

INFLUENCE OF CORE AND VENEER THICKNESS
RATIO AND VENEER THICKNESS ON FLEXURAL
STRENGTH OF LITHIUM DISILICATE.

Presented by Wagdi Hemali

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Abstract

Statement of the problem:

There is no clear guideline to the amount of reduction of the lithium disilicate core and its strength value as core/veneer thickness ratios changes.

Purpose:

The purpose of this study is to investigate the effects of different core/veneer thickness ratios and veneer thickness on the flexural strength of lithium disilicate.

Material and Methods:

Eighty-four disc-shaped IPS e.max Press (11.5mm diameter) were produced by investing and hot pressing according to manufacturer's recommendations, and divided to two main groups of 12 samples each. Set (A) comprised four groups made into four different core thicknesses (1.5mm, 1.25mm, 1.0mm and 0.8mm). Set (B) included three groups made into the same core thickness (0.8mm). Group 1 did not have a veneer layer, groups 2, 3, 4, 5, 6 and 7 were veneered with IPS e.max Ceram Dentin in layers of different thicknesses (0.25mm, 0.5mm, 0.7mm, 1.0mm, 1.5mm and 2.0mm respectively). The specimen's final thicknesses were adjusted and glazing procedures were performed onto the top surface of all specimens. All

specimens were then immersed in artificial saliva for two weeks prior to testing. Biaxial flexural strength was measured using a piston-on-three-ball test carried out in a universal testing machine (Model 4204, Instron Corb.Canton, MA) with a crosshead speed of 0.5 mm/minute.

Results:

The mean strength and standard deviation for the studied groups were as follows (MPa \pm SD). Group 1 exhibited the highest mean biaxial flexural strength value among all groups (266.96 ± 35.13), followed by group 2 (254.24 ± 37.89) than group 6 (253.81 ± 63.97). Group 3 and 5 exhibited similar mean flexural strength (231.53 ± 36.90) and (231.77 ± 40.55) respectively, followed by group 4 (228.25 ± 18.37). Group 7 exhibited the lowest mean value among all groups (199.68 ± 42.65). The result of One-way analysis of variances test was statistically significant $P = .004$. The result of post- hoc test (Bonferroni) indicated that there was statistically significant between group (6) and group (7).

Conclusion:

Within the limitations of this study, it can be concluded that different core/veneer ratios seem to affect on the flexural strength of lithium disilicate specimens but is not statistically significant.

Furthermore, there is no statistical evidence that veneer layer thickness has an effect on a fixed core thickness of 0.8mm. However, when a 2.0mm veneer thickness was applied to a core, the mean biaxial flexural strength was significantly lower.

Table of Contents

List of Tables	viii
List of Figures	viii
Introduction	1
Literature Review	2
History of Dental Ceramic	2
Porcelain Jacket Crowns	3
Dental Feldspathic Porcelain & Metal Ceramic	4
All Ceramic Systems	7
1) Aluminous Core Crowns	7
2) Zirconia Core Crowns	9
3) Glass Core Crowns	10
4) Pressing vs. CAD/CAM	13
Esthetics of All-Ceramic Crowns	15
Failure of All-Ceramic Crowns	16
Aim	17
Hypothesis	17
Clinical Significance	17

Materials and Methods	18
Core Ceramic Preparation	18
Veneering Ceramic Preparation	20
Samples Testing	21
Sample Size Calculation	22
Statistical Analysis	23
Results	24
Discussion	25
Limitations	32
Conclusion	32
References	42

Figure 7: Center- finding device.	36
Figure 8 A specimen was loaded into Instron machine. (Model 4204, Instron Corb.Canton, MA)	36
Figure 9: (A) A specimen of two pieces fracture, (B) A specimen to three pieces fracture.	37
Figure 10: (A) Fractured specimen from group 7, (B) Same specimen after the broken piece replaced.	37
Figure 11: Side view of specimen in group 7.	38
Figure 12: (A) Fractured specimen in group 6, (B) Side view.	38
Figure 13: Box plot graph for all groups.	41

Introduction

The demand for esthetic dentistry has become a standard in today's world. This has led to the development and advancement of highly esthetic dental materials. In an attempt to try to mimic the optical properties of teeth, all-ceramic restorations were created. All-ceramic restorations are made with cores of glass ceramics, aluminum oxide, and zirconium oxide.^{1,2} Lithium disilicate (IPS e-max Press, Ivoclar Vivadent, Schaan, Liechtenstein), a pressable glass-ceramic, can be used either as a simple monolithic material, or when surface characterization is necessary the material can be cut back and veneering fluoroapatite ceramic is applied (IPS e-max Ceram).³ Bilayered ceramic crowns composed of a strong core veneered with a weaker surface porcelain usually experience chipping, fracture or delamination of the veneering porcelain.^{4,5,6} The thickness ratio of the ceramic core to the veneering porcelain is a major factor in controlling crack initiation and potential failure.⁷ It also plays an important role in the mechanical behavior of the restoration.^{8,9} Although it is desirable to increase the thickness of the ceramic core, it is important not to compromise either the esthetics of the crown by overcontouring, or the tooth by overpreparing.¹⁰ Unfortunately, the focus of the literature has been mostly on

the monolithic state of lithium disilicate; only a few studies have evaluated the multilayered state of the material. Thus, there is no clear study as to how much reduction should apply to the core, and how much will affect the strength of the material. It is important, therefore, to evaluate the material with different core-veneer thickness ratios.

Literature Review

History of Dental Ceramic

Porcelain's introduction to dentistry was in the form of porcelain denture teeth. In 1808, Guiseppangelo Fonzi introduced the first removable porcelain restorations, but they were not successful because of their brittle and opaque nature.^{11,12} The dental profession continued improving and mastering the art of ceramics, resulting in the development of the first fixed restorations at the end of the nineteenth century. In 1889, Dr. Charles H. Land invented a technique to fabricate porcelain jacket crowns.^{11,13} Around the same time, Jan Adriaansen invented the technique of building up porcelain with a brush. The universal use of metal-ceramics would not become practical until 1962, when M. Weinstein and S. Katz developed the use of gold alloys

reinforcement, to overcome the inherent brittleness of porcelain. Other methods of reinforcing porcelain were developed in 1963 by McLean and Hughes, and taken a stage further in 1976 by McLean and Sced, with alumina reinforced crowns and platinum bonded alumina crowns.¹⁴

Porcelain Jacket Crowns

Charles H. Land, first introduced his porcelain jacket crowns in 1903. Feldspathic porcelain, in a dry clay form, was applied with a wet brush in layers over a piece of thin platinum foil that had been burnished onto a die. Since platinum lacks the oxide layer required to bond to porcelain, the platinum was removed after firing.¹⁵ However, removing the platinum foil after the crown was fired meant that there was always a substantial gap at the margin. Also, the porcelain tended to be too weak to withstand occlusal forces and too opaque to match the surrounding teeth. In an attempt to strengthen the porcelain, a technique was developed in which a tin oxide coat was applied to the platinum foil, enabling a bond with the ceramic. However, this technique did not solve the problems with esthetics and strength.¹⁶

Dental Feldspathic Porcelain & Metal Ceramic

Dental porcelain started out essentially as domestic porcelain, which was a form of pottery. The primary elements in domestic porcelain are kaolin, quartz and feldspar.¹⁷ A ceramic body includes both a refractory skeletal structure and feldspathic glass (feldspar). The refractory structure exists in the form of crystals, which remain unmelted but are fused at their points of contact. These crystals retain their shape through the sintering and fusing processes. During the sintering process the porcelain crystals bond together, and as the temperature increases the feldspar glass steadily flows to infiltrate between the refractory particles.¹⁸ The first element of domestic porcelain is Kaolin, and it is opaque. Thus, by 1938, little or no kaolin was left in the feldspathic porcelains chosen for dental use, and for a long time dental porcelain was manufactured with feldspar glass and finely ground quartz. The second element, feldspar, is a naturally occurring glass. Feldspar contains silica, fluxes and alumina, all bonded together. The process of turning the crystalline silica, along with associated fluxes and modifiers, into glass is called vitrification. Vitrification is the development of a liquid phase, by reaction or melting, which on cooling provides the glass phase. Silica forms an amorphous glass structure when it is heated to its melting

point, 1700°C, and cooled fairly rapidly to prevent devitrification (crystallization). On very slow cooling from a liquid state, silica would produce a devitrified quartz structure. Thus, the structure of silica glass is similar to that of crystalline quartz, with the exception that it has an amorphous structure.^{19,20} Quartz crystalline inclusions have the unique property of remaining hidden within the glassy matrix due to the similarity of their respective refractive indexes.²¹ Aluminum oxide exists in dental porcelain in two forms. It is a component of feldspathic glass, and in this form is used as a stabilizer without affecting its optical properties. However, aluminum oxide also exists in the form of small crystals dispersed throughout the glass matrix. In this form it strengthens the glass by acting as crack stopper, but it also diffuses light and causes opacity. Thus, it was replaced with various stabilizers such as boron and dissolved alumina, but these did not entirely make up for the low level of the crystalline alumina in the substructure.²² Very fine particles of refractory quartz were left in the formulation in order to give the glass enough structure to resist slumping and crack propagation through the structure. Thus, feldspathic porcelains slowly began to be formulated with only a weak refractory skeleton composed mostly of quartz particles. Feldspathic porcelain has a flexural strength of approximately 80 MPa however, they are highly esthetic materials for

building tooth-like structures. Additional oxides have been added to the glass-forming matrix, such as potassium, sodium and calcium oxide, which are used as glass modifiers and act as fluxes.¹⁸ Fluxes causes crystalline structures to melt at lower temperatures by interrupting the integrity of the silica network. Individual manufacturers use specific amounts of metallic oxides, and processes which can be altered to effect changes in the properties of the porcelain, such as viscosity, melting range, chemical durability, thermal expansion and resistance to devitrification.²

The metal ceramic system was originally designed with the idea of producing restorations that combined the esthetic qualities of feldspathic porcelain and the strength of cast metal.²⁴ Fusing feldspathic porcelain to metal was not possible until the discovery of leucite.¹ Leucite was developed by heating Orthoclase feldspar with alkali metal carbonates such as K_2CO_3 , Li_2CO_3 to approximately $1,093^\circ C$ to form a glass, and a high-expansion ceramic phase identified as leucite ($K_2O, Al_2O_3, 4SiO_2$).¹⁸ Leucite crystals raising the coefficient of the thermal expansion of feldspathic porcelain to match the values of gold alloys.⁵ PFM (porcelain-fuse- to- metal) ceramic is similar in nature to the feldspathic porcelains, based on a silica network and potash feldspar or soda feldspar. However, its metal oxides content is higher

than metal free feldspathic porcelains. The increased metal oxide content in the opaque porcelain act as opacifiers to mask the underlying metal.^{23,18}

All Ceramic Systems

With all ceramic systems a wider range of translucency and opacity can be achieved, which provides a better esthetic result for patients than metal-ceramic systems. All-ceramic systems design is to replace the metal core with more esthetic core, these cores are aluminum oxide, zirconium oxide or glass ceramics, and can be process by (heat-pressing, sintering, slip-casting or milling).³

1) Aluminous Core Crowns

In 1965, John McLean reported on the strengthening of a feldspathic glass by the addition of aluminum oxide particles.²⁶ Alumina particles are far stronger than the glass matrix, and are more successful than quartz crystals in preventing crack propagation. The addition of small alumina crystals into the glass in concentrations of 40-50% increases the flexural

strength of the feldspathic porcelain on the order of about 2-3 times.²⁶ While the added alumina crystals created the refractory internal skeleton strong, it also reintroduced the problem of opacity. The solution was to cut back the facing of the aluminous crown and cover it with a veneer made of more esthetic feldspathic porcelains formulated without alumina particles. Thus, the aluminous porcelain was used as a substructure, over which a veneer could be applied. This substructure known as an aluminous core.²⁷ Newer aluminous porcelains have flexural strengths of between 400 and 700 Mpa, depending on the amount of alumina the structure contains. Higher value aluminous porcelain systems currently used for fabricating stronger cores include In-Ceram Alumina (Vita Zahnfabrik, Sackingen, Germany) and Procera (Nobel Biocare AB, Goteborg, Sweden).^{4,5} In-Ceram Alumina core material contains 85% alumina, fabricated through the slip-casting technique. Slip-casting accomplish by applying a slurry of densely packed aluminum oxide to a refractory die and sintered at 1120°C for 10 hours.⁶ This produces a porous skeleton of alumina particles, which is infiltrated with lanthanum glass in a second firing at 1100°C for 4 hours to eliminate porosity, increase strength, and limit potential crack propagation.⁷ In-Ceram is also available in blocks that can be milled from a pre-sintered block, in combination with CEREC (Sirona Dental Systems, Bensheim, Germany).

The crystalline phase consists of alumina (In-Ceram Alumina), alumina/zirconia (In-Ceram Zirconia), or an alumina/magnesia mixture (In-Ceram Spinell).⁸ Recent aluminous core, Procera, was developed with copings that contain 99.9% high purity aluminum oxide.⁹ The three-dimensional shape of the preparation requires using a sapphire contact probe to scan the working die.¹⁰ The data then is sent electronically to a manufacturing facility, where an enlarged model is copy-milled and used for the dry-pressing technique.^{28,29} High purity aluminum-oxide powder is mechanically compacted on the enlarged die and sintered at 1550° C, which results in eliminating porosity and returning the core to the dimensions of the working die.^{29,30} The crown form produced by In-Ceram or Procera is completed by veneering the framework with low-fusing feldspathic porcelain, matching the coefficient of thermal expansion of aluminum-oxide.²⁸

2) Zirconia Core Crowns

Zirconium oxide ceramics have no glassy components; all of the atoms are densely packed into regular arrays that are much more resistant to cracking than atoms in the less dense and irregular network found in glass

ceramic.¹ Zirconium oxide is transformed from one crystalline state to another during firing. At room temperature zirconia is monoclinic, and is tetragonal at firing temperature, with a monoclinic unit cell occupying about 4.4% more volume than when tetragonal. Yttrium-oxide is added to pure zirconia to control volume expansion, and to stabilize it in the tetragonal phase at room temperature.³¹ Y-TZP (Yttria stabilized Tetragonal Zirconia Polycrystals) can be manufactured through computer-aided design and computer-aided manufacturing (CAD/CAM) technology. An enlarged coping framework can be designed and milled from a homogenous ceramic soft green body blank of zirconia.³² The framework structure has a linear shrinkage of 20-25% during sintering, until it reaches the desired final dimensions.^{33,34}

3) Glass Core Crowns

The development of methods for controlling over-crystallization in order to create high crystalline and uniform grain structure was discovered by accident in 1952 by S. D. Stookey. The technique of transforms ceramic to glass ceramic called nucleation.³⁵ Nucleation or Ceramming in glass ceramic will form stable center of long-range atomic order than is normally present in the original glass, which increase the strength and hardness. The

translation of a glass to a glass ceramic is essentially a two-stage process involving the formation of nuclei and the further growth of crystals upon these nuclei, and is accomplished by first heating the glass to a temperature appropriate to the formation of nuclei, which may be some 20°C to 80°C higher than the annealing point.³⁶ Annealing involves keeping the glass near transition temperature for an extended time so that the atoms in the glass are rearranged enough to relieve stress. After most of the stress has been eliminated, the glass is finally allowed to cool to room temperature. The time given for nucleation depends mainly on the viscosity-temperature relationship of the glass, and may vary from five minutes to several hours. After nucleation, the temperature raises slowly to the crystal growth temperature. The rate of heating must be sufficiently slow to allow the developing crystals to form a skeleton. The glass ceramic is then allowed to cool slowly to prevent cracking.^{37,38}

Glass ceramics are composed of a glass matrix (feldspar) and crystalline (crystal filler). Feldspar is similar to feldspathic ceramic, which is based on silica and alumina (alumino-silicate glass). The crystalline usually exists in the form of particles shaped like needles or plates, formed during the nucleation process. This crystalline serves to interrupt the propagation of cracks in the material, which results in increased strength and toughness.¹⁸ A

material called Dicor (Dentsply/Caulk, Milford, DE), the first commercial glass-ceramic available for fixed restoration, contained 55vol% tetrasilicic fluormica crystals in the glass matrix.³⁹ The Dicor core was fabricated through a casting technique; after casting the core was subjected to a special heat treatment that caused crystalline material (mica) to grow within the glass matrix.⁴⁰ However, due to its high failure rate, the material is no longer in use.^{41,18} High-leucite-containing IPS Empress (Ivoclar Vivadent, Schaan, Liechtenstein) is composed of 35% crystalline leucite added to the glass matrix. IPS Empress has been developed in both machinable CAD/CAM and pressable forms.^{42,43} These materials are ideally suited for inlay and onlay restorations, single anterior crowns and veneers. Lithium disilicate IPS e.max consists of approximately 70% needle-like lithium disilicate crystals embedded in a glassy matrix. The crystals created by adding Li₂O (lithium oxide) to the alumino-silicate glass using special nucleation and growth heating treatments. These crystals measure approximately 3 to 6 microns in length.⁴⁴ Similar to IPS Empress, IPS e.max can be processed either by lost-wax pressing or CAD/CAM milling fabrication techniques.⁴⁵ IPS e.max Press is composed of quartz, lithium dioxide, phosphor oxide, alumina, potassium oxide, and other components in the chemical formula (SiO₂, Li₂O, K₂O, MgO, ZnO, Al₂O₃, P₂O₅).⁴⁴ The pressable form (IPS e.max Press) is

produced using a bulk casting production process to create the ingots. This involves a continuous manufacturing process based on glass technology (melting, cooling, simultaneous nucleation of two different crystals, and growth of crystals), which is constantly optimized to prevent defects. This composition produces a highly thermal, shock-resistant glass ceramic, as a result of the low thermal expansion that occurs when it is processed. Due to the shape and volume of the crystals, the flexural strength and fracture toughness of IPS e.max is approximately doubled compared to IPS Empress. A mean biaxial flexural strength 440 ± 55 MPa was reported for IPS e.max Press, compared with 104 ± 23 to 175 ± 32 MPa for IPS Empress.⁴⁶ The interlocked microstructure and layered crystals are also likely to contribute to increase strength of IPS e.max compared to IPS Empress. Moreover, the increased number of crystals and their orientation make cracks more difficult to propagate.^{47,48,49}

4) Pressing vs. CAD/CAM

The pressing method involves forming an accurate crown shaped wax pattern in a refractory mould, which is burnt out to allow a ceramic ingot made of crystalline particles distributed within a glassy material to be

inserted, under pressure from a pneumatic press furnace. The pressing method is among the most popular options, and has the advantage of reducing the likelihood of the formation of large flaws, and minimizing thermally induced residual stresses.^{50,51,52} Moreover, the pressing technique ensures that framework thickness and design are more easily controlled than alumina-zirconia infiltration systems.^{53,54,55} Alumina oxide and zirconium oxide ceramics are much tougher and stronger than glassy ceramics. Thus, these ceramics are more complicated to process into complex shapes than are glassy ceramics. Prostheses made from aluminum oxide or zirconium oxide ceramics were not convenient prior to the availability of CAD/CAM.^{9,56} The advantages of using CAD/CAM ceramics include reduced porosity levels and reduced patient appointments.¹⁸ However, chair-side or laboratory based CAD/CAM systems have factors that may affect or limit their accuracy, including software limitations in designing restorations, as well as hardware limitations in cameras, scanning equipment, and milling machines.⁵⁷ Systems dependent upon optical impressions experience problems with rounded edges due to the scanning resolution and positive error, which simulates peaks at the edges.⁵⁸ Other systems that use a surface contacting probe cannot accurately reproduce proximal retentive features less than 2.5mm wide or more than 0.5mm deep.⁵⁹

Esthetics of All-Ceramic Crowns

The primary advantage of using an all-ceramic restoration is increased translucency with improved esthetics. Although metal ceramics have esthetic limitation for restoration in the esthetic zone, they are the choice for posterior restoration where the occlusion forces are high.¹⁸ All ceramic systems offer the flexibility of leaving the margin at the gingival or supra-gingival level, which is difficult to achieve with metal-ceramic.^{60,3} Highly esthetic dental ceramics are feldspathic ceramic, however as the crystalline contents increase the ceramic become opaque and strength value will be increased.²⁵ Ceramic cores made of aluminum oxide or zirconium oxide are much stronger, but they tend to be relatively opaque compared to glassy ceramics.⁶¹ Heffernan et al⁶² studied the relative translucency of several ceramic material and found that In-Ceram Spinell, IPS Empress and lithium disilicate had higher levels of translucency, recommended for single crowns in the anterior region. Lithium disilicate was recommended for both anterior and posterior single crowns. These were followed by Procera, In-Ceram Alumina, In-Ceram Zirconia, which were only recommended to match opaque teeth, or in the posterior region.⁶³

Failure of All-Ceramic Crowns

Bilayered ceramic crowns compose of a strong core which is veneered with a weaker surface porcelain generally ranges between 60 and 120 Mpa, and usually experiences chipping, fracture or delamination of the veneering porcelain.^{64,65,66} It has been reported that veneered specimens generally have lower strength values than their non-veneered counterparts.⁶⁵ Delamination of In-Ceram Alumina has been observed in many in vitro studies.^{65,66,67} Cracks leading to fractures of veneers can severely compromise the esthetics and function of all-ceramic restorations.³² This has been a particular problem with zirconia-based ceramics. Chipping of veneered zirconia core restoration ranges from 8 to 50% at one to two years, while the reported rate of veneer fracture with metal-ceramic restoration has been between 4 and 10% after 10 years.^{68,69} However, specific changes in the core design are aimed at optimizing porcelain support, as well as reducing the size of fractures in the porcelain veneer, relative to a uniform thickness core for Y-TZP crowns.⁷⁰

Aim

The aim of this study is to investigate the effects of different lithium disilicate core and ceramic veneer thickness ratios on biaxial flexural strength with minimum thickness of 0.8mm.

Hypothesis

A higher ratio of core/veneer thickness results in greater strength. As core thickness increases and veneer thickness decreases, the flexural strength of the specimens increases. Additionally, when core thickness is held constant, a higher veneer thickness results in reduced flexural strength.

Clinical Significance

The results of this study may help to further understanding of the effects of the veneer layer thickness on the flexural strength of the restoration. It may also help provide a guideline for the optimal amount of core reduction without compromising the strength of the restoration.

Materials and Methods

Core Ceramic Preparation

Eighty-four disc-shaped (11.5mm diameter) wax patterns were fabricated in four different thicknesses (1.5mm, 1.25mm, 1.0mm, and 0.8mm) using a metal mold (Figure1). The metal mold was adjusted to different height; when set at zero a counter clockwise rotation to the first notch resulted in a 0.1mm increase in thickness. Each notch on the vertical component was equivalent to a 1.0mm increment. The disc height was adjusted for each group with an additional 1.0mm to their respected target thickness. After waxing was completed, the patterns were smoothed with a sand paper 600 grit using speed grinding polishing machine (Ecomet3, Buehler, USA, IL).

Wax thickness was measured using a digital caliper (Dentagauge1, Erskine Dental, Marina Del Rey, CA) sensitive up to 0.001mm. The wax patterns were than divided into two main groups (set A, and set B). Set (A) was subsequently divided into four groups 12 specimens each (1.5mm, 1.25mm, 1.0mm, and 0.8mm respectively) figure 2. Set (B) was also divided into three subgroups of 12 samples each made all with same core thickness (0.8mm).

Each waxed patterns was attached to a spruing wax (3mm in diameter X 5mm in length). The sprue patterns were attached to a base former and invested in a 200g silicone ring. A solution of 32ml IPS Press liquid and 22ml distilled water was added to 200g of IPS pressVEST speed powder (Ivoclar-Vivadent, Schaan, Liechtenstein) and vacuum mixed for 2.5 minutes (350 rpm) (figure 3).

The investment ring was left to set for 30 minutes on a table bench then placed in a preheated burn-out furnace (Ney Vulcan 3-130) at 850°C for 60 minutes. Following wax elimination, a cold ingot and an Alox plunger were placed into the investment ring, and immediately placed in a preheated furnace (EP 5000 Ivoclar Vivadent). After the press cycle was completed, the investment ring was removed from the furnace and allowed to cool at room temperature for 60 minutes. A separating disk (Dedeco international Inc, USA) was used to separate/break the investment ring. Rough divestment procedure was done using sandblasting (Renfert, Basic Master, USA, IL) with glass beads at 4 bar 60psi until the discs were visible, the pressure were then lowered to 2 bar 30psi for the fine divestment phase. A diamond disk (Brasseler USA, GA, 005412U0) was used to cut all sprues (Figure 4).

The specimens were then immersed in a special liquid (IPS e.max Press Invex liquid) and cleaned for 15 minutes in an ultrasonic cleaner to remove the reaction layer (Sandblasting with glass beads at 2psi were than preformed). Specimens were than steam cleaned and divided into seven groups (Table1). Thickness was measured in five different points, one in the center and 3mm in each direction (Figure 5).

Veneering Ceramic Preparation

Group 1 will not have a veneer layer. Groups (2, 3, 4, 5, 6 and 7) were veneered with different veneer layer thickness using IPS e.max Ceram (0.25mm, 0.5mm, 0.7mm, 1.0mm, 1.5mm and 2.0mm respectively) (Table1). The core ceramic discs were replaced in the same metal mold previously used to wax up, and their heights were adjusted based on their desired group thickness. The veneering ceramic space and an additional 0.3mm (to compensate for ceramic shrinkage) were filled with veneering ceramic powder mixed in IPS e.max Ceram buildup liquid until the metal mold was full (Figure 6).

The core discs with the veneering porcelain were removed from the metal mold and fired in a porcelain furnace (P300 furnace). Following cooling at room temperature, the discs final thickness was adjusted by sandpapering (600 grit) under water using a polishing machine. A digital caliper was used to measure the final thickness of all specimens at five different points, in the center and 3mm away in all four directions, to ensure uniform thickness. IPS e.max press glazing paste was then applied onto the veneered surface and the samples were fired according to manufacture recommended firing temperature. All specimens were immersed in artificial saliva (Caphosol, EUSA Pharma, NJ, USA) for two weeks prior to testing.

Samples Testing

A center finder device (ID-NR. 170593009, Preisser company, Germany) was used to located the center of each sample (Figure 7) Biaxial flexural strength was determined using piston-on-three- ball test carried in a universal testing machine (Model 4204, Instron Corb.Canton, MA) with a crosshead head speed of 0.5 mm/minute and 3,0mm piton diameter (Figure 8). The load was applied vertically with a compressive force until fracture. The veneered surface for each sample was facing the piston; as for group 1,

the glazed surface will be facing the piston. The sporting three balls were adjusted 8mm apart. The dependent variable (flexural strength) was measured in MPa and the independent variable (veneer thickness) was measured in millimeters.

Sample Size Calculation

A sample size calculation was conducted using nQuery Advisor (version 7.0). Assuming an effect size of $\Delta^2 = 0.50^*$, a sample size of $n = 12$ per group provides a Type I error rate of $\alpha = 0.05$ and power over 99%.

*Effect size obtained from pilot study.

Statistical Analysis

A histogram of the data for each group was used to check for equality of variances. One-way analysis of variance was used to compare means of all groups, followed by the post-hoc test (Bonferroni) to check for significant differences between groups. Specifically, Bonferroni test was used to compare all groups in set A to one another, and all groups in set B to one another. Means, standard deviations, and p-values were reported. Data for the groups was presented in a side-by-side box plot.

Results

The mean and standard deviation of flexural strength for the studied groups were as follows (MPa \pm SD) (Table 2). Group (1), exhibited the highest mean biaxial flexural strength value among all groups, 266.96 ± 35.13 MPa, followed by group (2), 254.24 ± 37.89 MPa, then group (6), 253.81 ± 63.97 MPa. Groups (3) and (5) exhibited similar mean values of 231.53 ± 36.90 MPa and 231.77 ± 40.55 MPa respectively. These were followed by group (4), 228.25 ± 18.37 MPa, and finally group (7), which exhibited the lowest mean value among all groups, 199.68 ± 42.65 MPa. The result of one-way analysis of variances test was statistically significant $P= 0.004$. There was no statistical difference between flexural strength values for groups 1, 2, 3 and 4 when analyzed using Bonferroni multiple comparison test (alpha level was set at $\alpha = 0.004$) (Table 3). Moreover, there was no statistically significant difference between flexural strength values for groups 4, 5 and 6. Group 7 was not statistically significant when compared with groups 4 and 5. However, there was a statistically significant difference between group 6 and 7 (Table 4).

Discussion

All-ceramic dental restorations have been used extensively for replacing missing teeth because of their superior esthetic appearance, durability and biocompatibility. Ceramic materials generally exhibit high compressive strength compared with their low tensile strength. Therefore, tensile strength is an important factor that influences the clinical success of dental restorations.⁷¹ Compressive occlusal forces produce tensile stresses along the inner surface of ceramic crowns, which can initiate crack propagation from surface flaws and porosities.⁷² Surface flaws are usually introduced during the fabrication process and maybe responsible for early restoration failures. It has been demonstrated that ceramic materials show considerable variation in strength, primarily due to their extreme sensitivity to the presence of cracks of different size. The unstable fracture of ceramic starts from critical flaws, and this phenomenon maybe explained by the “weakest link” theory, which states that fracture always propagates from the largest flaw favorably oriented to the tensile stress. For a given ceramic material, the distribution of crack size, shape, and orientation differs from sample to sample and its strength is statistically distributed according to the flaw size distribution.^{72,73}

Laboratory load-to-failure tests attempt to simulate clinical failure to investigate variables thought to influence the success of fixed prostheses and to assess new materials or designs.⁷⁴

Standard test methods for determining the flexural strength of ceramic material are either uniaxial such as three-or four-point bending of beams or Biaxial flexural test such as piston-on-ring, piston-on-three-ball, ball-on-ring, and ring-on-ring tests. Biaxial flexural strength tests have several advantages over uniaxial tests, as they produce stress and eliminate edge failures.⁷¹ It has been noticed that when dental ceramic bars are tested for uniaxial strength measurements, defects are commonly formed by bend processing. These defects are not found in clinical dental crowns, and they are also absent in disc specimens that are used for biaxial strength measurement.^{75,76} Thus, biaxial flexural test became the standard and the for testing dental ceramic material by the International Organization for Standardization ISO 6782 and American Society for Testing Materials ASTM 1499.^{77,78,79}

It has been agreed that crack propagation produced by tensile stresses, can cause a brittle ceramic to fracture. However, if the microstructure and surface quality are improved, then the performance of all ceramic dental restorations should improve.¹⁸ The fractured specimens in the present study were investigated using light microscopy in an attempt to understand the fracture behavior of the material. None of the broken samples failed adhesively, only cohesive failure was observed.

Most of the broken samples in set A and few in set B broke to two or three pieces (Figure 10). This has been observed in previous study by Gonzaga et al,⁸⁰ wherein the fracture patterns were similar to the present study. However, different fracture pattern than those mentioned previously were observed in some samples in set B (Figure 11). This fracture pattern could be due the presence of increased number and size of air bubbles, as this type of fracture was seen in samples with thicker veneer thickness. Although the size and the volume of the air bubbles, group 6 showed less air bubble in the suction surface than group 7 (Figure 12).

Since the veneering ceramic was applied in to the core using conventional technique with the brush, it is possible that as the veneer thickness increased

more air trapped and this air traps was hard to escape in the firing. This might explain the low value flexural strength that was obtained in group 7, however, the presented study was not aimed to look for mode of fracture. Fractographic analysis and scanning electron microscopy in future studies may help to assess to answer this assumption.

The finding of this study was in agreement with Cattell et al,⁸¹ the mean biaxial flexural strength for the monolithic specimens in this study was 266.96 ± 35.13 MPa, which is consistent with their reported mean of 265.5 ± 25.7 MPa. However, other studies reported a higher mean flexural strength range between (340 - 407 MPa).^{82,83,84} This difference may be attributed to test design.

A small piston tip diameter will result in a higher biaxial flexural strength because a smaller area of the specimen is subjected to the maximum tensile stresses.⁷⁴ It has been shown that a piston tip between 0.5mm and 3.0mm in diameter will match wear facets that are seen on fractured surfaces in clinical failure restoration.⁷⁴ Previous studies reported a low biaxial flexural strength for IPS- Empress was due to the piston tip diameters, which ranged from 1.3 to 1.6mm.^{85,86,87,88} However, a higher biaxial flexural strength for

IPS- Empress reported used piston tip of 0.75mm diameter.^{89,50} The piston diameter and used in the presented study is 3.0mm and this may explain why the mean biaxial flexural strength is less than that mentioned in previous studies.

Dental ceramics have been shown to be sensitive to the presence of water. Water will acts chemically at crack tips and causes hydrolytic rupture of Si-O-Si bond and weakening of the surface region of the glass phase ceramic material.⁹⁰ Since water exists in saliva, in order to mimic the oral environment samples in this study were stored in artificial saliva for two weeks. This may have played a role in the low mean reported in this study compared to these reported in the previous studies.^{81,82,83,84}

In the presented study, monolithic samples (group 1) that consist of 1.5mm core exhibited the highest flexural strength of 266.96 ± 35.13 MPa among all groups. When 0.25mm was reduced from the core (group 2) and replaced by 0.25mm of veneer ceramic, the mean flexural strength was reduced to 254.24 ± 37.89 MPa. However, the difference in mean flexural strength was not statically significant between these groups. A further reduction from the core of 0.5mm (group 3), which was replaced with 0.5mm veneer ceramic,

resulted in a decrease in mean flexural strength to 231.53 ± 36.90 MPa. This also was not statistically significant. More reduction of the core (group 4) and thicker veneer replacement of 0.7mm resulted in further decrease in mean flexural strength to 228.25 ± 18.37 MPa. The difference was not statistically significant. On the other hand, in set B when the core was held at a constant thickness of 0.8mm (in groups 4, 5, 6, and 7) and the veneer thickness was increased the mean flexural strength increased (as seen in groups 5, and 6). Group 4 was considered a part of set A as the sum of thickness for this group was 1.5mm, and a part of group 4 since it consists of 0.8mm core thickness. Group 5 specimens, consisting of 0.8mm core and 1.0mm veneer, showed higher mean flexural strength 231.77 ± 40.55 MPa than group 4 and similar to that in group 3. In group 6, the veneer thickness was increased to 1.5mm while maintaining a core thickness of 0.8mm, this group exhibited a mean flexural strength of 254.81 MPa, which was higher than groups 4, 5, and 7. However, the difference was only statistically significant with group 7, which has 0.8mm core thickness and 2.0mm veneer thickness. The mean flexural strength reported for group 7 was 199.68 ± 42.65 MPa, which indicates that an increase in the veneer thickness beyond 1.5mm with a fixed core thickness of 0.8mm will significantly effect the core strength. Unfortunately, the result of these groups cannot be compared

to other studies since none of the previous studies compared the affect of varying core/veneer thickness.

Because they are brittle and sensitive to crack propagation, although all-ceramic prostheses represent the most esthetically pleasing restoration, all-ceramic prostheses are the most fracture-prone prostheses.¹⁸ However, with adequate tooth reduction, an excellent quality impression, and a ceramic with reasonably high flexure strength (≥ 250 MPa) a high success rate can be achieved.¹⁸ The findings of this study indicate that a monolithic prostheses (groups 1) or a minimum veneering thickness (group 2), show the flexure strength more than 250MPa; this will ensure a high success rate for the prostheses. However, increasing the veneer thickness in groups, 3, and 4 will result in flexural strength less than the recommended 250MPa.

The findings of this in *vitro* study have to be confirmed by clinical study furthermore, to better simulate clinical situation such as, cyclic loading, anatomic crowns, as well as cemented crowns may be used in future studies.

Limitations

- The presented study was conducted in *vitro*.
- The test was performed on disc shape samples.
- No cyclic loading was performed.

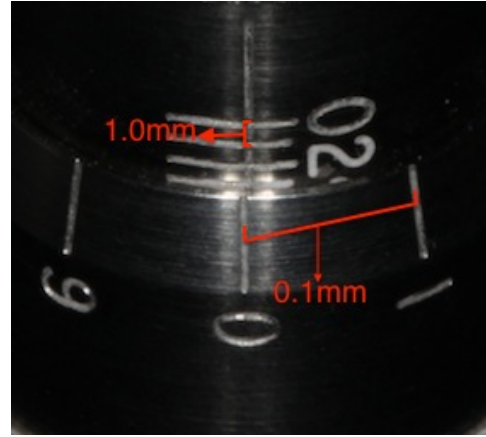
Conclusion

Within the limitations of this study, it can be concluded that different lithium disilicate core/ veneer ratios seem to have an effect on the flexural strength with minimum thickness of 0.8 mm lithium disilicate specimens but is not statistically significant.

Furthermore, there is no statistical evidence that ceramic veneer layer thickness has an effect on a fixed lithium disilicate core thickness of 0.8mm. However, when a 2.0mm ceramic veneer thickness was applied to a lithium disilicate core, the mean biaxial flexural strength was significantly lower.



A

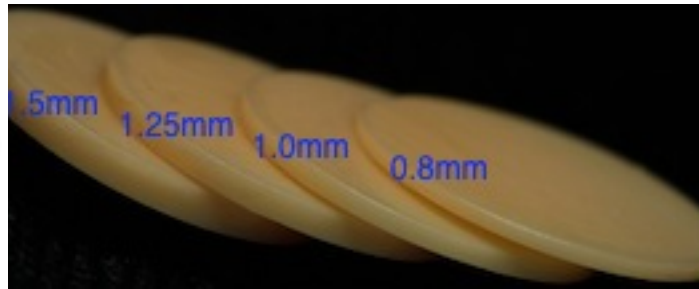


B

Figure 1: (A) Metal mold, (B) Metal mold measurement.

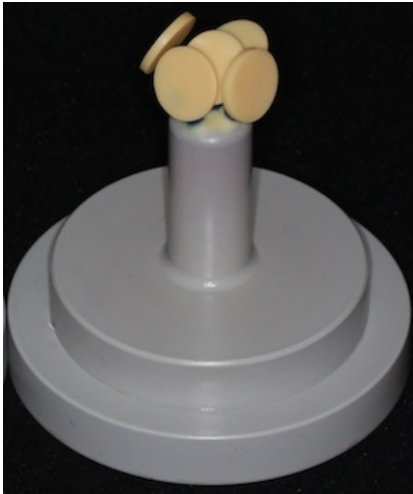


A



B

Figure 2: (A) Wax pattern in the metal mold,
(B) Different thickness of disc shape wax patterns.



A



B

Figure 3: (A) Wax patterns sprued to ring former, (B) Investment filled into the ring.



A



B

Figure 4: (A) Lithium disilicate specimens after divested, (B) Sprue separation.

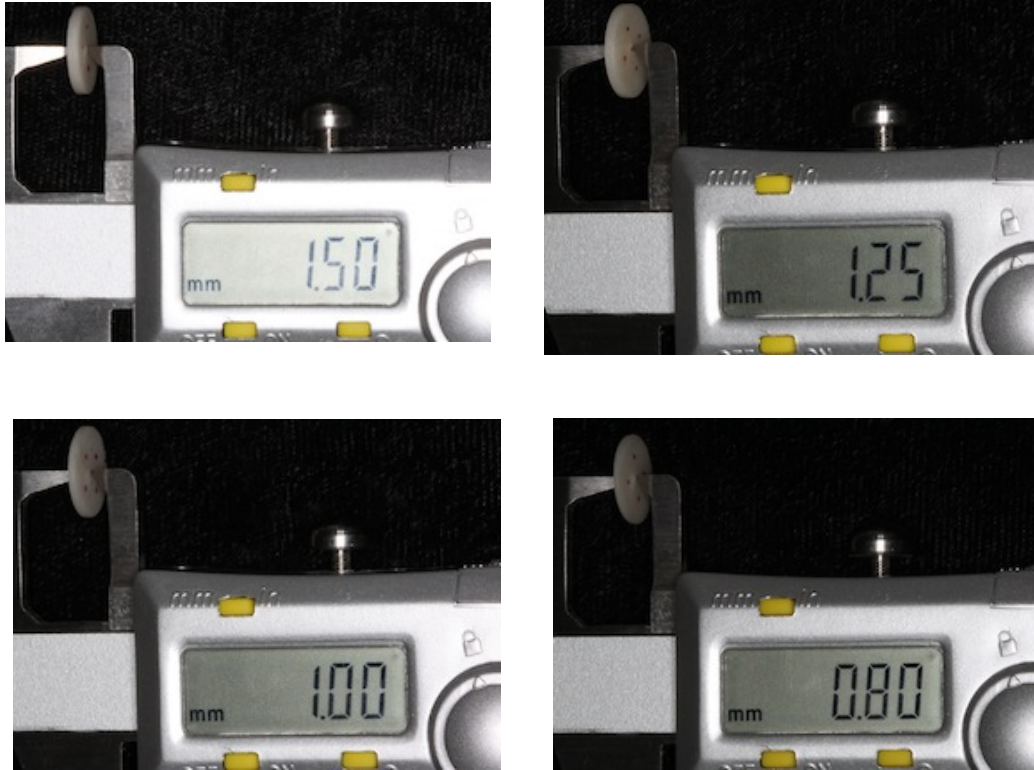


Figure 5: Thickness measurement of the different lithium disilicate specimen.

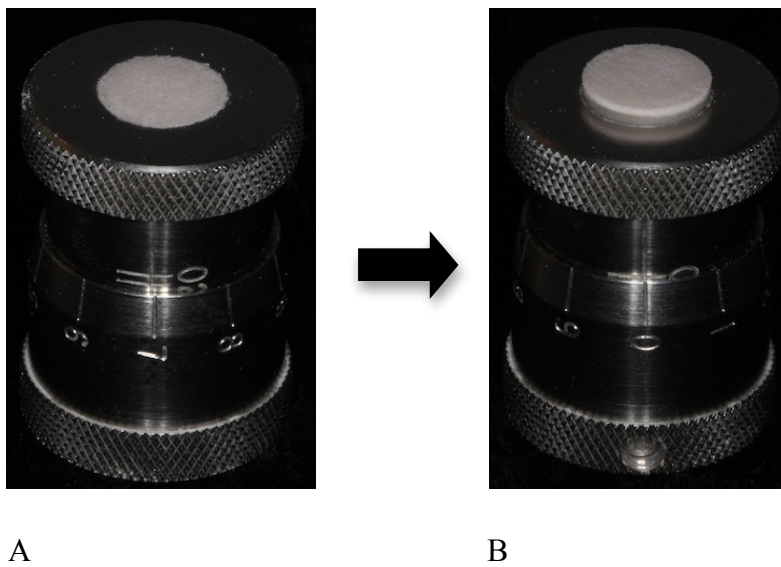


Figure 6: (A) The veneered ceramic is filled into lithium disilicate core on the metal mold, (B) Removing lithium disilicate core and veneer ceramic.



Figure 7: Center- finding device.

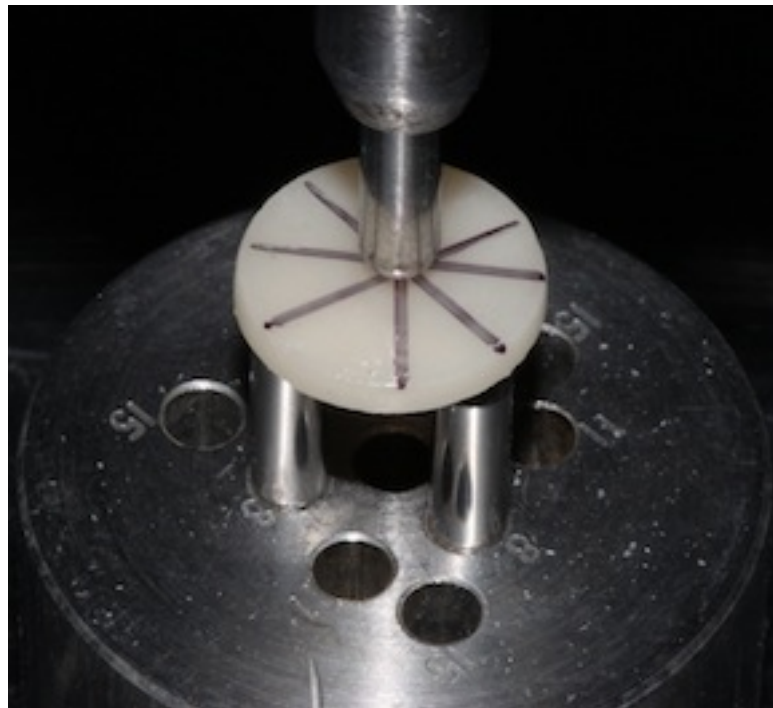
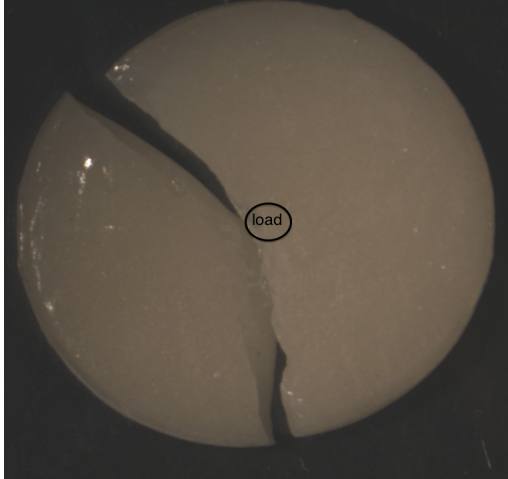
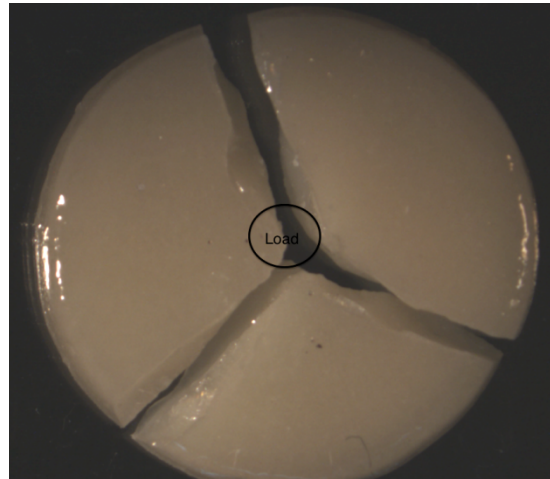


Figure 8 A specimen was loaded into Instron machine. (Model 4204, Instron
Corb.Canton, MA)

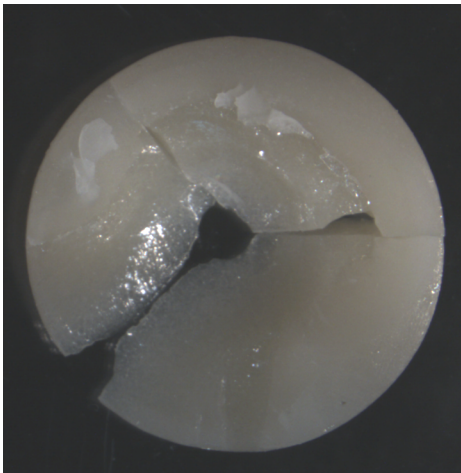


A

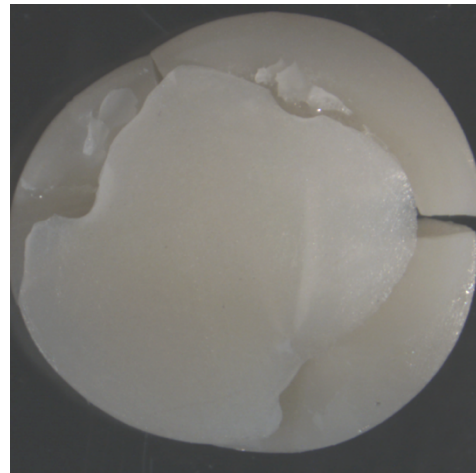


B

Figure 9: (A) A specimen of two pieces fracture, (B) A specimen to three pieces fracture.



A



B

Figure 10: (A) Fractured specimen from group 7, (B) Same specimen after the broken piece replaced.

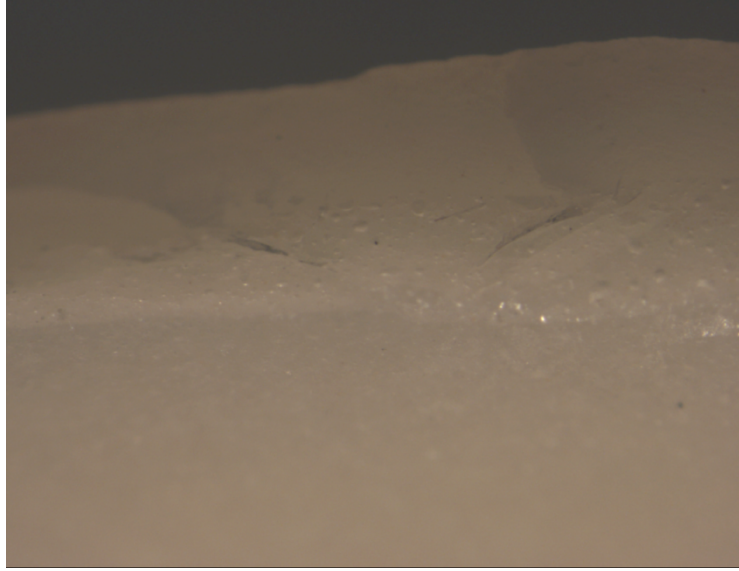
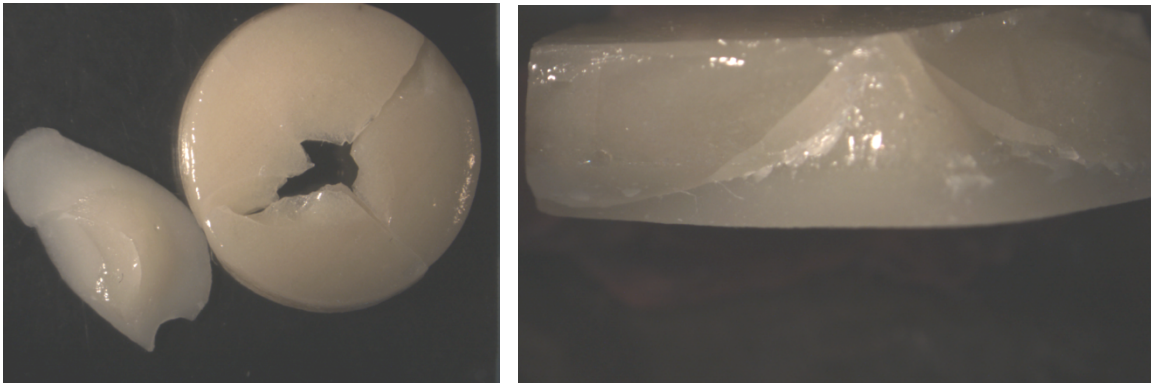


Figure 11: Side view of specimen in group 7.

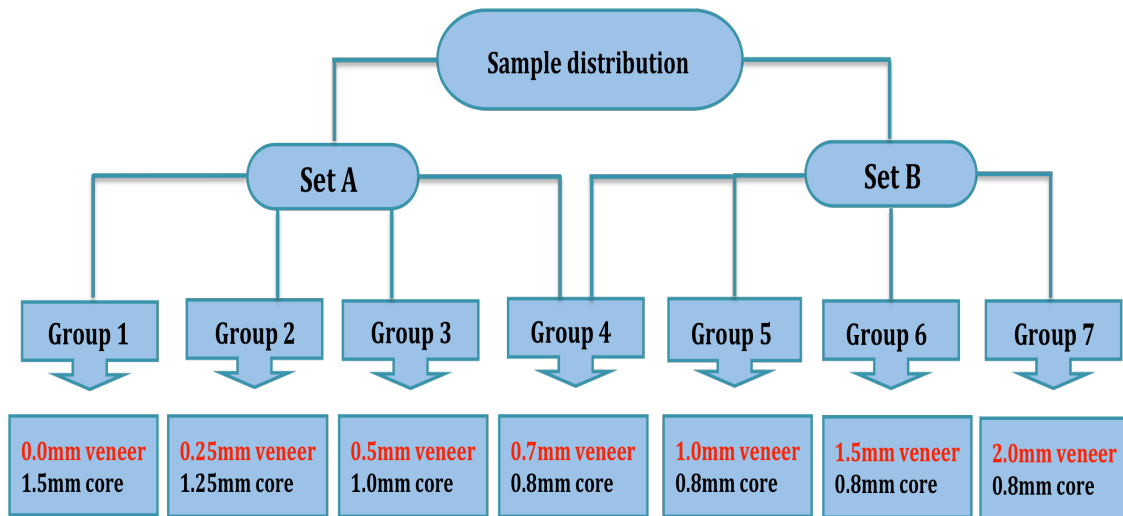


A

B

Figure 12: (A) Fractured specimen in group 6, (B) Side view.

Table 1: Group distribution.



Group	N	Veneer Thickness	Core Thickness	Total Thickness	Mean MPa	S.D
1	12	0mm	1.5mm	1.5mm	266.96	35.13
2	12	0.25mm	1.25mm	1.5mm	254.24	37.89
3	12	0.5mm	1.0mm	1.5mm	231.53	36.90
4	12	0.7mm	0.8mm	1.5mm	228.25	18.37
5	12	1.0mm	0.8mm	1.8mm	231.77	40.55
6	12	1.5mm	0.8mm	2.3mm	253.81	63.97
7	12	2.0mm	0.8mm	2.8mm	199.68	42.65
Total	84				238.04	44.87

Table 3 Bonferroni test result for set A					
Group		N	Mean difference (I-J)	S. E	P- value
I	J				
1	2	12	12.727	16.849	0.452
1	3	12	35.430	16.849	0.039
1	4	12	38.714	16.849	0.024
2	3	12	22.702	16.849	0.182
2	4	12	25.986	16.849	0.127
3	4	12	3.284	16.849	0.846

The mean difference is significant at the **0.004 level**

Table 4 Bonferroni test result for set B					
Group		N	Mean difference (I-J)	S. E	P- value
I	J				
4	5	12	-3.519	16.849	0.835
4	6	12	-25.558	16.849	0.133
4	7	12	28.565	16.849	0.094
5	6	12	-22.039	16.849	0.194
5	7	12	32.039	16.849	0.060
6	7	12	54.124*	16.849	0.002

The mean difference is significant at the **0.004 level**

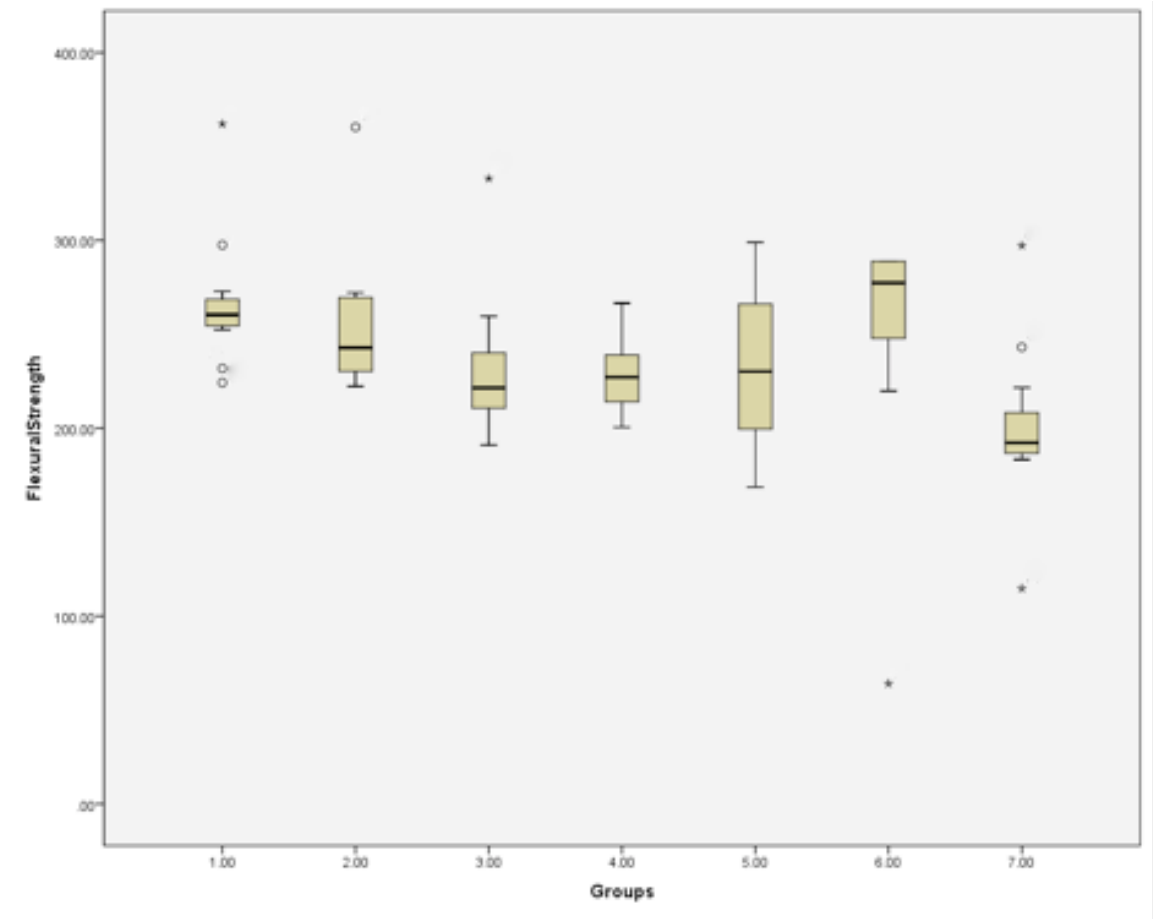


Figure 13: Box plot graph for all groups.

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