, porcelains are veneered on polycrystal-line ceramics, such as zirconia or alumina. However, this introduces a point of vulnerability to the structure. The veneering porcelain on the zirconia crown chips or fractures because of the brittleness or inherent residual stress. Long-term clinical trials have shown that this is a relevant issue among treatment...
failure factors [5–14]. If zirconia crowns are to be used as a treatment option to replace molars, they must withstand posterior mastication forces of approximately 700 N [37,38]. The results of the current study show that the veneer fracture loads of all groups exceeded 700 N. However, it would be difficult to draw a conclusion by comparing the vertical loads on 3-point occlusal contact to the actual, multi-directional, dynamic functional loading, which occurs in the mouth.

There have been several attempts to prevent veneering porcelain fractures in terms of coping designs. The results of the present study correspond well with those of the earlier studies [28–31]. Silva et al. [28] examined the improvement of zirconia crown reliability by modifying the coping designs. They showed that modified coping designs had significantly higher reliabilities than standard coping designs [28]. Their coping had a 2.5-mm-high shoulder on the lingual wall extending to the proximal. In their study, a step-stress fatigue load was applied at one point of the buccal incline of the mesiolingual cusp to simulate laterotrusive movement of the mandible. Bonfante et al. [29] investigated the fatigue reliability of yttria tetragonal zirconia crowns. They reported that copings with alternative designs exhibited significantly higher reliabilities [29]. They designed one arbitrary lingual shoulder to reduce the veneering porcelain bulk of the cusps. In their study, a load was applied at one point of the occlusal surfaces. These studies mentioned above documented the fatigue and reliability of zirconia crowns. However, they did not provide information about fracture resistance with variations in different shoulder designs under a maximum intercuspal position. Therefore, in the current study, variations in the shoulder

Fig. 4. Representative images of the fractured zirconia crowns. (a) Chipping, (b) delamination, and (c) cracks that extended to the coping (radial cracks); (d) catastrophic failure (bulk fracture).

Fig. 5. Fracture mode distribution of the experimental and control groups.
positions (buccal, lingual/proximal) and heights (0 mm, 1 mm, 2 mm, and 3 mm) were designed in the copings. The design concept was empirically derived from porcelain-fused-to-metal restorations in which the amount of support is not understood well [39]. Loading points play important roles in stress distribution and concentration in complex geometries, such as FDPs. Magne et al. showed that peak stresses were observed in the central groove of the posterior maxillary dentitions during mediotrusive contact and on the lingual surface of the mandibular posterior teeth during laterotrusive movement [40]. Therefore, fracture modes were evaluated under 3-point occlusal contact in this study. We manufactured veneering porcelain via the heat pressing method to prevent undesired variations in shapes and sizes of the models that occur when they are constructed by hand. The coping adjustments may produce flaws, which could influence the fracture loading test. To minimize this problem, we adjusted coping as little as possible. In addition, we used a bur under water spray when adjustments had to be made. Moreover, the main objective of the present study was not zirconia fractures but veneer fractures. Since zirconia coping fractures did not happen before veneer fractures in all groups, the influence from the coping adjustments was deemed negligible.

Kokubo et al. [32] evaluated the fracture loads of the four different coping designs in zirconia molar crowns. Those coping designs were conventional copings with a 0.6-mm-high shoulder, copings that followed the original cuspal configurations, and copings with supporting configurations against occlusal force [32]. They demonstrated that the copings that follow the original cuspal configuration showed the highest mean fracture load value under both vertical and lateral loads. Guess et al. [19] examined the fatigue reliability and failure modes of the two different zirconia coping designs (conventional and anatomically designed) veneered with either a hand-layer or a pressed technique. They showed that the hand-layer veneered conventional coping designs had better reliabilities than the counterpart press-veneered coping design [19]. Moreover, the anatomically designed coping showed significantly increased reliability and led to a decreased veneer porcelain chip size. The restorations differ from each other for obvious reasons, such as different patient factors,
varying space under the restorations, varying tissue damage, and different material properties; therefore, the in vitro test data are difficult to compare with those of other studies. It is important to exclude sample differences in laboratory experiments and simulation tests.

The crowns that had approximately 3-mm shoulders supported the largest fracture loads in the present study. As the shoulder height increased, so did the values of the veneer fracture load. The existence and height of the lingual shoulder had larger influences on the veneer fracture load than that of the buccal shoulder. Comparing the veneer fracture load difference between Group 2 and Group 1 to that between Group 2 and Group 3 shows that the increase in the veneer fracture load gained by adding 1 mm to the shoulder height on the lingual side is greater than the increase gained by adding the same shoulder height to the buccal side. Again, compared to the group without any shoulder on either the buccal or lingual sides, the group with the 1-mm lingual shoulder had a larger increase of the fracture load. However, adding 1 mm of buccal shoulder to the coping with a pre-existing lingual shoulder did not significantly increase the fracture load. The same tendency appears in all of the other groups. Therefore, we understand that the existence of a lingual shoulder in the posterior mandibular zirconia crown is more important than the existence of a buccal shoulder. This result is clinically important as well. Usually, the exposure of the lingual surface of a mandibular posterior zirconia crown during a smile, speech, and mastication is not much greater than the exposure of the buccal surface. In addition, the lack of translucency of zirconia is an esthetic drawback. Adding a lingual shoulder under veneering porcelain compromises esthetics less while adding remarkable strength to the mandibular posterior zirconia crowns. The shoulder on the coping may allow for an even distribution with adequate thickness of the veneering porcelain, supporting the porcelain and leading to a decrease in the tensile stresses. Additionally, the bulk fracture load values increased with increasing shoulder height. The reason for this increase could be explained by the fact that the width of the shoulder is 1 mm, while the thickness of the coping in general is 0.5 mm. In other words, an increase in the shoulder height also increases the coping thickness. However, it should be noted that the load values from all groups of this study were higher than reported maximum bite force (700 N).

Furthermore, as the height of the shoulder increased, the amount veneer porcelain chipping decreased. The increase in the shoulder height provided more support and resulted in more delamination. The support from underneath the veneering porcelain may reduce the tensile stress on the veneering porcelain while adding a larger compressive load. Moreover, as the shoulder height increased, the veneering porcelain volume decreased substantially. This may decrease the thermal gradients in the veneering porcelain, which, in turn, decreases the residual stresses in the veneering porcelain [20]. The residual stresses in the veneering porcelain may not arise from tempering associated with rapid cooling because the slow cooling was performed after pressing the veneering porcelain. Paula et al. reported that with slow cooling, the reliability of zirconia crowns was the same, regardless of coping design [18]. This may suppose that the coping design is not a single critical factor contributing to the reliability of zirconia crown. In the present study, many crowns with shoulders showed exposure of the veneer–coping interface, with cracks propagating to the zirconia coping. The effects of the shoulder were more dramatic as the shoulder height increased. These effects were more remarkable at the lingual surface because of the shorter clinical crown height of the lingual surface than that of the buccal surface. Moráguez et al. classified fractures of ceramic restorations into 4 categories [16]. Moreover, they subdivided veneer cracking and chippling into 2 categories: non-critical and critical [16]. In the present study, most of the fracture sizes were large in the test specimens because of the nature of single loads for the fracture test. All of the fractured specimens showed "critical fractures" and required a replacement of the crown. The purpose of this test was to see if the shoulder increased the veneer porcelain fracture load. Thus, we used a single load to test for fractures. The single load fracture testing showed that the test produced damage that was not observed clinically because of the high contact stresses [41]. For ball-on-ball contact between the indenter and the cusps in the present study, it is likely that many cracks needed to be produced before the loads required for bulk fractures were reached [41]. The contact stresses reported in many studies were measured to be between 1000 MPa and 5000 MPa, whereas these of intra-oral wear facets were calculated to not exceed 40 MPa irrespective of the bite force [41]. Therefore, the load values in this study could not be actual stresses leading to crown fracture. Moreover, the single load fracture testing showed load values that were far different from those found in fatigue tests that simulated the oral environment.

The direction of the functional loads depends on the relation between the jaw and the occlusal surface of each patient. The results of in vitro testing cannot be extrapolated to the clinical setting, as the design of the present study did not consider factors that exist in the actual oral environment, such as the dynamic forces of mastication or fatigue loading. In addition, in the single loads for fracture testing, a titanium abutment tooth was used instead of a human tooth to support the zirconia crown until fracture occurred. Since titanium has different modulus of elasticity from dentin, the stress distribution on the titanium abutment during loading should have different pattern on human tooth. That limit this study to some extent. It should be noted that this was a comparative study that had the same variables except for the shoulder height. Hence, it should be emphasized that single load fracture testing is only one of many experiments that can be used and that maximum strength is only one property of zirconia crowns. In addition, this experimental study design provided no data on thermal cycling or cyclic loading. Further investigations are necessary to assess the fracture resistance for longer periods of time and to evaluate the effect of thermal cycling on fractures.
5. Conclusion

The null hypothesis, which stated that there is no difference in the fracture resistance of posterior mandibular zirconia crowns with various coping designs, was rejected. The highest veneer fracture load was observed for the crowns with shoulders of approximately 3 mm. The increase in shoulder height provided higher veneer fracture load values. The lingual shoulders of approximately 3 mm. The increase in shoulder height provided higher veneer fracture load values. The lingual shoulder had a greater influence on the veneer fracture load than the buccal shoulder.

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