Original Study

EFFECT OF ZIRCONIA COPING DESIGN ON FRACTURES RESISTANCE OF VENEERING PORCELAIN IN ALL-CERAMIC POSTERIOR CROWNS

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ABSTRACT

Aim: When the zirconia core is fabricated in a uniform thickness, and veneering porcelain has variable thicknesses, it maybe undergoes chipping. This study aimed to compare the fracture load of restorations fabricated with two different zirconia core designs.

Materials & Method: Twenty four metal dies were divided into 2 groups for fabrication of zirconia cores: conventionally (0.5-mm thickness); customized design. After the porcelain was applied, the crowns were cemented. Using a universal testing machine a compressive vertical load was applied until failure.

Results: Data were analyzed with SPSS 17 software, using t-test, chi-squared test and Mann-Whitney U test. The mean fracture loads were 2254.1773±404.4 N and 2885.4327±670.5 N for conventional and customized zirconia core designs, respectively.

Conclusion: Student's t-test revealed that the fracture load of the customized group was significantly higher than that of the conventional group (P<0.001). Crowns with customized design have higher fracture resistance than those with conventional design.

Key words: All-ceramic restoration, Fracture, Zirconia core.

Introduction

At present, there is plenty of information available about new materials and techniques in dentistry. This has increased the demand for highly esthetic dental restorations. Patients are well aware that ceramics are the most suitable option to simulate natural teeth and know that ceramic dentistry has been revolutionized in recent decades. However, the esthetics of dental restorations must not compromise their resistance against maximum clenching forces, which is approximately 1031 N in partially edentulous and 1243 N in fully dentate patients.[1]

Metal–ceramic restorations have been used in dentistry for more than half a decade century. Improvements in alloys, substrates and porcelain veneers in recent years have resulted in high acceptance of these restorations by patients. However, due to the presence of metal frameworks and its related opacity, metal–ceramic restorations do not provide excellent esthetic results, especially when observed directly.[2] Metal frameworks may lead to discoloration of the teeth and soft tissues and may also cause allergic reactions.[3,4]

The advances in science and technology, the increased need for esthetic treatments, and the questionable biocompatibility of metals and alloys used in metal–ceramic restorations have all contributed to the growing popularity of all-ceramic restorations in contemporary dentistry.[5] Favorable optical characteristics, esthetic properties and biocompatibility of ceramics are the main reasons for the popularity of all-ceramic restorations in dentistry. Due to its brittle nature, porcelain has high compressive and low tensile strengths.[6] Porcelain fracture is the common form of failure of all-ceramic restorations.[7,8] Thus, for all-ceramic restorations, a core must be fabricated to minimize the tensile load applied to the porcelain veneer. Adequate thickness of the porcelain and core can decrease the concentration of internal stresses and the consequent mechanical failure and increase restoration's esthetics.[9] Researchers and manufacturers came up with an advanced formula using the yttria tetragonal zirconia polycrystalline ceramics called zirconia to prevent crack formation.[10,11]

In the majority of all-ceramic systems, the zirconia substructure is fabricated via a specific computer-aided manufacturing (CAM) process. Next, the fabricated core is veneered by the conventional porcelain, using the pressing method or the layering technique. Thus, the zirconia core provides optimal support for the veneering porcelain.[12] However, factors such as the thickness of the veneering porcelain, defects in the bond between the veneering porcelain and the zirconia core and the weak nature of this bond can cause porcelain delamination, exposure of the zirconia core, and chipping of the veneering porcelain, resulting in the eventual failure of zirconia restoration.[13] In some all-ceramic systems, the zirconia core is conventionally formed in a single layer with a uniform thickness. Thus, the veneering porcelain has variable thicknesses at different areas and thus is subjected to excess loads and more rapidly undergoes chipping and eventual failure.[14] Previous studies have investigated the effect of coping design on the fracture resistance of porcelain veneers and this effect has been confirmed by some[15–17] and rejected by some other studies.[8,18,19]

This study aimed to compare the in vitro fracture load of the porcelain veneer in all-ceramic posterior restorations with two different zirconia coping designs.

Materials and Method

A maxillary first premolar phantom model (Nissin, Dental Product Inc., Kyoto, Japan) was embedded in an acrylic resin block (GC Pattern Resin, ALSTP, IL, USA) in such a way that the acrylic surface was 3 mm below the cemento-

enamel junction (CEJ) of the model. Using putty soft silicon impression material (Zhermack, Elite HD, Italy), a putty index was obtained from the model. Anatomical preparation for an all-ceramic crown was carried out to one millimeter above the CEJ using a milling machine (Degussa, Germany). The details of the preparation design were as follows: Occlusal reduction of 1.8 mm, axial wall reduction of 1.5 mm with 8° taper and radial shoulder finish line at the margins measuring one millimeter in width. Sharp angles and points were rounded. The prepared model was covered with Easy-Vac Gasket (3A MEDES, Korea) polyethylene sheets with a 2-mm thickness using a vacuum former (UltraVacm Ultradent Product Inc., USA). Using light-cured resin (Megatray, Germany), 24 special trays were fabricated from the prepared model.

Using trays, 24 impressions were made from the prepared model using Impregum (3M ESPE, USA) impression material and poured with hard wax. The fabricated wax models were cast with base metal alloy (Verabond, USA), using the lost wax technique. Thus, 24 master metal dies were fabricated for the production of zirconia cores. [Figure 1]

Figure 1: Master matel die with 8 ^{o} taper

In this study, CAD/CAM was used for the fabrication of zirconia cores from Cercon zirconia (DeguDent, GmbH, Germany)

Material	Manufacturing Company	Youngs Modulus (GPA)	CTE 20-500^{o}C (ppm/^{o}C
Cercon Core	DeguDent, GmbH, Hanau- Wolfgang, Germany	205	10.5
Cercon Cerams	DeguDent, GmbH, Hanau- Wolfgang, Germany	69	9.7

Table 1: Physical properties of the core material and the veneering porcelain.

Table 1 summarizes the physical properties of the core material and the veneering porcelain used. All the metal die specimens were scanned and divided into two groups. In group one, 12 conventional zirconia cores were fabricated with 0.5-mm thickness, 25-µ space for the cement and 85% die coverage with die spacer. In group two (customized), the software virtually performed a full contour wax-up and cut-back and anatomically designed the core in such a way that a uniform thickness of porcelain veneer could be achieved in all areas. A reinforcing collar with 1-mm height was also created. Then the specimens were placed in the Cercon Heat (Degudent, GmbH, Hanau-Wolfgang, Germany) and subjected to sintering for 6 hours to reach their ideal size and final strength. After preparation of zirconia cores, they were primarily seated on the metal dies. Zirconia cores were surface treated with 50-µ alumina particles from a 10-mm distance at 3-bar pressure using the air abrasion machine (Easy, Blast, Bego, Germany) [Figures 2 and 3].

Figure 2: Conventional Zirconia Core

Figure 3: Customixed Zirconia Core

Porcelain powder (A4 shade, Cercon Ceram, DeguDent, GmbH, Germany) was applied to the surface of all the specimens by an expert technician, using the previously prepared silicone index. The porcelain was applied at a firing temperature of 830°C in two phases of opaque porcelain application and one phase of dentin baking and the restoration was then glazed. The crowns were cemented with Panavia F 2.0 (Kuraray Medical Inc., Osaka, Japan) over the respective metal die. Using finger pressure, a gentle force was applied to the die–crown complex for 5 minutes. Primary curing was carried out for 5 seconds using Coltolux light-curing unit (Coltene, Germany). Excess cement was removed by the tip of an explorer and each surface was light-cured for 40 seconds for final setting.

A universal testing machine (Zwick, UTM, Germany) was used for load testing. A polyethylene sheet, 2 mm in thickness, was placed over each crown in order for the loads to be applied to the specimen surfaces more efficiently. Static vertical load was compressively applied

with a stainless steel metal ball measuring 4 mm in diameter at a crosshead speed of 0.5 mm/min to the specimen surface until fracture. The load at fracture for each specimen was recorded by the machine. A stereomicroscope (Koops Pazhoohesh, Iran) was used to assess the failure modes.

Normal distribution of data in the conventional and customized groups was tested and confirmed using Kolmogorov-Smirnov test (p=0.046). Data were analyzed with SPSS 17, using t-test, chi-squared test and Mann-Whitney U test. The fracture resistance in the two groups was assessed and compared using Student's t-test. Equality of variances in the two groups was tested and confirmed using Levene's test (p=0.038). Type I error was considered at 0.05 and P≤0.05 was considered statistically significant.

Results

During fracture resistance testing by the UTM one specimen in each group was lost. Thus, assessments were made on the remaining 11 specimens in each group.

The mean fracture load was 2254.1773±404.4 N for crowns with the conventional zirconia core and 2885.4327±670.5 N for crowns with the customized zirconia core. Student's t-test revealed that the fracture load in the customized group was significantly higher than that of the conventional group (P<0.001). These data are demonstrated in Table 2 .

Group	No.	Mean	SD	Standard Error
Convertional	11	2254.17	404.4	121.93
Customized	11	2885.43	670.5	202.16

Figure 2: The mean fracture strength in the customized and convertional groups in Newton

Evaluation of fracture modes under a stereomicroscope in the customized group revealed that in two specimens the zirconia core had fractured along with the porcelain veneer. In one specimen, chipping of the zirconia core margin had occurred along with bulk porcelain separation from the underlying zirconia core surface. In the remaining eight specimens, only bulk separation of porcelain from the zirconia core surface was noted.

In the conventional group, in three specimens fracture of the zirconia core had occurred along with the porcelain veneer. Chipping of the zirconia core margin was observed in two specimens along with bulk separation of porcelain from the surface of zirconia core. In the remaining six specimens, only the bulk separation of porcelain from the zirconia core surface had occurred. In both groups, the fracture of porcelain had occurred in the lingual surface with greater extension towards the mesial surface.

Discussion

Based on the results, the hypothesis on the higher fracture load of all-ceramic restorations with customized core design compared to conventional core design was accepted.

Kokubo *et al*, in 2011, reported that the zirconia coping design with anatomical cusps had the highest amount of fracture load (4100 N).[17] Silva *et al*, in 2011, reported a fracture load of 1134 ± 182 N for standard design and 767 ± 154 N for cores with a modified design. The most common mode of fracture was the bulk fracture of the veneering porcelain.[15] Tsalochou *et al*, in 2008, reported a fracture load of 2135.6±330.1 N with the use of a metal die and zinc phosphate cement.[20] Sim HB *et al*, in 2010, evaluated the fracture resistance and marginal fidelity of zirconia crowns based on coping design and type of cement. The zirconia core designs were as follows: Group I: uniform thickness of 0.3 mm; group II: 0.3 mm in the buccal half and 0.6 mm in the lingual half; group III: 0.6 mm in both buccal and lingual halves; and group IV: one millimeter of thickness in the lingual and 0.6 mm of thickness in the buccal half. Some specimens were cemented with Cavitec and some with Panavia F. Irrespective of the crown design, the fracture resistance of crowns cemented with Panavia F was higher than that in the other group. The lowest marginal fidelity was seen in groups I and III, which were bonded with Cavitec, while the greatest marginal fidelity was seen in group II. Fracture resistance was higher in groups III and IV, which were cemented with Cavitect; among groups cemented with Panavia F, group II exhibited the highest fracture resistance.[19] Pallis *et al*, in 2004, reported a fracture load of 918–1183 N for zirconia crowns cemented with Rely X cement on resin dies.[21] Based on the results of the current study, the mean fracture load was 2254.1773±404.4 in the conventional group and 2885.4327±670.5 N in the customized group. Gibbs in 2002 reported a maximum clenching force of approximately 1243 N in fully dentate patients; on the other hand, he stated that loads within this range are compatible with clinical conditions.[1] According to Scherrer and de Rijk in 1993, dies with high modulus of elasticity increase the fracture load of porcelain crowns attached to them; their study was not an exception, either. Thus, these results are not comparable with the maximum clenching force applied in the clinical setting.[22]

Pattern of fracture and separation of porcelain from the zirconia core in this study occurred in the lingual surface slightly towards the mesial aspect, which was consistent with the results reported by Rosentritt *et al* in 2009. They reported that lingual inclination of the crown complex might play a role in this respect and further direct the loads towards the lingual surface.[23] Regarding the chipping of the zirconia core margins, it appears that the shear stresses were concentrated at the sharp edges, and inadequate support of the zirconia by the core material at this area might be responsible for the chipping of zirconia margins. Fracture of the core together with the veneering porcelain occurred in 5 specimens in this study. Sundh and Sjogren in 2004 reported that the odds of occurrence of this type of fracture increases if a metal die or other materials with high modulus of elasticity are used as the master die, while they also reported the fracture of natural teeth when used as the master die.[24]

Currently, fatigue failure due to cyclic loading or thermal loading has been suggested as a possible factor responsible for the failure of dental restorations.[25] Future studies are required to assess the fracture load of zirconia-based crowns with this design under humid conditions with cyclic loading.

In the current study, Panavia F 2.0 cement was used and a 25-µ space was allowed for the cement. Different cements, including conventional and adhesive types, have been used for cementation in previous studies to bond the crown to the respective die. Attia *et al*, in 2006, demonstrated that adhesive cements significantly increased the strength of crown complex and consequently the fracture load compared to the conventional types.[26] Also, concerning the space required for the cement, Rosentritt *et al*, in 2009, reported a difference in cement space size (from 10 to 40 µ) had no significant effect on the fracture load of all-ceramic crowns with zirconia cores.[23]

Based on the available statistics, despite great advances in the fracture resistance of dental ceramics (using alumina and zirconia cores), the failure rate of posterior all-ceramic restorations is 3–4% annually. This finding indicates that a complex scenario, other than the catastrophic fracture due to overload, plays an important role in the initiation of damage to the ceramic system.[27] A significant difference exists between zirconia and metals in bonding to porcelain. Bonding of metals to the veneering porcelain is favorable due to the chemical nature of the bond and its optimal quality (attributed to the adequate thickness of the oxide layer and suitable exchange of ions at the interface) as well as the micromechanical interlocking. However, no clear data is available regarding the porcelain bond to zirconia. The wettability of the zirconia core by the porcelain and the micromechanical bond between them are the only mechanisms known so far; this bond is weaker than the metal–ceramic bond.[28] Therefore, before adding the porcelain, the core surface must be sandblasted according to the manufacturer's instructions.

Guess showed that the process of sandblasting with particles 100 µm in diameter had no significant impact on the shear bond strength of zirconia to the veneering porcelain in the Cercon system compared to systems not requiring sandblasting.[29] In this study, surface treatment of cores was performed at a 10-mm distance with 50-µ particles at 3 bars of pressure.

Thermal conductivity (TC) is also an influential factor in this regard. Metal alloys have a high TC (300 $Wm^{-1}k^{-1}$) while zirconia cores act as insulators. Based on the data provided by different manufacturers, the TC of zirconia cores is 2–2.2 $Wm^{-1}k^{-1}$. Veneering ceramics also have a TC within the same range, i.e. 2.39 $Wm^{-1}k^{-1}$. The low combination of the TC of the core and the veneering porcelain delays thermal loss at the interface in comparison with metals, changes the linear contraction of porcelain and the zirconia core, and creates thermal stresses at this area, which per se can cause porcelain delamination over time. On the other hand, a change in the ratio of the thickness of core to the veneering porcelain at different areas of the crown can lead to the formation of excess stresses in the thermal cycles of porcelain baking.[30]

Conclusion

Within the limitations of this study, it may be concluded that customized core design significantly increases the fracture resistance of all-ceramic posterior crowns compared with the conventional design, and the obtained crowns undergo fracture at a significantly higher load.

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