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The effect of zirconia thickness on the biaxial flexural strength of zirconia-ceramic bilayered discs

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The aim of this study was to assess the effect of zirconia core thickness on the biaxial flexural strength values of zirconia-porcelain bilayered discs. A total of 60 discs with 0.3, 0.4, and 0.5 mm thickness were obtained from a fully sintered zirconia block. A 1.5-mm thick layer of veneer porcelain was fired on the zirconia specimens and biaxial flexural strength tests were performed on the bilayered discs. In each group, the loading surface was the veneer porcelain in half of the specimens (core in tension) and the zirconia core surface in the other half (core in compression). The zirconia core thickness had no effect on the biaxial flexural strength of zirconia-porcelain bilayered discs when the core was in tension ($p>0.05$). Whereas, when the core was in compression, an increase in the zirconia core thickness resulted in an increase in the biaxial flexural strength ($p<0.05$).

Keywords: Biaxial flexural strength, Bilayered ceramic, Zirconia core thickness

INTRODUCTION

All-ceramic restorations have been used extensively in dental practice. Ceramic materials are resistant to compression but are brittle and cannot withstand high tensile stresses occurring under functional loading. Tensile strength is an important factor in the clinical success of all-ceramic dental restorations[1)]. Therefore, there are two main types of all-ceramic systems. The first system, in which reinforced glassy materials have been successfully used to fabricate single crowns for anterior and premolar regions, involves the use of a single material[2)]. In the second system, by using two different materials (bilayered), esthetic ceramics (such as porcelain and other glassy materials) are fused to frameworks composed of high-strength ceramics instead of alloys. Dense sintered polycrystalline zirconia-based materials are promising candidates for all-ceramic restoration frameworks[3,4)]. The most common method for fabricating a zirconia framework is the computer-assisted design/computer-assisted manufacturing (CAD/CAM) milling of a solid block.

The three main types of zirconia available for use in clinical dentistry are chemically identical with slightly different physical properties (*e.g.*, porosity, density, purity, and strength). The first type is the fully sintered or "hot isostatic pressed" zirconia; hot isostatic pressing (HIP) is a sintering technique used in the ceramic industry that simultaneously utilizes high temperature and pressure to increase the density of the material. The second is a partially sintered zirconia, and the third does not undergo sintering (non-HIP) and is called a "green state" zirconia; both the second and third types involve similar manufacturing and fabricating processes[5)]. Zirconia is the strongest and toughest ceramic used in dentistry, however its esthetics due to its translucency constitutes a major disadvantage. Particularly in case of placing a crown or shortspan bridge restoration in the anterior region in the presence of other natural anterior teeth, the opacity of zirconia becomes a considerable problem[6)].

Although fabrication of monolithic crowns using translucent zirconia may be an alternative in the posterior region recently, in the anterior region zirconia cores or frameworks should be veneered with porcelain to achieve acceptable esthetics[7)]. The failure of bilayered ceramic restoration is a very complex process, and research on layered ceramic structures demonstrates that a veneer of relatively weak porcelain may result in failure at low loads if the porcelain veneer is placed under tension[8-10)]. Although uniaxial strength tests, such as the three- or four-point bending of beams, have been used to determine the strength of dental ceramics in the past, most prosthetic applications experience a state of biaxial stress[10-12)].

The biaxial flexural strength tests are more useful than the uniaxial flexural strength tests because dental materials are generally subjected to multiaxial loading[13,14)]. The International Organization for Standardization (ISO) described the piston-on-three-ball test in the ISO 6872 for measuring the biaxial flexural strength of dental ceramics; however, this test

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has been limited to monolithic materials[15]. The strength of bilayered ceramic restorations may be compromised by the complex distribution of tensile stresses, and these restorations may fail at unexpectedly low stresses unless tensile stresses are considered in the structure design[12]. Hsueh *et al.*[16,17] derived an analytical formula from the existing equation on monolithic systems for calculating the stress distribution through the thickness of bilayered specimens that are tested in a number of biaxial flexural testing configurations. The results of the flexural strength tests of bilayered specimens can be considerably different depending on the testing configuration (core or porcelain material on the bottom surface) and addressed as core in tension or core in compression, respectively. Therefore, the clinical failure conditions expected at the connector area of fixed partial dentures can be simulated *in vitro* by loading the veneering porcelain in tension and the core material in compression[18].

In general, the manufacturers recommend a minimum zirconia core thickness of 0.3 mm for anterior teeth and 0.5 mm for posterior teeth[5]. Insufficient framework thickness has been mentioned as the main reason for zirconia core bulk fractures[19,20]. The failure types for the zirconia framework are divided into the following three categories: substructure failure, failure of the bond at the interface between the zirconia and the veneer porcelain, and breakage and chipping of the porcelain veneer[5]. White *et al.*[8] recommended constructing a bilayered ceramic restoration from the strongest core material available, in the greatest thickness possible, and with the thinnest veneer layer possible.

Fleming *et al.*[21,22] assessed the effect of the core/dentine thickness ratio on the biaxial flexural strength. Core thicknesses of 1 mm[21] and 1.5 mm[22] with core/dentine thickness ratios of 2:1, 1:1, and 1:2 were tested with either the core or the veneer porcelain loaded in tension. For a core thickness of 1 mm, the core/dentine thickness ratio failed to influence the biaxial flexural strength values[21]. However, for a core thickness of 1.5 mm, when the core/dentin thickness ratio was increased to 2:1, the biaxial flexural strength values increased[22]. Using 1-mm thick bar-shaped zirconia specimens, Lima *et al.*[23] reported that the thickness of the veneer ceramic influences the mechanical strength of the bilayered ceramic system.

The core thickness is important in achieving esthetics and zirconia is a semi-translucent material that is slightly more opaque than dentin[24], the level of opacity can be controlled by varying the core thickness[5]. A restoration with a core that is too thick or with connectors that are too large will be esthetically unpleasant because of the bulky appearance and may lead to periodontal failure over time due to increased plaque accumulation and poor hygiene[25]. Moreover, there may not be enough space for veneer porcelain[5].

A conservative approach involves minimal tooth preparation in the clinic, and a certain thickness of veneer porcelain is needed to adequately mask the color of an opaque porcelain material such as zirconia. However, in many studies, in which biaxial flexural strength tests were performed on zirconia cores veneered with porcelain, the thicknesses of the zirconia framework were higher than the thicknesses for clinical practice[21-23].

The aim of this study was to assess the effect of zirconia core thickness on the biaxial flexural strength values of zirconia-porcelain bilayered discs. Since, it has been well established for metal-ceramic systems, that the thickness of the veneering porcelain layer should not exceed 1.5–2 mm[26], biaxial flexural strengths as core in tension or compression were tested after application of 1.5-mm veneer porcelain on HIP zirconia cores, which have different thicknesses. The purpose of testing core in compression was to simulate the configuration of the connector area and pontic of a fixed partial denture. Additionally, the biaxial flexural strength values obtained using the ISO 6872[15] (for monolithic materials) were compared with those obtained from the equation developed by Hsueh *et al.*[16].

The null hypothesis of this study was that the zirconia core thickness does not affect biaxial flexural strength values of layered porcelain restorations under both core in tension (CIT) and core in compression (CIC) stresses.

MATERIALS AND METHODS

Preparation of specimens

Hint-Els HIP zirconia core blocks (Hint-Els DentaCad Systeme, Hint-Els, Griesheim, Germany) with a diameter of 16 mm were cut into 60 specimens of 0.3-mm thickness (n=20, Group 1), 0.4-mm thickness (n=20, Group 2), or 0.5-mm thickness (n=20, Group 3) using an Hahn and Kolb Type A1 industrial cutting tool (Hahn and Kolb, Stuttgart, Germany). Specimens from all groups were treated with a thin layer of Effect Bonder (VITA VM9, VITA Zahnfabrik, Bad Säckingen, Germany) according to the manufacturer's instructions and were fired in a porcelain firing oven (VITA Vacumat 40T, VITA Zahnfabrik). Subsequently, dentin porcelain (VITA VM9, VITA Zahnfabrik) with a 1.5-mm thickness was applied according to the manufacturer's instructions. Dentin porcelain powder was mixed with its special liquid to create slurry and was then transferred to a metallic mold on the core specimen. The mold was overfilled with the slurry and condensed on a vibrating table (Electro Vibrator, Porex, Renfert, Hilzingen, Germany) for 90 s; excess water appearing on the specimen's surface was blotted away with an absorbent tissue. To provide uniform thickness, the specimen's surface was leveled with a razor blade, and the material was fired. Grading was performed with a diamond porcelain grinding rotary instrument (Torr Dental DD2373 blue/medium grit, Febe Dental, Istanbul, Turkey) to obtain 16-mm specimens with a standard porcelain thickness of 1.5 mm. The final thicknesses of the specimens were measured with a micrometer (0–25 mm range, 0.01 mm graduation, 293 MDC-MX Lite, Mitutoyo, Tokyo, Japan) to ensure that they had equal thickness and opposing faces of the test pieces were flat and parallel to a tolerance within 0.05 mm in compliance with ISO 6872[15]. The specimens were kept in a desiccator to prevent uncontrolled environmental effects during storage periods.

Biaxial flexural strength tests

Disc specimens were centered and placed on three steel spheres (3.18 mm in diameter) positioned at 120º from each other on the perimeter of a 10-mm diameter circle. The biaxial flexural tests were conducted in a universal testing machine (Shimadzu Autograph AG-IS 50 kN, Shimadzu, Kyoto, Japan) with a 1-kN load cell at a crosshead speed of 1 mm/min until failure. A flat-ended loading cylinder, with a 0.7-mm radius, was used as the ram tip. To distribute the load evenly, a thin plastic sheet (0.08 mm in thickness) was placed between the ram tip and the specimen surface. All groups were divided into two subgroups. In each group, the overlying veneer porcelain was used as the loading surface in half of the specimens (CIT, Groups 1, 2, and 3; n=10), and the zirconia core surface was used in the other half (CIC, Groups 1, 2, and 3; n=10). The mechanical test was interrupted when the porcelain layer failed, and the load at the point of fracture (N) was recorded. The biaxial flexural strength values were determined using the following two equations:

First, the biaxial flexural strength values for monolithic materials (σ_{mon}) were calculated using the equation given in the ISO 6872[15]:

$$\sigma_{mon}=-0.238\,7P\,(X-Y)/b^2 \qquad \text{(eq.1)}$$

where σ_{mon} is the maximum center tensile stress [MPa], P is the total load causing fracture (N), and b is the specimen thickness (mm). X and Y were determined as follows:

$$X=(1+v)\ln(r_2/r_3)^2+[(1-v)/2](r_2/r_3)^2$$

$$Y=(1+v)[1+\ln(r_1/r_3)^2]+(1-v)(r_1/r_3)^2$$

where v is Poisson's ratio and is taken as the mean value between the Poisson's ratios of VM9 (v_1) and Y-TZP (v_2), r_1 is the radius of the support circle (mm), r_2 is the radius of the loaded area (mm), and r_3 is the radius of the specimen (mm).

Additionally, the biaxial flexural strength values for bilayered specimens were calculated using the equations developed by Hsueh *et al.*[16]:

$$\sigma_1=\frac{-E_1(z-z^*)P}{8\pi(1-v_1)D^*}\{1+2\ln(\frac{a}{c})+\frac{1-v}{1+v}[1-\frac{c^2}{2a^2}]\frac{a^2}{R^2}\}$$
$$\text{x}(for\ 0\leq z\leq t_1 \text{ and } r\leq c) \qquad \text{(eq.2)}$$

$$\sigma_2=\frac{-E_2(z-z^*)P}{8\pi(1-v_2)D^*}\{1+2\ln(\frac{a}{c})+\frac{1-v}{1+v}[1-\frac{c^2}{2a^2}]\frac{a^2}{R^2}\}$$
$$\text{x}(for\ t_1\leq z\leq t_1+t_2 \text{ and } r\leq c) \qquad \text{(eq.3)}$$

σ_1: Biaxial flexural strength value at the interface between the two layers of the bilayered specimens

σ_2: Biaxial flexural strength value at the bottom surface of the bottom layer in the bilayered specimens

[The values of the elastic modulus (E_1) and Poisson's ratio (v_1) for VM9 obtained from the manufacturer are equal to 64.57 GPa and 0.21, respectively. The values of the elastic modulus (E_2) and Poisson's ratio (v_2) for Y-TZP obtained from the literature are equal to 210 GPa and 0.23, respectively[27-30]. t_1=1.5 mm and t_2=(0.3 mm), (0.4 mm), (0.5 mm).]

P is the measured load required for a fracture; a, c, and R are the radii of the supporting ring, piston, and disc respectively; r is the radial distance from the center of the disc; z is the interface between layers in vertical cylindrical coordinates; t_1 is the thickness of the core layer; t_2 is the thickness of the veneer layer; v_1 and v_2 are the Poisson's ratio for the core and veneer materials, respectively; and z^*, D^*, and v are the physical meanings of the position of the neutral plane, the flexural rigidity, and the average Poisson's ratio of the multilayer, respectively, such that:

$$z^*=\frac{E_1t_1^2/2(1-v_1^2)+E_2t_2^2/2(1-v_2^2)+E_2t_1t_2/(1-v_2^2)}{E_1t_1/(1-v_1^2)+E_2t_2/(1-v_2^2)}$$

$$D^*=\frac{E_1t_1^3}{3(1-v_1^2)}+\frac{E_2t_2^3}{3(1-v_2^2)}+\frac{E_2t_1t_2(t_1+t_2)}{1-v_2^2}$$
$$-\frac{[E_1t_1^2/2(1-v_1^2)+E_2t_2^2/2(1-v_2^2)+E_2t_1t_2/(1-v_2^2)]^2}{E_1t_1/(1-v_1^2)+E_2t_2/(1-v_2^2)}$$

where E_1 and E_2 are the Young's modulus of the core and veneer layers, respectively.

$$v=v_1t_1+v_2t_2/t_1+t_2$$

A schematic drawing showing the stress distributions corresponding to the bilayered specimens is given in Fig. 1. The positive and negative values represent the tensile and compressive stresses, respectively. Compressive and tensile stresses, which induce strains on the disc, develop in perpendicular direction to the disc axis under bending loads. Although the maximum compressive stress is found at the uppermost edge of the disc, the maximum tensile stress is located at its lower edge. There is a point on the linear path where no bending stress exists known as the "neutral axis". Since only the flexural stress is considered in analytical solutions, the stress is linear through the thicknesses in each layer. However, as the two layers have different elastic properties, the stress is discontinuous at the interface and the stress gradients are different between the two layers[31]. As shown in Fig. 1, the zirconia core section of the plain specimen remains in the tensile region but is in the compressive region for the inverted specimen.

Although the minimum sample size was calculated as n=8 for each group based on power analysis of the data (Power:0.80, α:0.05, Δ:22 and SD:15 for biaxial flexural strength parameters), minimum sample size was adjusted as n=10 as recommended by ISO 6872[15].

After mechanical testing, the fractured surfaces of the specimens were examined by visual inspection under 8× magnification using an optical stereomicroscope (Zeiss Discovery V12, Carl Zeiss Microscopy, Jena, Germany) to determine the mode of failure.

Statistical analysis

The statistical analysis was performed using the NCSS (Number Cruncher Statistical System) 2007 and PASS (Power Analysis and Sample Size) 2008 Statistical Software (Utah, USA). One-way ANOVA with Tukey HSD as *post-hoc* test and student's t-test were used for intergroup comparisons of the parameters. A p value <0.05 was considered statistically significant. ICC (Intraclass Correlation Coefficient) was used to evaluate the consistency between the σ_{mon} and σ_2 values obtained in Groups 1, 2, and 3 for both CIT and CIC. ICC can range between −1 and +1, in the manner of the standard Pearson Product Moment Correlation. Cicchetti and Sparrow[32] classified levels of ICC, in terms of practical or clinical significance, as follows: <0.40=poor; 0.40–0.59=fair; 0.60–0.74=good; and 0.75 and above=excellent levels of agreement.

RESULTS

The σ_{mon} values were calculated using eq.1, and the σ_2 values were calculated using eq.3; no significant differences in the σ_{mon} and σ_2 values were found among the CIT groups (p>0.05). With respect to CIC groups, significant differences were observed in one-way ANOVA analysis for the σ_{mon} and the σ_2 values (p<0.05)

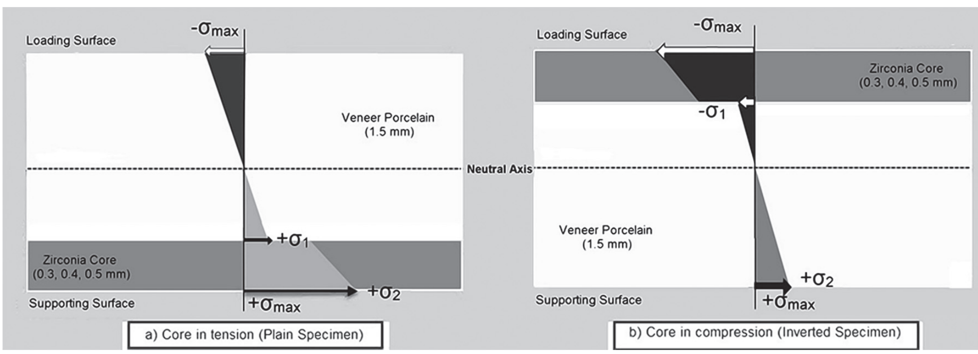

Fig. 1 Schematic drawing of the stress distributions in the bilayered test specimens.

Table 1 Comparison of the groups in relation to the σ_{mon}, σ_1, and σ_2 values [MPa]

			Group 1	Group 2	Group 3	^{a}p	^{b}*post-hoc*
σ_{mon}	CIT	Mean±SD Range	343.77±84.92 213.53–456.03	356.34±40.46 308.60–435.13	333.40±48.93 233.13–394.24	0.757	—
	CIC	Mean±SD Range	116.33±28.98 63.16–144.39	121.61±18.82 97.68–147.60	163.40±36.12 119.15–228.83	0.009*	G1, G2<G3
		^{c}p	0.001*	0.001*	0.001*		
σ_1	CIT	Mean±SD Range	92.30±22.80 57.33–122.44	68.80±7.81 59.58–84.01	44.85±6.58 31.36–53.04	0.001*	G1>G2>G3
	CIC	Mean±SD Range	103.93±25.89 56.42–129.00	78.13±12.09 62.76–94.82	73.15±16.17 53.34–102.44	0.010*	G1>G2,G3
		^{c}p	0.356	0.092	0.001*		
σ_2	CIT	Mean±SD Range	541.04±133.65 336.05–717.72	526.99±59.83 456.38–643.51	476.24±69.89 333.01–563.13	0.364	—
	CIC	Mean±SD Range	87.70±21.85 47.61–108.86	91.16±14.11 73.22–110.63	122.50±27.08 89.33–171.56	0.010*	G1, G2<G3
		^{c}p	0.001*	0.001*	0.001*		

CIT: Core in Tension CIC: Core in Compression G1: Group1; G2: Group2; G3: Group3
[a] One-way ANOVA, [b] Tukey HSD, [c] Student's t-test *p<0.05

(Table 1). Accordingly, in CIC groups, both the σ_{mon} and σ_2 values obtained in Group 3 were significantly higher than in Groups 1 and 2 (p<0.05), whereas no significant differences were observed between Groups 1 and 2 (p>0.05) (Table 1).

The σ_1 values calculated with eq.2 were compared, and significant differences were observed in both CIT and CIC groups (p<0.05) in one-way ANOVA analysis (Table 1). Tukey HSD test indicated that, for the CIT groups, the σ_1 values in Group 1 were significantly higher than those in Groups 2 and 3 (p<0.05). Additionally, the σ_1 values of Group 2 were significantly higher than those of Group 3 (p<0.05). However, for CIC groups, the mean σ_1 values of Group 1 were observed to be significantly higher than those of Groups 2 and 3 (p<0.05) (Table 1).

Student's t-test comparison of CIT and CIC groups with the same zirconia framework thickness showed significant differences in all of the paired groups (p<0.05) except the σ_1 values of Groups 1 and 2 (p>0.05) (Table 1). According to ICC, consistency between the σ_{mon} and σ_2 values of the groups was found to be excellent (Table 2).

Upon visual inspection of the specimens that were subjected to the flexural test, the following types of failure modes were identified: total fracture with delamination, total fracture without delamination, and

Table 2 Consistency of σ_{mon} and σ_2 values of the groups in the case of core in tension or compression

		n	ICC(95%CI) (σ_{mon} and σ_2)
CIT	Group 1	10	0.950 (0.752–0.990)
	Group 2	10	0.963 (0.814–0.993)
	Group 3	10	0.969 (0.845–0.994)
CIC	Group 1	10	0.980 (0.902–0.996)
	Group 2	10	0.980 (0.898–0.996)
	Group 3	10	0.980 (0.881–0.996)

CIT: Core in Tension, CIC: Core in Compression, ICC: Intraclass Correlation (Consistency)

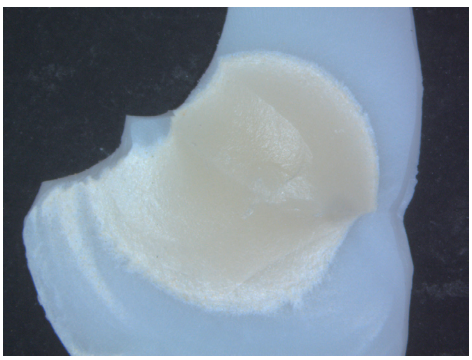

Fig. 2 Total fracture and a substantial level of partial delamination (8× magnification).

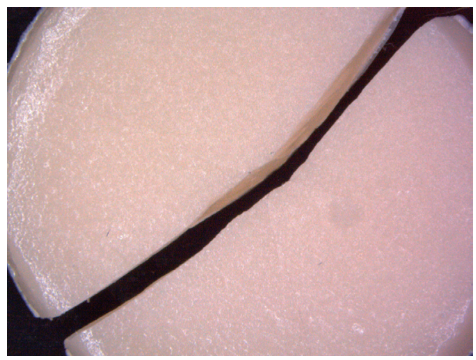

Fig. 3 Total fracture without delamination (8× magnification).

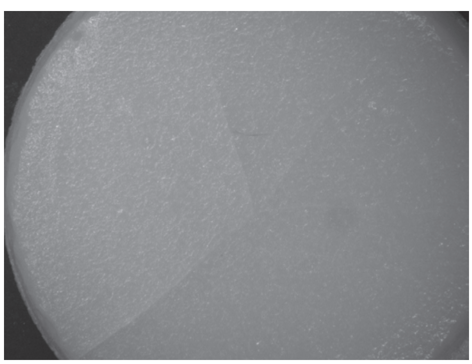

Fig. 4 Partial failure showing the cracks on the veneer with no exposure of the zirconia layer (8× magnification).

Table 3 Failure modes of the groups

Failure mode		Group 1	Group 2	Group 3
Total fracture with delamination	CIT	10	10	10
	CIC	—	1	—
Total fracture without delamination	CIT	—	—	—
	CIC	2	—	—
Partial failure	CIT	—	—	—
	CIC	8	9	10

CIT: Core in Tension, CIC: Core in Compression

partial failure (fracture of only one layer) (Figs 2, 3, and 4, respectively). Partial failure consists of cracks on the veneer with no exposure of the zirconia layer. Total fracture with delamination was observed in all of the specimens of the CIT groups. In the CIC groups, two specimens from Group 1 had total fracture without delamination, and eight specimens had partial failure. Additionally, one specimen from Group 2 showed total fracture with delamination, and nine specimens showed partial failure. Partial failure was observed in all Group 3 specimens (Table 3).

DISCUSSION

Chipping or fracturing of veneer porcelain occurs at a relatively higher rate for zirconia-based fixed dental prostheses than for conventional porcelain-fused-to-metal systems, as observed in recent clinical studies[33]. Strength is one of the most important mechanical parameters for understanding the clinical potential and limitations of dental ceramics[13,14]. This study evaluated the effect of zirconia core thickness on the biaxial flexural strength values of zirconia-porcelain bilayered discs. The null hypothesis of this study was partially rejected. No significant differences were found among the mean biaxial flexural strength values obtained for the 0.3-, 0.4-, and 0.5-mm zirconia cores in the CIT groups. In contrast, in the CIC groups, the mean biaxial flexural strength values for the 0.5-mm core group were calculated as significantly higher than those for the 0.3- and 0.4-mm core groups.

Because dental ceramics are brittle and have limited ability to absorb elastic energy, tensile stresses and structural flaws may cause premature failure even under low functional stresses[34]. In many studies, the material subjected to tension during flexural testing and the interaction between the materials in ceramic structures have had a significant effect on the strength and fracture mode of the layered structures[7,8,12]. The fracture strength values for the layered structures tend to be similar to those for monolithic specimens constructed with framework material when the framework material is under tension[10]. However, when the porcelain is subjected to tension, the framework material has a small effect on the fracture strength of these structures[35].

This present study showed that the core material had an influence on the biaxial flexural strength when the veneer porcelain was subjected to tension. The biaxial flexural strength values were calculated using eq.1 (σ_{mon}), eq.2 (σ_1), and eq.3 (σ_2). The σ_1 values were not compared with the σ_{mon} values because these values correspond to the interface between the two layers. The biaxial flexural strength values (σ_{mon} and σ_2), which represent the tensile stresses, were compared and found to be higher in CIT and lower in CIC groups. In addition, consistency between the σ_{mon} and σ_2 values of groups was found to be excellent. The values showed similar variations with regard to zirconia core thicknesses (0.3, 0.4, and 0.5 mm) and stresses (tensile or compressive). Consequently, both equations were found to be appropriate for the calculation of biaxial flexural strength.

In a study by Fleming *et al.*[22], the fracture resistance increased as the core/dentin thickness ratio was increased when the core layer was tested in tension. Additionally, when the reinforcing core layer was tested in compression (and the weaker veneer layer in tension), the fracture resistance increased with the core/dentin thickness ratio. In the same study, the core thicknesses (core/dentin height) were 1.5/0.75 mm, 1.5/1.5 mm, and 1.5/3.0 mm. It was found that the core/dentin thickness ratio influenced the biaxial flexural strength and flexural strength data reliability when both the reinforcing core and veneer dentin porcelain were tested in tension. The strength and reliability increased for a core/dentin thickness of 1.5/0.75 mm. However, in our study, lower core thicknesses (0.3, 0.4, and 0.5 mm) were used, and the veneer porcelain thickness of 1.5 mm was the same for all specimens. Accordingly, in the CIT groups, no differences were observed between the 0.3/1.5 mm, 0.4/1.5 mm, and 0.5/1.5 mm core/dentin thickness ratios.

In all studies performed on bilayered specimens, the strength, reliability, and mode of fracture were mainly determined by the material on the bottom surface under biaxial tensile stress[7,36]. Therefore, undersurfaces of the connectors of fixed dental prostheses and other areas of high tensile stress should not be veneered with porcelain[7,35]. For this reason, the effect of core thickness on the biaxial flexural strength values were investigated in our study when core was in compression or the veneer porcelain was in tension as for the connector areas. An increase in the core thickness resulted in an increase in biaxial flexural strength values that were calculated with two different equations. Although the veneer porcelain thickness was the same in all groups, the group with the 0.5-mm zirconia core thickness had higher biaxial flexural strength values than the groups with the 0.3 mm and 0.4 mm zirconia core thickness. White *et al.*[8] reported that the strongest available core material should be used in the greatest thickness possible, with minimal coverage, and the weaker veneer layer needs to be as thin as possible.

The compressive stresses compensate for the tensile stresses accumulated inside the porcelain, but under tensile functional loads, failure may occur at lower strength values[37]. Considering the core and veneer porcelain thickness in the core in tension, as the σ_2 and σ_1 are both positive (both tensile), the bottom surface of the core and the interface between the core and the veneer porcelain are influenced by tensile stresses (Fig. 1). Thus, as σ_1 is tensile, fracture with delamination was observed in all CIT groups. When the core was in tension, failure of the bilayered disc is governed by the low tensile strength of the veneer. The results of the present study indicated that the tensile stress on the bottom surface of the veneer porcelain (tension at the interface, σ_1) decreased as the core thickness increased, in the CIT groups; however, there were no considerable changes at the bottom surface of the core. Both ceramic layers were under the influence of tensile stresses. As a result, as previously reported in some studies[10] bilayered disc behaved as a monolithic material since the core was in tension and total fracture was observed in all groups (Fig. 2). In the CIC groups, as σ_1 is compressive and σ_2 is tensile in nature, the whole core layer and the interface are influenced by compressive stresses ($-\sigma_1$). In this respect, the compressive stresses decreased when the core thickness increased. However, the bottom surface of the veneer porcelain is affected by tensile stresses ($+\sigma_2$) (Fig. 1). Thus, partial failure was observed in all CIC groups except for with three specimens (Fig. 4), where the core resisted compressive stresses and did not fracture such that cracking of the veneer porcelain was a result of the tensile stresses.

Ceramic materials are strong under compression; however, they are brittle under tension and are unable to withstand high tensile stresses. Lawn *et al.*[1] observed that the first damage in bilayered structures with a thicker core layer (>1 mm) is initiated from the ceramic top surface as the cone cracks, whilst in the thinner core layers (<1 mm) as radial cracks first. Similarly, Fleming *et al.*[22,38,39] examined bilayered structures with thicker ceramic layers (=1.5 mm) in which cone cracks in the top surface were evident. For bilayered, disc-shaped specimens, the fractures were identified as radial cracks initiating from the ceramic/substrate interface[40]. In a subsequent study by Fleming *et al.*[21], it was concluded that the design criteria for all-ceramic crown and bridge structures need to be optimized to ensure the acceptance of the high strength ceramic systems amongst practitioners. The core to dentine thickness ratio was identified as influencing the fracture resistance and fracture mode and failure origin of bilayered ceramic composite specimens.

Although valuable information can be obtained from laboratory studies, care should be taken not to over emphasize the results of *in vitro* investigations such as those reported in the present study. Numerous variables play a role in the immediate and long-term behavior of layered ceramic restorations, and they all deserve careful consideration. The present study has several limitations, making it difficult to compare results directly with clinical situations. First, dental ceramics are susceptible to the effects of thermal and chemical fatigue as well as to the effects of mechanical stresses in oral environment. Further research might be needed on the effect of chemical and/or thermal fatigue of the core and veneering materials tested in this study by considering aqueous environment of the mouth. Secondly, differences in thermal expansion coefficients between core and ceramic, firing shrinkage of ceramic, flaws on veneering, and poor wetting by veneering on core may also be regarded as other factors influencing veneer cracking[41] and need to be tested under *in vitro* laboratory circumstances. In the study also only one veneering porcelain thickness (1.5 mm) was used. The possible adverse effect of decreasing veneer thickness on each specific core material was not investigated. Therefore, the information derived from this study needs to be verified by future clinical studies.

CONCLUSION

Within the limitations of this study the following conclusions can be drawn:
The zirconia core material thickness had no effect on the biaxial flexural strength values of zirconia-porcelain bilayered discs when the core was in tension. However, when the core was in compression and veneer porcelain was subjected to tensile forces, an increase in the zirconia core thickness resulted in an increase in the biaxial flexural strength of bilayered discs. On the other hand, as the core thickness increased, the stress at the interface decreased in both circumstances no matter the core was in tension or compression.

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