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Effect of Core Thickness Differences on Post-Fatigue Indentation Fracture Resistance of Veneered Zirconia Crowns

Abdulrahman Alhasanyah, BDS, MSD, Tritala K. Vaidyanathan, MASc, PhD, & Robert J. Flinton, DDS, MS

Department of Restorative Dentistry, NJ Dental School, University of Medicine and Dentistry of NJ, Newark, NJ

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Keywords
Indentation loading; post-fatigue fracture; core/veneer thickness ratio; failure modes; zirconia crowns.

Correspondence
Tritala K. Vaidyanathan, Department of Restorative Dentistry, NJ Dental School, University of Medicine and Dentistry of NJ, 110 Bergen St., Newark, NJ 07103. E-mail: vaidyatk@umdnj.edu

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Abstract

Purpose: Despite the excellent esthetics of veneered zirconia crowns, the incidence of chipping and fracture of veneer porcelain on zirconia crowns has been recognized to be higher than in metal ceramic crowns. The objective of this investigation was to study the effect of selected variations in core thickness on the post-fatigue fracture resistance of veneer porcelain on zirconia crowns.
Materials and Methods: Zirconia crowns for veneering were prepared with three thickness designs of (a) uniform 0.6-mm thick core (group A), (b) extra-thick 1.7 mm occlusal core support (group B), and (c) uniform 1.2-mm thick core (group C). The copings were virtually designed and milled by the CAD/CAM technique. Metal ceramic copings (group D) with the same design as in group C were used as controls. A sample size of $N = 20$ was used for each group. The copings were veneered with compatible porcelain and fatigue tested under a sinusoidal loading regimen. Loading was done with a 200 N maximum force amplitude under Hertzian axial loading conditions at the center of the crowns using a spherical tungsten carbide indenter. After 100,000 fatigue cycles, the crowns were axially loaded to fracture and maximum load levels before fracture was recorded. One-way ANOVA ($P < 0.05$) and post hoc Tukey tests ($\alpha = 0.05$) were used to determine significant differences between means.
Results: The mean fracture failure load of group B was not significantly different from that of control group D. In contrast, the mean failure loads of groups A and C were significantly lower than that of control group D. Failure patterns also indicated distinct differences in failure mode distributions. The results suggest that proper occlusal core support improves veneer chipping fracture resistance in zirconia crowns.
Conclusions: Extra-thick occlusal core support for porcelain veneer may significantly reduce the veneer chipping and fracture of zirconia crowns. This is suggested as an important consideration in the design of copings for zirconia crowns.

Metal ceramic restorations are popular in prosthodontics because of their long-term durability with a survival rate of 95.5% reported in a 7-year study.[1] However, the use of a metal backing for porcelain makes it necessary to provide axial tooth reduction of no less than 1.5 mm to allow space for the opaque porcelain needed to mask the metallic gray color.[2] Today, ceramic restorations can overcome this limitation.[3,4] They are stronger, and no opaque porcelain is needed. Conservative tooth preparation involving minimum tooth reduction is possible.[5] Elimination of opaque porcelain enhances spatial availability for veneering porcelain, making it easier to achieve optimum esthetic results.[2,6] Biocompatibility is also improved by reducing corrosion, hypersensitivity, and allergy complications involving some metals.[2,7] The elimination of metal also removes esthetic concerns at the margin and allows placement of the margin at the gingival level or at a slightly subgingival level. This may help prevent any iatrogenic periodontal trauma during soft tissue manipulation in making the final impression. It also results in a more accurate impression and minimizes the risk of biological width violation.[6] Natural soft tissue color around restored teeth and implants is also better maintained.[8,9] Thus, the ceramic restoration is now a popular alternative to the metal ceramic restoration.[10]

Unfortunately, concern about mechanical properties of conventional ceramics have limited their use to single-unit crowns and 3-unit anterior fixed partial dentures (FPDs).[11-13] The survival rate of conventional ceramic posterior restorations is known to be lower than that of ceramic anterior restorations.[13]

Zirconia in the form of yttrium-stabilized tetragonal zirconia polycrystals (Y-TZP) is now available as an alternative coping material with improved mechanical properties. Y-TZP coping with porcelain veneer also has esthetics superior to its metal ceramic counterpart.[15] Typical mechanical properties of Y-TZP are a flexural strength of 900 to 1200 MPa, compressive strength of 2000 MPa, Young's modulus of 210 GPa, and hardness of 1200 HV.[14,15] Previous reports in the literature indicate that typical intraoral forces during mastication and clenching may range from 5 to 376 N and 216 to 880 N, respectively.[16-20] This makes Y-TZP a very promising alternative coping material to metal for posterior crown and FPD applications.

Unfortunately, the veneering porcelain on zirconia coping is susceptible to more chipping attrition,[21-23] because the zirconia coping/veneer interface is weaker than the metal/porcelain interface.[24,25] The chipping of veneering porcelain on zirconia (13.6%) has been reported to be four times that of metal ceramic FPDs (2.9%) over 5 years.[12] In addition, Walton reported only 3% of mechanical failures in metal ceramic crowns in a 10-year longitudinal study.[26]

Thermal expansion mismatch between coping material and porcelain, improper firing and cooling, or improper coping build-up are leading causes of chipping in porcelain veneer.[27-31] Feldspathic porcelains with thermal expansion compatibility with zirconia are now available.[5,23,27] The effect of cooling rates on post-firing internal stresses and shear bond strength have been overcome with better laboratory procedures.[28,29] However, feldspathic porcelain has a low flexural strength (55–87 MPa),[7] and proper core support and cementation procedures are important to ensure long-term durability of the restoration.

Several investigators have previously studied fracture resistance of veneered zirconia crowns under a variety of conditions. These studies included scanning electron microscopy (SEM) of fracture surfaces under compression,[32] edge chipping tendencies,[33,34] differences in veneer thickness on moisture retention and diffusion into the core,[35] effects of shoulder and chamfer preparation of margins,[36] fabrication techniques and aging,[37] supporting die structures,[38] different cementation techniques,[39,40] differences in finish lines,[41,42] differences in veneering methods and materials,[43,44] effect of connector design,[45] and effect of off-axis loading.[46] Other authors have also shown that core/veneer thickness ratios have a significant effect on the hardness, fracture toughness, and residual stress in zirconia crowns.[47,48]

In this research, we hypothesized that thicker core thickness at the load-bearing site may help improve postfatigue fracture resistance of zirconia crowns. The main objective of this study was to investigate whether increased core thickness designs of copings can improve the post-fatigue fracture resistance of veneered zirconia crowns. The null hypothesis tested was that post-fatigue fracture failure loads in zirconia crowns using thicker core supports are not significantly different ($p > 0.05$) from those of traditional metal ceramic crowns.

Materials and methods

Sample size selection using power analysis

This study focused on the post-fatigue fracture resistance of zirconia restorations due to thickness variations in the Y-TZP core and a compatible feldspathic porcelain veneer. Four coping preparation designs in a one-way layout were tested in the experiments. The sample size was determined using power analysis with an assumed effect size of 0.50. This effect size was estimated as the ratio of the largest difference between experimentally observed means to the (SD) in a previous study.[39] The total number of specimens estimated for 90% power at $\alpha = 0.05$ was 60. We used a higher total sample size of 80, corresponding to approximately 96% power.

Fabrication of replicate tooth models

An ivorine mandibular molar (tooth #19) (Columbia Dentoform, New York, NY) mounted in an orthodontic resin (Caulk Orthodontic Resin, Dentsply Caulk, Milford, DE) block and condensation silicon putty index (Sil-Tech®, Ivoclar Vivadent, Amherst, NY) was made before tooth preparation, to be used to attain the full-contoured wax-up. This was scanned, and this index was used as a guide for the ceramist during porcelain application to standardize the porcelain buildup.

After the index was made, teeth were prepared according to clinical guidelines using a diamond bur. Poly(vinyl siloxane) light- and heavy-body impression material (Aquasil Ultra Smart Wetting, Dentsply Caulk) was used to make an impression of the prepared tooth, and then composite resin material (Z100, 3M ESPE, St. Paul, MN) was applied in layers and light cured according to manufacturer's instructions to fabricate a model of the prepared tooth. Composite models of each group were stored in normal saline for 30 days as recommended by Coelho et al.[21] Figure 1 shows the tooth before (A) and after (B) and the index placed on the prepared tooth, showing the available space for the crown (C).

Fabrication of experimental and control crowns with different core/veneer thickness designs

The prepared ivorine tooth and the full-contoured wax-up were scanned in the Procera optical scanner (Nobel Biocare, Mahwah, NJ), and zirconia coping core structures were prepared. Virtual designs of the copings were made as follows:

Group A = lowest uniform core thickness (0.6 mm)
Group B = Extra-thick occlusal core support (1.7 mm), 1.2 mm non-occlusal core thickness
Group C = 1.2 mm uniform core thickness (thickness of the core was 0.6 mm higher than that in group A).

Group A, B, and C core substructures were designed and milled by computer-aided design/computer-aided manufacturing (CAD/CAM) technique. For group D, a 3D dental scanner with its scanning and design software (3-Shape A/S, Copenhagen, Denmark) and a PRO JET 3D Printer (3D Systems Corporation, Rock Hill, SC) were used to make identical wax patterns similar to group C design. Figure 2 shows the occlusal and lingual views of the CAD/CAM milled zirconia copings and virtually generated wax pattern for metal ceramic crown. Figure 3 qualitatively illustrates how core thickness differences affect the space (between the coping and the index) available for veneering.

The metal copings for group D were then invested and cast using a 62.5% Pd, 22% Ag, 2% Au ceramic alloy

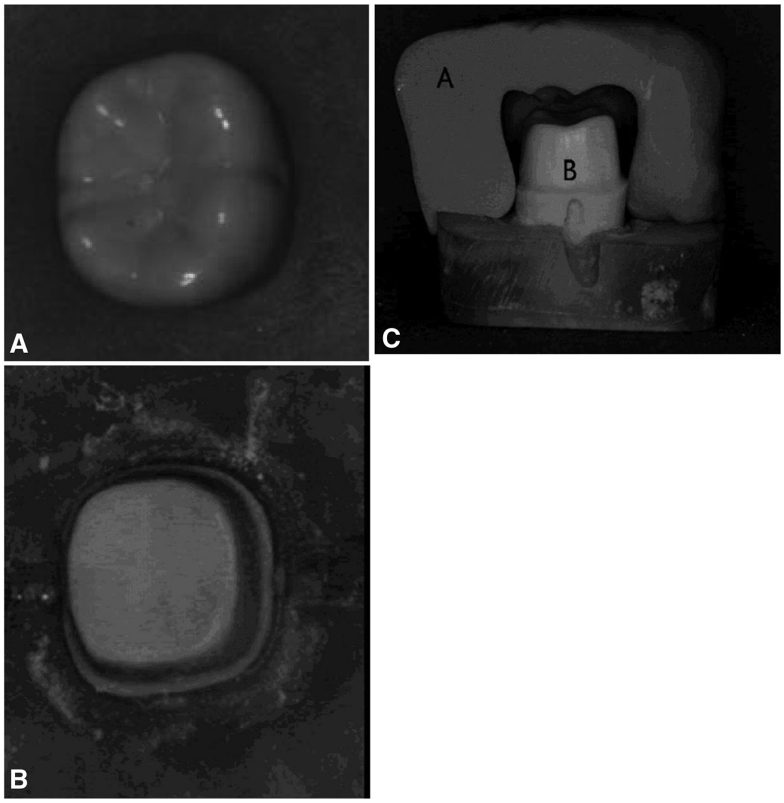

Figure 1 Preparation of tooth and index. (A) Unprepared tooth, (B) prepared tooth, and (C) index placed on prepared tooth (A: index, B: prepared tooth). The gap between the tooth and the index determines the available space for the crown within which the core and veneer thickness are varied for different groups.

(Superior Plus Dental Alloy, Jensen Dental Inc., North Haven, CT) in an induction furnace according to the manufacturer's instructions. The same ceramist (Marotta Dental Lab, Farmingdale, NY) applied the layering porcelain to all specimens, following the recommended techniques. Jensen Creation CC (Jensen Dental Inc.) for metal ceramic crowns and Jensen Creation ZI (Jensen Dental Inc.) for zirconia crowns were used. The core thickness variations in different groups restricted the space for maximum occlusal veneer thickness as follows: group A 1.8 mm; group B 0.7 mm; group C 1.2 mm; group D 1.2 mm.

The porcelain application for the ceramic alloy followed the recommended procedures in the Jensen Creation CC manual as follows: (a) first opaque layer: preheat to 550°C, dry 6 minutes, heat to 980°C at 80°C/min under vacuum, turn off vacuum, hold 1 minute, cool inside furnace with door open; (b) second opaque layer: same as the first layer except for the final temperature of 950°C; (c) third and fourth (dentin) layers: preheat to 580°C, dry 6 minutes, heat to 920°C at 55°C/min under vacuum, turn off vacuum, hold 1 minute, and cool as in (a); final (glaze) firing: preheat to 600°C, dry 2 minutes, heat to 930°C at 55°C/min (no vacuum), and cool as in (a). The porcelain application for the zirconia crowns followed the recommended procedures for Jensen ZI in its manual as follows: (a) first and second pore firing: preheat to 450°C, dry 6 minutes, heat to 810°C at 45°C/min under vacuum, turn off vacuum, hold 1 minute, and cool in furnace with door open; (b) final glaze firing: preheat to 480°C, hold 2 minutes, heat to 820°C at 45°C/min (no vacuum) and cool as in (a). The same operator cemented all crowns to the prepared tooth models using self-adhesive cement (RelyX Unicem, 3M ESPE, St. Paul, MN) according to manufacturer's instructions.

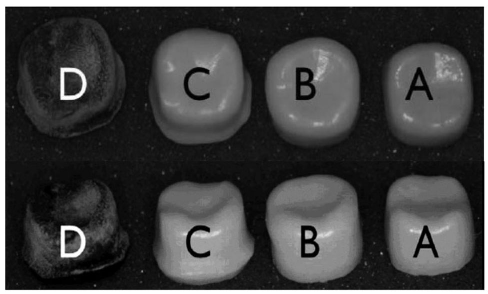

Figure 2 CAD/CAM prepared zirconia copings for groups A, B, and C, and wax pattern to be used to cast metal coping for metal ceramic control group D. The top half shows the occlusal views, and the bottom half the lingual views.

Loading of crowns cemented to tooth models

All loading experiments were carried out in a pre-calibrated MTS Universal Mechanical Testing Machine Model 810 (Material Testing Systems, St. Paul, MN). The loading was done at the center of the tooth using a 6 mm diameter tungsten carbide ball attached to a stainless steel rod (Test Resources, Shakopee, MN) (Fig 4). Axial loading with a spherical indenter, mimicking Hertzian contact conditions, is now routinely used to

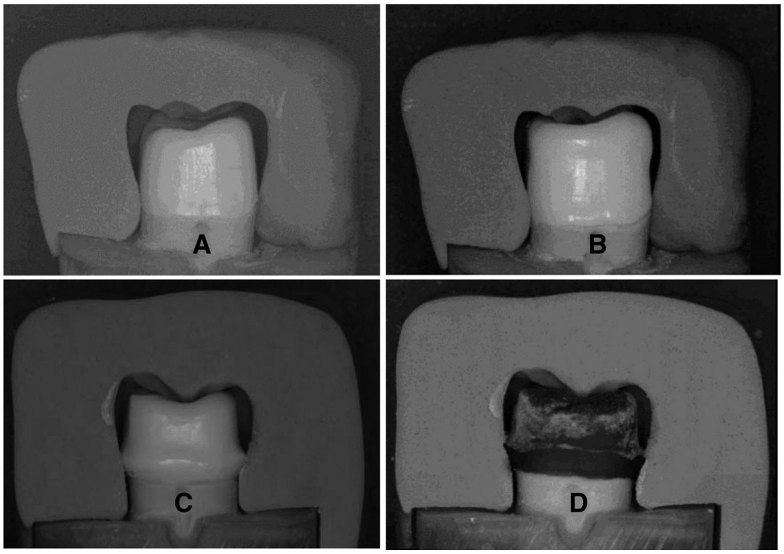

Figure 3 Illustration showing how different coping designs (for groups A, B, C, and D) affect the space for veneering porcelain. The ceramist uses the index to guide him/her to build up the feldspathic porcelain layer within the available space.

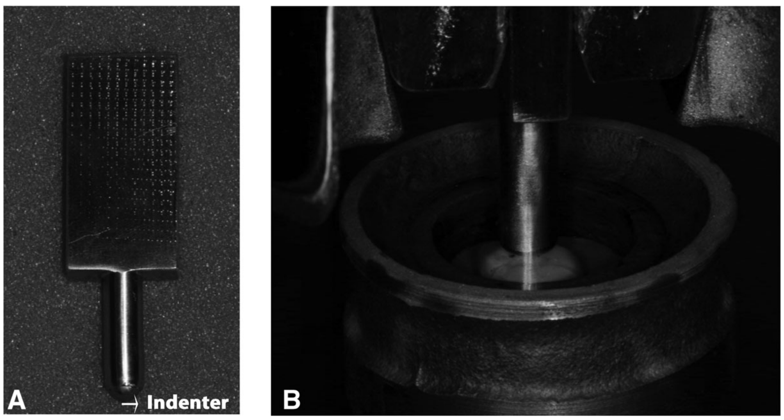

Figure 4 Loading setup for fatigue and fracture tests. (A) Loading fixture with 6 mm silicon carbide sphere (see arrow) attached to the bottom of the cylindrical rod of the fixture. (B) Loading test setup with the fixture positioned to apply load on the crown in a container used to provide an aqueous environment.

study fatigue and fracture resistance in ceramic restorations.[46] Such loading is especially important for following the chipping and fracture failure of the weaker veneer material. The loading consisted of two sequential steps of (a) programmed sinusoidal fatigue loading to simulate chewing attrition followed by (b) a single controlled incremental load to failure to assess the effect of prior fatigue loading on subsequent fracture failure load (FFL).

Fatigue loading of each specimen was done in an aqueous environment to simulate the salivary environment in the mouth. All tests were done at the ambient temperature of 27°C. Each crown cemented to the tooth model support was cyclic-loaded under a preprogrammed fatigue-loading regimen involving sinusoidal compressive cyclic indentation loading (minimum compressive load: 10 N; maximum load: 210 N; mean level: 110 N; frequency: 10 Hz; total of 100,000 cycles) in the MTS machine. After the fatigue loading simulations were completed, the crowns were subjected to a single controlled incremental compressive axial loading at a 1 mm/min crosshead speed under stroke control (with the same tungsten carbide indenter) until fracture occurred. Each specimen was loaded to onset of fracture, and the fracture failure load (FFL in N) was recorded as the peak (maximum) load recorded prior to the onset of fracture.

Fracture surfaces were also examined both visually and under a light microscope to characterize the failure modes. Failure modes were classified microscopically into two types.

1. *Cohesive failure*: Chipping or fracture occurring within the veneer material. This type of fracture is the cone fracture reported under axial loading. The fracture was classified into major or minor depending on the extent of fracture in the veneer and the clinical options to restore function and esthetics to the restoration (minor: can be repaired clinically by finishing and polishing; major: requires replacement clinically).
2. *Adhesive failure*: where the fracture reached the core/veneer interface and propagated through the interface. This type of failure can be expected in a bilayer core/veneer structure where the core is tough, but the veneer is brittle. The core toughness allows it to flex, but the brittle veneer cannot support flexure, resulting in interfacial crack or tear propagation.

Table 1 Fracture failure load (FFL) means, standard deviations (SD), standard errors (SE), Tukey contrast, and *p* value

Groups	Sample size	Mean FFL, (N)*	SD	SE	*p* value
D	20	2118.99[a]	144.46	32.30	$p = 0.001$ (indicates significant differences between group means)
B	20	1841.59[a,b]	372.60	83.31	
A	20	1653.94[b]	391.45	87.53	
C	20	1586.74[b]	418.71	93.63	

Identical superscript letters indicate that the means belong to a homogenous subset.

Table 2 Distribution (number and percent) of failure types among fractured specimens

	Failure mode					
	Minor cohesive		Major cohesive		Adhesive	
Group	# of specimens	%	# of specimens	%	# of specimens	%
A[a]	1	5	2	10	16	80
B	9	45	2	10	9	45
C	5	25	6	30	9	45
D	1	5	3	15	16	80

[a]One specimen in group A failed by fracture through core and was not included in the table.

Statistical analysis

Statistical analysis of FFL data was performed using SPSS v.15 for Windows (SPSS Inc., Chicago, IL). The procedure involved Shapiro−Wilk *W* test for normal distribution followed by appropriate one-way ANOVA of the FFL data. Post hoc Tukey multiple comparisons were used to identify statistically significant differences ($p < 0.05$) between means of individual groups.

Results

The range of FFL values (N) for the groups were as follows: group A, 1160 to 2179; group B, 1240 to 2180; group C, 1018 to 2178; group D, 1720 to 2178. The data showed differences in fracture load means between groups. The results suggest that core thickness differences may influence the fatigue effects on the post-fatigue fracture resistance of the crowns.

Analysis by Shapiro−Wilk *W* test revealed that the FFL data were approximately normally distributed. The variances were not, however, homogenous, and the Welch method of one-way ANOVA was therefore used to statistically characterize the significant differences between means. Post-hoc Tukey tests ($\alpha = 0.05$) were also used to determine significant differences between pairs of group means. Table 1 gives the means, standard deviations (SDs), and standard errors (SEs) of each group, together with the Tukey test. The FFL means fall into two homogenous subsets (within which group means were not statistically significantly different) with one subset containing D and B groups, while the other subset contains A, B, and C groups. The results suggest that the FFL mean of group B crowns is not significantly different from that of metal ceramic control (group D) mean ($p > 0.05$), whereas group means of A and C are significantly different ($p < 0.05$) and lower than that of control group D. It was concluded that zirconia copings designed with extra occlusal core thickness of 1.7 mm (group B) improved the chipping/fracture failure resistance of the veneer porcelain.

Visual examination of the fractured specimens revealed that in 79 specimens (out of 80), fracture occurred only in the veneer or veneer/core interface. We have treated both fracture within the veneer and fracture along the veneer/core interface as chipping in this study. Bulk fracture may also occur through the core, but this occurred only in one specimen in group A, and was treated as an outlier. Visual and microscopic examination nevertheless revealed that chipping failure modes were not identical and included major and minor cohesive as well as adhesives types of failures, as described in the Materials and Methods section.

The number of specimens in different failure mode categories is summarized in Table 2. One zirconia crown from group A failed catastrophically through the core, and was not included in the table. The distribution of failure patterns and the corresponding failure loads were also useful in assessing differences between groups. In groups A and C, failures mainly occurred in major cohesive or adhesive mode of fracture, and typically occurred at relatively lower FFL values than in B and D groups. In group B, failures in 45% of the specimens were only minor cohesive failures, and were clinically repairable. The rest of the fractures were major cohesive or adhesive failures. In group D, failures typically occurred by adhesive mode, but at higher FFL values.

Discussion

We tested the null hypothesis that there is no difference in chipping or fracture resistance of veneer porcelain with different zirconia core thickness support designs in the crown preparation. The null hypothesis was rejected, as the results indicated that the mean fracture load of post-fatigued zirconia crowns with improved occlusal core support was higher, and not statistically significantly different from that of the metal ceramic positive control.

The range of FFL values (1018–2179 N) in this study is in close agreement with the load range (1111–2295 N) reported by Rosentritt et al[39,40] and mean load values (2135–2190 N) reported by Tsalouchou et al[43] for veneer fracture in zirconia crowns after simulated fatigue loading. Other reports in the literature show lower as well as higher veneer fracture loads.[36,41–42,46–48.] It is well known that fracture load is dependent on the type and direction of loading, location of loading on the crown, the tooth model used as crown support during loading, and other variables (e.g., strain rate) that may be present. The differences between reported fracture loads may be the result of these variables in experiments used by different authors.

Nobel Procera zirconia used in this study is a partially stabilized TZP with 4.5% to 5.5% yttrium oxide (Y_2O_3). It has a high flexural strength (1150 MPa) and fracture toughness

(6–9 MPa.m$^{1/2}$) because of a very fine grain size (0.3–0.5 μm) and stress-induced transformation toughening property. The fracture toughness of veneer porcelain is unfortunately not optimal because of its brittle characteristics. In addition, the veneer porcelain at the occlusal surface of the crown is in contact with the opposing dentition during chewing. Consequently the masticatory stresses generated during chewing are directly applied on the veneer porcelain in the occlusal region. We designed the core support variations to test our hypothesis that extra thick core support at the occlusal loading site may help improve veneer fracture resistance. Group A had a uniform core thickness of only 0.6 mm, group C had a uniform core thickness of 1.2 mm, and group B had extra occlusal core support (1.7 mm), but only 1.2 mm thickness elsewhere, as in the case of group C. The (occlusal) veneer thickness in zirconia crowns was highest (1.8 mm) in group A, intermediate (1.2 mm) in group C, and lowest (0.7 mm) in group B. These results suggest that when the veneer porcelain is thicker and is inadequately supported by zirconia core at the occlusal surface, as in groups A and C, the brittle nature of the veneer porcelain may determine the chipping and fracture behavior. The lower FFL means of groups A and C are in keeping with this rationale. In contrast, the mean FFL of zirconia crowns in group B was higher, and it is reasonable to suggest that extra occlusal core support is responsible for this improved performance. The additional occlusal core thickness was achieved with only 2.4 mm occlusal tooth reduction. From a clinical point of view, optimized thickness of tougher occlusal core support under the weak veneering porcelain may help enhance resistance against veneer fracture in zirconia restorations under chewing attrition.

As pointed out in the introduction, thermal expansion mismatch between the coping material and porcelain, improper firing and cooling, or improper coping build-up are leading causes for chipping in porcelain veneer.[27-29] In this research, the veneering porcelain used was selected with thermal expansion compatibility to zirconia, and the proper guidelines for porcelain firing and cooling were followed to avoid internal residual stresses. The only variable was the difference in core support for the veneer. In addition, actual crowns were tested in the experiments in contrast to tests on flat discs reported in the past. We also used an aqueous environment simulating the clinical situation. Our tests therefore better represent potential clinical outcomes. The results of our study are in agreement with the results of White et al,[48] who also used crowns in their tests to demonstrate differences in fracture resistance with increased core/veneer thickness ratios. On the other hand, our results appear to differ from reported results of Lohbauer et al,[34] where they used flat specimens in an edge-chipping study and reported no difference in chipping resistance between metal ceramic and zirconia veneered specimens. Since the types and distributions of stress may differ markedly between loading sites and between anatomic and flat surfaces, the difference between our results and those of the Lohbauer study may be due to these differences.

In a SEM study, Tholey et al[35] studied veneered zirconia specimens prepared with different methods of porcelain build-up, and suggested there was evidence of a greater degree of faceting on Y-TZP grains underneath a thicker veneering porcelain. Their interpretation was that the thicker layer of veneering porcelain would contain more moisture, which would cause the water radicals to diffuse to the crystal grains during porcelain firing, leading to tensile stress formation, creating a destabilized tetragonal phase at the core/veneer interface. This moisture effect may also potentially interact with stress distribution effects to weaken the interface between the veneer porcelain and zirconia core. The adhesive failures at lower FFLs in groups C and A are in keeping with this effect.

Analysis of the distribution of failure modes and the respective load levels of fracture also suggest that core support at the loading site significantly influences failure differences. In control group D and extra-thick occlusal core support group B, 80% and 45% were adhesive failures, respectively, and typically occurred at higher load levels; 45% of failures in group B were also minor cohesive failures. In contrast, adhesive failures (group A: 80%, group C: 20%) in the lower occlusal core thickness groups occurred at relatively lower load levels. Thus, the reduced veneer thickness at the loading sites in zirconia crowns may enhance the role of the core support in resisting fracture, influencing the load levels of adhesive fracture, and the mode of cohesive fracture within the veneer, leading to differences in fracture modes.

One of the limitations of this study was the lack of periodontium simulation. The presence of the periodontal ligament (PDL) around the teeth allows for some degree of mobility during the applied load. This will lead to a decrease in the stress accumulated within the restoration.[30] Another limitation is that axial compressive forces were applied using a tungsten carbide sphere to the center of the crowns in this study. This is somewhat different from a full chewing simulation involving posterior teeth where the tooth is not only subjected to axial loading at the centric occlusion, but also to off-axis loading on the cuspal inclines.[46] Flexural stresses generated due to off-axis loading were not addressed in this study. The use of a 10-Hz fatigue loading frequency is also significantly higher than the typical normal in-mouth chewing frequency of 1 to 2 Hz. Ten Hz was used to reduce the duration of individual tests, since it is a common laboratory practice to use accelerated conditions of testing to assess test effects for a longer attrition period in a shorter test duration without compromising the essential features of testing simulation. Composite resin tooth replicas were used instead of the metal dies to simulate the resiliency of dentin and to avoid possible internal crack initiation to the core material. Coelho et al[21] also used this material to successfully simulate tooth support for zirconia crowns in their fatigue studies.

Although the forces generated during mastication vary widely between approximately 5 and 364 N,[16-17] typical total chewing/swallowing forces of mastication average about 100 N,[18] and therefore sinusoidal load-cycling between 10 and 210 N (with a mean level of 100 N) is relevant to normal chewing simulation. The maximum clenching force (N) reported in the literature ranges from 216 to 880.[19-20] Thus, the mean post-fatigue FFL (>1840 N) for crowns with extra-thick occlusal core support (group B) was more than twice the expected highest force level encountered by restorations in posterior teeth. This is better than a 100% safety factor for their use in posterior applications, and provides a valid rationale for their use.

Conclusions

Within the limitation of this study we conclude the following.

1. The mean post-fatigue fracture failure loads recorded for all zirconia groups (A-1653 N, B-1841 N, C-1586 N) were significantly higher than the maximum clenching force that can be generated intraorally (880 N). The fracture resistance of group B was more than twice the maximum clenching force, providing a greater than 100% safety factor against failure, even under the highest intraoral stress levels.
2. When the zirconia core support was maximized occlusally as in group B, the mean failure load came closer to that recorded for the metal ceramic group.
3. When the zirconia core support was maximized occlusally as in group B, 45% of the failed specimens had minor chipping, which can be corrected clinically. In group A where the support was minimal, only 5% of the failed specimens showed minor chipping. In group C, where the core support was uniform, but intermediate, 25% showed minor chipping.
4. It is important to maximize the support occlusally where the masticatory stresses are concentrated to achieve the best results.

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