Effect of Zirconia Substructure Design on the In-Vitro Fracture Load of Molar Zirconia Core Crowns

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Abstract

Objective: In the available all-ceramic systems the zirconia core is commonly fabricated as one layer with an even thickness, but the veneering porcelain has different thicknesses in different parts of the restoration and therefore, undergoes chipping and fracture more rapidly under masticatory forces. This study aimed at comparing the fracture load of all-ceramic crowns with 2 different zirconia core designs in Cercon system and in-vitro conditions.

Methods: In the present experimental study, 10 metal dies of mandibular first molar were fabricated with NNB base metal alloy using lost wax technique. Ten standard zirconia cores were then fabricated with an even thickness of 0.5 mm using Cercon CAD/CAM System. Also, 10 more were fabricated in the customized form with a 1 mm labial collar and 2 mm lingual shoulder. Porcelain was applied on all samples by an expert technician using an index. The prepared crowns were placed on their respective dies and cemented with Panavia F resin cement and under the constant pressure of 25 N. Vertical compressive force was applied to the samples by means of a stainless steel ball at a crosshead speed of 0.5 mm/min using Universal Testing Machine until failure. Data were analyzed using student t test.

Results: Our study results demonstrated the fracture load of 1852.11±587.9 N for zirconia core crowns with standard core design and 3332.63±916.38 N for custom design crowns. Statistical analyses demonstrated that the fracture load was significantly higher for customized core designs than the standard design cores(P<0.0001).

Conclusion: Considering the obtained results we can conclude that crowns with customized core design have a greater fracture resistance than those with standard design.

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

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Introduction:

During the past 40 years, metal ceramic restorations have always been a reliable treatment modality. This treatment is still considered among the ideal dental therapies. However, advancements in science and technology, increased demand for improving the esthetics and doubtful compatibility of metals and alloys used in metal ceramic restorations have all resulted in growing popularity of all ceramic restorations in contemporary dentistry (1).

The brittle nature of dental porcelains necessitates the need for an appropriate substructure (core) to support the veneering porcelain in all ceramic restorations. In the early nineties, partially stabilized Yttrium Oxide Tetragonal Zirconia Polycrystal (Y-TZP) was introduced as a sub-structure for all ceramic
restorations. Due to the transformation toughening mechanisms, zirconia has better mechanical properties compared to other core materials in full ceramic systems. Therefore, we are now witnessing an increasing trend in application of this material (2).

In the majority of zirconia all ceramic systems, this substructure is fabricated via a special CAM process. Then the fabricated core is veneered with the conventional porcelains through layering or pressing technique. Zirconia core provides a good support for the veneering porcelain (3). However, factors like the thickness of veneering porcelain, limitations in the bond of veneering porcelain and zirconia core and weak nature of this bond can result in porcelain delamination, exposure of zirconia substructure or chipping of veneering porcelain and subsequent fracture and failure of zirconia fixed prosthesis (4).

In the available full ceramic systems, zirconia core is commonly formed as one layer with an even thickness. Therefore, the veneering porcelain has different thicknesses in different parts of the restoration and undergoes chipping and fracture more rapidly under masticatory forces (5). Limited number of studies have evaluated the effect of core design on the fracture resistance of porcelain. Therefore, this study assessed the effect of core design on fracture resistance of veneering porcelain and compared the fracture load of all ceramic crowns in 2 different zirconia core designs in Cercon system and in-vitro conditions.

Methods:

In the present experimental study, number of understudy samples was calculated to be 20 through review of previous studies (5-8), consultation with our statistician and evaluation of trial and error method; out of which, 10 received standard core design and the remaining 10a customized core design. A sound extracted human mandibular first molar was placed and embedded in a mold containing Pattern resin (GC, Japan) in a way than acrylic resin surface was 3 mm below the CEJ. Then, an index was obtained from the tooth with silicone impression material (Zhermack, Elite® HD+ putty soft, Italy) for later use in the veneering phase. Preparation for a full ceramic crown was done using a hand piece and water and air spray up to 1 mm above the CEJ and in an anatomical form with the following characteristics: 1.8-2 mm occlusal reduction, 1.5 mm axial reduction with 8° taper, 1.5 mm radial shoulder at the preparation margin and rounded edges. Tooth surface was then covered with Easy-Vac Gasket polyethylene sheets (3A MEDES, Korea) with 2 mm thickness using a vacuum former (SCHEU minister, Germany), and by VLC light cure acrylic resin (Megatray, MegaDenta, Germany), 10 special trays were fabricated for the prepared tooth. Ten impressions were made from the tooth with Impregum impression material (3M ESPE, USA). The impressions were poured with hard wax. In the next phase, the castings were formed by the lost wax technique and NNB base metal alloy (Sankin-Dentsply, Germany). Ten master metallic dies were fabricated as such (Figure 1).

Figure 1- Metallic die with 8° taper

In order to make impression for fabrication of sample models, 20 special trays (2 special trays on each metallic die) were made using light cure acrylic resin. Twenty final impressions were
made from metallic dies (2 impressions from each metallic die) using Impregum impression material (3M ESPE, USA). Impressions were poured with type IV dental stone (Type IV, Fuji Rock, GC Japan). Twenty stone dies were prepared as such for the fabrication of zirconia cores.

Samples were divided into 2 groups of 10 each. In the first group, after scanning the surface of stone dies, 10 standard zirconia cores were fabricated. The thickness of the copings was set at 0.5 mm. A 25 micron space was provided for the cement using a die spacer. Die coverage was 85% (Figure 2).

Figure 2- Standard zirconia core

For the customized group, the computer software did a virtual full contour wax-up for each sample and did the cut back to form a 2 mm buttressing shoulder at the lingual surface and a 1 mm reinforcing collar at the labial surface (Figure 3).

Figure 3- Customized zirconia core

The contact line of buttressing shoulder and reinforcing collar was placed 1 mm lingual to the proximal surface half line. Other adjustments were made similar to the standard group. After completion, obtained samples were placed in Cercon heat high-temperature sintering furnace and subjected to sintering process for 6 hours to reach their ideal size and final hardness.

After fabrication of zirconia cores, each sample was placed on its respective stone die and the primary seating of all cores was evaluated. Final placement of cores on their master metallic dies was done afterwards. Thus, for each metallic die 2 zirconia cores, one with standard design and the other in the customized form were available.

After completion of this phase, surface treatment of zirconia cores was done from 10 mm distance with 50 micron diameter particles at 3 bar pressure with Air Abrasion device (Easy-Blast, BEGO, Germany). According to the silicone index obtained before tooth preparation veneering porcelain was applied to the models by an expert technician using Cercon® Ceram S porcelain (Degudent, GmbH, Germany) and fired at 830°C temperature in a 2-step opaque firing (in 2 layers) and a single step dentin firing and glazed afterwards.

In the next phase, Panavia F2.0 cement (Kurary Medical Inc, Osaka, Japan) was applied to the inner surface of cores and they were seated on their respective dies. Using a device specifically designed for this purpose, 25 N force was applied to the crown-die complex for 5 minutes. Excess cement was removed with the tip of an explorer and each surface was light cured for 40 seconds with a light curing device (CE/ISO LK-G13, Ivoclar Vivadent) to achieve final setting. To perform load testing, a polyethylene sheet with 2 mm thickness was placed on each crown to efficiently distribute the force on the surface of each sample. Then, compressive vertical static load was applied to the sample surface by means of a stainless steel ball (4 mm diameter) at a crosshead speed of 0.5 mm/min using Universal Testing Machine and continued until failure. The amount of force at which fracture occurred was
recorded by the machine for each sample (Figure 4).

**Figure 4- Application of force to the sample using UTM**

Normal distribution of data for standard and customized groups was analyzed and confirmed using Kolmogorov-Smirnov test ($P=0.69$). Data were statistically analyzed using SPSS version 16 software. Fracture toughness in 2 groups was compared using student t test. The hypothesis of equality of variances in the 2 groups was tested and approved with Levene’s test ($P=0.12$). $P$-values ≤ 0.05 were considered statistically significant.

**Results:**

The mean fracture load was $1852.11\pm587.9$ N in the standard samples and $3332.63\pm916.4$ N in the customized samples. Student t test demonstrated the fracture load to be significantly higher in the customized group compared to the ones with standard design ($P<0.0001$). Data in this respect are presented in Table 1.

**Table 1- Central dispersion indexes for fracture resistance (N) of samples with standard and customized designs**

<table>
<thead>
<tr>
<th>Group</th>
<th>Number</th>
<th>Mean</th>
<th>Standard Deviation</th>
<th>Standard Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standard</td>
<td>10</td>
<td>1852.11</td>
<td>587.9</td>
<td>185.9</td>
</tr>
<tr>
<td>Customized</td>
<td>10</td>
<td>3332.63</td>
<td>916.4</td>
<td>289.9</td>
</tr>
</tbody>
</table>

**Figure 1- Mean fracture resistance (N) of the samples in 2 groups with 95% confidence interval**

Macroscopic evaluation of the mode of fracture in the customized group revealed that in one sample (die number 1) fracture was in the form of a crack on the occlusal surface of crown and separation of porcelain from the core surface was not observed. In 2 cases, fracture of zirconia core along with its veneering porcelain was noticed (Figure 5). In the remaining 7 cases,
bulk separation of porcelain from the core occurred (Figure 6).

Figure 5- Fracture of zirconia core along with porcelain

Figure 6- Bulk fracture of porcelain

Figure 7- Marginal chipping of the zirconia core

On the other hand, marginal chipping of the zirconia core was observed in 9 cases (except for die number 1)(Figure 7). In the standard design group, bulk separation of porcelain from the zirconia core occurred in all samples. In both customized and standard groups, pattern of porcelain fracture was in the lingual surface and towards the mesial.

Discussion:

Considering the obtained results, our suggested hypothesis regarding the increased fracture load of all ceramic crowns with custom core design compared to those with standard design is confirmed. When comparing our study findings with those of others in terms of type of model, the amount of measured force, technique of force application, pattern of porcelain fracture, technique of impression making, type of impression material used, nature of bond between porcelain and substructure material and type of cement used, the following points are worthy of further evaluation:

The amount of measured force:
Coelho et al, in 2009 reported a mean fracture load of 1227±221 N for all ceramic crowns with zirconia cores that had been cemented on acrylic resin dies with Rely X resin cement (6). Sundh and Sjögren in 2004 used metallic dies and zinc phosphate cement and proposed the mean fracture load of 4114±321 N for crowns with zirconia cores (7). Tsalouchou and colleagues in 2008 reported the mean fracture load of 2135.6±330.1 N when using metallic dies and zinc phosphate cement (8). Pallis and coworkers in 2004 reported a fracture load in the range of 918-1183 N for zirconia crowns cemented with Rely X cement on acrylic resin dies (9). Based on the results of the present study, the mean fracture load was 1852.11±587.9 N for the standard and 3332.63±916.4 N for the customized group.
Burke in 1992 reported the maximum masticatory force of about 800 N for natural teeth and considered the forces within this range to be compatible with clinical conditions (10).
According to Scherrer and de Rijk in 1993, dies with high Modulus of Elasticity result in increased fracture load of their veneering porcelain (11). The present study is no exception to this rule. Therefore, our study results are not comparable with the amount of masticatory forces or maximum bite force in a clinical context.

Porcelain fracture pattern:
Porcelain fracture and delamination from the zirconia core in our study occurred in the lingual surface towards the mesial which was in accord with the finding of Posentritt et al, study in 2009 and can be due to the lingual inclination of the crown complex which can be a factor for further lingual transfer of forces (5). Regarding the marginal chipping of zirconia core at the buttressing shoulder, it seems that shear stresses developed at sharp edges and lack of adequate support of zirconia by the core material itself in that area can be the possible reasons for this problem.

Fracture of the core and veneer occurred in 2 of our understudy samples. Sundh and Sjögren in 2004 stated that the risk of this mode of failure increases if metallic die or other materials with high modulus of elasticity are used as the master die. However, they also mentioned fracture of normal teeth when used as the master die (7).

Test environment:
In all ceramic systems, fatigue is defined as subcritical crack propagation within the veneering material under stress and aqueous environment. Despite the high strength reported for zirconia-base ceramics, they are very sensitive to fatigue failure which in long term can significantly decrease their toughness. At present, fatigue failure due to cyclic loading or thermal loading is considered as a possible factor responsible for the failure of dental restorations (12). Further investigations are required to evaluate the amount of fracture load in zirconia-base crowns with this design in aqueous environment and cyclic loading conditions.

Type of cement used:
In the present study, Panavia F cement was used and 25 micron space was provided for cementation. Various conventional and adhesive cements have been used in different studies for the adhesion of crown to the respective die. Attia et al, in 2006 demonstrated that adhesive cements significantly increased the strength of the crown complex and its fracture load compared with conventional cements (13). In terms of cement space, Rosentritt et al, in 2009 explained that changing the cement space (from 10 to 40 micron) did not have a significant effect on the fracture load of all ceramic crowns with zirconia cores (5).

Nature of the porcelain-substructure bond and related factors:
Based on the available statistics, despite great advancements in the field of dental ceramics (use of cores made of alumina and zirconia), failure rate of posterior full ceramic restorations reaches to 3-4% per year. It indicates that a very complex scenario other than catastrophic failure due to overload, plays an important role in initiation of damage to the ceramic system (6). A significant difference exists between zirconia and metals when it comes to bonding with porcelain. In metals, due to the presence of a good quality chemical bond (which is due to the adequate thickness of the oxide layer and adequate ion exchange at the interface) along with micromechanical interlocking, a good bonding forms between the metal and the veneering porcelain. However, there are no distinct findings about the bond between the veneering porcelain and the zirconia core and the wet ability rate of the zirconia core by the porcelain and the micromechanical bond between them are the only known mechanisms in this respect which make this bond weaker than the metal-ceramic bond (14). Therefore,
before applying the porcelain, core surface should be sandblasted according to the manufacturer’s instructions.

Surface treatment:
Regarding the effect of surface treatment on the physical properties of Y-TZP, Kosmac et al, in 1999 proposed sandblasting as an effective technique for improving the strength of Y-TZP in comparison with grinding in a clinical setting (15). Grinding may significantly decrease the strength and reliability of the zirconia components. On the other hand, Guess demonstrated that sandblasting with 100µm particles did not have a significant effect on the shear bond strength between zirconia and veneering porcelain in Cercon system in comparison with the systems that do not require sandblasting (16).

In the present study, core surface treatment was done from 10 mm distance with 50 micron particles at 3 bar pressure. In order to obtain better results, evaluation of the effect of sandblasting particles of different sizes (diameters) on the bond strength between the zirconia core and veneering porcelain in a specific system may be helpful.

Coefficient of Thermal Expansion (CTE):
Another point that should be addressed is the different CTEs of zirconia core and veneering porcelain which has been discussed in many studies and can affect the bond strength between these two. Table 2 summarizes the physical characteristics of the core material and veneering porcelain used in this study.

<table>
<thead>
<tr>
<th>Name of material</th>
<th>Manufacturing company</th>
<th>Young’s modulus (GPa)</th>
<th>CTE 20–500 ◦C(ppm/◦C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cercon Core</td>
<td>Degudent, GmbH, Hanau-Wolfgang, Germany</td>
<td>205</td>
<td>10.5</td>
</tr>
<tr>
<td>Cercon Ceram S</td>
<td>Degudent, GmbH, Hanau-Wolfgang, Germany</td>
<td>69</td>
<td>9.7</td>
</tr>
</tbody>
</table>

CTEs of the substructure and veneering porcelain should be compatible. If the CTE of the substructure material is greater than that of the porcelain tangential compressive stress develops causing cracks propagating parallel to the substructure surface. If the CTE of porcelain is greater than the substructure tangential tensile stress will develop causing cracks growing from the surface of substructure towards the free surface of the veneer. The latter results in porcelain flaking. Ideally, by selecting an appropriate substructure and porcelain material, cracks can be prevented (16).

According to the manufacturer, the difference in CTEs of the zirconia core and veneering porcelain used in this study was 0.8X10⁻⁶ K⁻¹. Although this difference in CTEs and its related stresses cannot disturb the bond in metal-ceramic restorations (due to the presence of strong chemical-micromechanical bond between the framework and porcelain), it may compromise the bond strength in all ceramic
crowns (where the nature of the bond between zirconia core and veneering porcelain is questionable).

Thermal Conductivity (TC):

The last but not least subject to consider is the thermal conductivity. Alloys have high TC (300 Wm⁻¹ K⁻¹) but zirconia cores act as a thermal insulator. According to the information provided by different manufacturers, TC of zirconia cores is about 2-2.2 Wm⁻¹ K⁻¹. TC of the veneering ceramics is also within the same range (2.39 Wm⁻¹ K⁻¹). Low total sum of the TCs of core and veneering porcelain results in a delay in losing the heat at the interface compared to metals and subsequent changes in the rate of porcelain and zirconia core’s linear contraction and development of thermal stresses at this area which per se may result in porcelain delamination in time. On the other hand, different core/veneer thickness ratios at different parts of the restoration can result in development of additional stress during thermal cycles of the porcelain firing process (18).

Conclusion:

This aim of this study was to evaluate the effect of zirconia core design on the in vitro fracture load of molar full ceramic crown. Considering the limitations of the present study, we can conclude that customized core design in comparison with the standard design significantly increased the fracture resistance of full ceramic crowns and failure of the restorations fabricated as such occurs in higher load levels.

References: