TRANSLUCENCY AND DEGREE OF CONVERSION OF

RESIN CEMENT WITH DIFFERENT THICKNESS OF

FULL CONTOUR ZIRCONIA

by

Noppamath Supornpun

Submitted to the Graduate Faculty of the School of
Dentistry in partial fulfillment of the requirements
for the degree of Master of Science in Dentistry,
Indiana University School of Dentistry, 2015.

Thesis accepted by the faculty of the Department of Prosthodontics, Indiana University School of Dentistry, in partial fulfillment of the requirements for the degree of Master of Science in Dentistry.

_____
David Brown

_____
Suteera Hovijitra

_____
Marco C. Bottino

_____
Tien-Min Gabriel Chu
Chair of the Research Committee

_____
John A. Levon
Program Director

_____
Date

ACKNOWLEDGMENTS

TABLE OF CONTENTS

LIST OF ILLUSTRATIONS

INTRODUCTION

In the past decade, all-ceramic crowns have been introduced to replace porcelain-fused-to-metal crowns because of their high aesthetics, excellent biocompatibility and sufficient flexural strength.[1-4] However, ceramic is brittle with low tensile strength and fracture toughness, potentiating cracks when subjected to stress. The most common complication with all-ceramic crowns is veneering fracture.[5] As a result, improvement in glass-ceramic processing techniques has continued. Current research aims to improve mechanical and optical properties by incorporating new materials, reducing particle size, and modifying the processing technique.[6,7] Among these materials, zirconia is a popular core material for all-ceramic crowns due to its high strength and toughness. However, given certain limitations observed in the use of conventional zirconia, an updated product called full contour zirconia has been developed with greater translucency.

OBJECTIVE

The overall objective of this study was to investigate the translucency parameter and degree of conversion of resin cement underneath several recently marketed full contour zirconia ceramics. The specific aims were:

1.	To investigate and compare the translucency parameter of recently marketed full contour zirconia (translucency zirconia), traditional zirconia, and lithium disilicate glass ceramic at different thickness.

2.	To evaluate the degree of conversion of the resin cement through different thicknesses of full contour zirconia, traditional zirconia and lithium disilicate glass ceramic.

HYPOTHESES

The null hypotheses of this study were:

1.   The full-contour zirconia disks have the same translucency as lithium disilicate glass ceramic.

2.   The full-contour zirconia disks have the same translucency as traditional zirconia.

3.   The resin cement underneath the full contour zirconia demonstrates the same degree of conversion compared with that of lithium disilicate glass ceramic after curing.

4.   The resin cement underneath the full contour zirconia demonstrates the same degree of conversion compared with that of traditional zirconia after curing.

The alternate hypotheses were:

1.   The full-contour zirconia disks have fewer translucencies than lithium disilicate glass ceramic.

2.   The full-contour zirconia disks have greater translucency than traditional zirconia.

3.   The resin cement underneath the full contour zirconium demonstrates a lesser degree of conversion than that of lithium dislocate glass ceramic after curing.

4.   The resin cement underneath the full contour zirconium demonstrates a greater degree of conversion than that of traditional zirconium after curing.

REVIEW OF LITERATURE

HISTORY OF CERAMIC

The term "ceramic" originated from the word "keramos" which means "pottery" in Greek. Since ceramic was developed, it has been utilized in daily human applications, such as stained-glass windows or potteries. In dentistry, it was first introduced to a single crown fabrication by Charles Land, a French dentist in 1789.[8] Since then, the glass-ceramic technology has been gradually improved in many industries, including in renewable engineering, medical technology and dentistry.

Modern glass-ceramics encompass both traditional and advanced glass-ceramics.[9] The traditional glass-ceramics are commonly derived from naturally raw materials such as clay minerals, quartz sands, and silicate glasses, which have been made into household products such as tableware, bricks and cements. The advanced glass-ceramics consist of carbides, oxides, nitrides, and non-silicate glasses (e.g. alumina or zirconia), which have been produced as electrical-thermal insulators, lightweight armors, aerospace frameworks, and biomimetic composites in the medical field.

DEFINITION AND COMPOSITION OF DENTAL CERAMIC

Theoretically, ceramic is defined as the reaction product between nonmetallic and metallic atoms, mainly with ionic and covalent bonds with the various proportions of crystal and amorphous phases. In dentistry, modern glass-ceramic fixed dental prostheses utilize the advantages derived from properties of crystalline ceramics in order to restore function and aesthetic, and maintain physical integrity of dental structure. Generally, this

material is brittle, strong in compression while weak in tension, and brittle. The various types of bonds between the atoms are responsible for brittleness and low ductility.[10] The glass ceramic also plays a critical role in oral rehabilitation while bridging the chasm between synthetic and naturalistic aesthetics. Therefore, importantly, it provides better aesthetic results relative to polymer and metal restorations. With these advantages, glass-ceramics are widely used for many dental applications.

Some ceramic materials consist of glass matrix and crystals. Glass is responsible for the optical quality, and crystals are responsible for the strength. The greater the glass matrix content, the higher the translucency; the larger number of crystals, the stronger and more opaque the ceramic. The properties of these ceramics vary depending on the proportion of crystals and glass content, type of bonding, crystal size, and processing technique.[10,11] However, the glass phase is the weaker part, responsible for crack propagation and restoration failure.[12]

CLASSIFICATION OF DENTAL CERAMIC

Ceramic dental prostheses can be classified according to either fabricating techniques or microstructural phases.[10,13] From fabricating techniques, ceramic can be classified as follows: 1) powder-liquid condensation, 2) glass infiltration, 3) heat-pressed, and 4) CAD-CAM machined.[14,15] According to their microstructural phases ceramic can be classified into four groups: 1) predominantly glass-based, 2) glassy-crystalline, 3) interpenetrating phase, and 4) polycrystalline.[16]

Fabricating Technique[10,13]

Powder-liquid condensation

Powder-liquid condensation is the simplest and most economical method for layering and veneering dental porcelain. The technique relies upon the technical artistry and experiences of the dentists. First, glass-ceramic powders are mixed with diluting agent. The slurry is applied layer by layer using a sculpturing blade or brush in order to craft the tooth anatomy. Then, the stacked porcelain is dried and heated. The stacked porcelain usually contains feldspar-based silicate glass with minimal crystalline fillers, which create a set of excellent aesthetics for natural custom veneers. Nevertheless, the porosity from the manually stacked porcelain shows a high degree of variability, which can affect the strength and the toughness of the prostheses.

Glass infiltration

Glass infiltration processes use both ceramic slips and glasses in a two-stage heat-treatment. The slips are a liquid suspension of ceramic particles and behave like hydrocolloids. They are poured into a mold, which is designed to absorb water. After water sorption, a thin coating of the ceramic particles is condensed tightly against the mold. After the first thermal treatment, sintering of the ceramic particles takes place and creates a porous microstructure. During the second firing process, in which the molten glass penetrates into the porous framework, the ceramic skeleton is surrounded by the glassy matrix to form the core of the dental prosthesis. Finally, feldspathic porcelain is stacked and glazed onto the glass-ceramic core for its final finish. The glass-infiltrated ceramic cores exhibit higher fracture resistance and strength than those fabricated by

powder-liquid condensation due to the high polycrystalline contents of core material and less manual interference.

Heat-pressed technique

The heat-pressed process is similar to the lost-wax casting method, consisting of designing, investing, burnout, and casting (pressing). In the designing stage, a wax model of the desired fixed dental prosthesis (FDP) is sculptured. Then, the mold is made of gypsum materials, which are heated, burning out wax and leaving behind only a cavity. Finally, molten glass-ceramic is pressed or injected into the mold's cavity.

Computer-aided design and computer-aided
manufacturing (CAD-CAM)

The CAD-CAM technique is the newest generation of glass-ceramic using machine-able blocks. The machine-able blocks are designed to allow ease of milling and to maximize cutting efficiency. In this partially crystallizing state, an ingot exhibits mild to moderate strength and hardness, which can be easily milled by a CAD-CAM system. After milling, it is then transformed by a heat treatment into a dental prosthesis containing both glassy phase and crystalline phase. Different heating parameters can either promote or interfere with crystal growth and can affect the overall percentage of residual glasses.[17-22] Theoretically, glass-ceramic prostheses, containing residual glassy phase of glass-ceramic or porosity of polycrystalline ceramic, are more likely to adversely impact a number of properties including load-bearing capacity, resistance to acidic attacks, and fracture toughness.[23] The blocks can also be in the form of porous crystalline ceramic. The firing will induce the sintering process, where a final product will be a completely dense, pore-free dense polycrystalline ceramic.

Microstructural Phase Classification[10,13]

Predominantly glass–based system

In dentistry, feldspathic porcelain is defined as an amorphous aluminosilicate network that is interspersed with feldspar or leucite crystals and is classified as a predominantly glass-based structure. It consists mainly of silicate and aluminosilicate glass. The elements can be derived from the melting of silicate $[SiO_2]$, alumina $[Al_2O_3]$, and feldspathic minerals $[X_nAlSi_3O_8]$.[16] The major advantage of a "predominantly glass-based" prosthesis like feldspathic porcelain is its inherent translucency and enamel-like luster, which is highly esthetic and mimics natural tooth color. Nevertheless, its disadvantage is its strength, which is much weaker than the glassy-crystalline or polycrystalline restorations. It has low flexural strength, ranges from 60 MPa to 70 MPa; therefore, it can be used only as veneering material for metal or ceramic. This material is called glassy porcelain.

Glassy-crystalline system

The glassy-crystalline group consists of a wide variety of glass-ceramic system, including binary system [e.g. $Li_2O\text{-}SiO_2$ or $Li_2O\text{-}2SiO_2$], ternary system [e.g. $Li_2O\text{-}Al_2O_3\text{-}nSiO_2$ (LAS-System), $MgO\text{-}Al_2O_3\text{-}nSiO_2$ (MAS-System), or $ZnO\text{-}Al_2O_3\text{-}nSiO_2$ (ZAS-System)], and multicomponent system [e.g., IPS E-max® Press and IPS E-max® CAD; Ivoclar Vivadent, Schaan, Liechtenstein]. These glass-ceramic systems exhibit only 30 percent to 50 percent of an amorphous, glassy network with a glass-to-crystal ratio that ranges from 50 percent to 70 percent volume of crystallinity.[24]

Glass-ceramic fabrication can be achieved with dispersion-strengthening techniques or the preparation of a monolithic glass with controlled crystallization. The benefit of increasing crystallization phase is to resist crack advancement and ultimately to stop fracture. It is generally perceived that the crack advancement can be restrained by toughening the material through compositional or microstructural modifications.

The method to produce a glass-ceramic is the preparation of the composition in the monolithic glass and then to be treated by controlled crystallization. The most popular controlled-crystallization system that is commercially available for dental application is the lithium disilicate glass-ceramic. This category includes:

- Leucite content feldspathic glass ceramic.

The addition of potassium content to the glass-ceramic increases mechanical strength and maintains optical quality. Two common Leucite reinforced ceramic brands are IPS Empress (Ivoclar Vivadent), and OPC (Pentron). IPS Empress was developed at University of Zurich, Zurich, and was introduced to the dental market in 1990.[25] It has a 160 MPa flexural strength,[26] with crystal sizes of 1.5 μm to 2.6 μm that grow evenly in a multistage process.[27] OPC ceramic material has crystal sizes of 1.9 μm to 6.6 μm.[28] In addition, machine-able blocks of Leucite reinforced glass ceramic are available, such as Empress CAD (Ivoclar). Both machinable and pressable systems are shown to have higher fracture resistance and are reported to have good clinical results when used for veneers, inlays, onlays and anterior crowns.[11,29]

- Lithium disilicate glass ceramics.

Lithium disilicate glass ceramic was introduced by Ivoclar as IPS Empress II for the purpose of increasing the strength of dental ceramic but maintaining the optical

quality. It contains glass matrix and 70-percent-micron lithium disilicate crystals. The

crystals are derived from adding lithium oxide to the aluminosilicate glass, which

decreases the melting temperature of the material.[10] Modification of crystal size and an

increase in the amount of crystals lead to flexural strength of 360 MPa, which is about

three times stronger than Leucite glass ceramic.[30] Furthermore, Lithium disilicate crystals

have a low refractive index, which provides translucency even with high crystal content.

In 2005, IPS E-max (Ivoclar) was introduced by enhancing mechanical properties, but it

still has lower esthetic property than IPS Empress II.[31] Clinically, it was recommended

for a 3-unit bridge in the anterior region extending up to the second premolar.[32,33]

The machinable system, IPS E-max CAD is available as CAD-CAM blocks in

this system. They are partially sintered and need further heat treatment to complete the

sintering and complete the growth of the crystals. Generally, CAD-CAM blocks were

reported to have better mechanical properties than a pressable system because of the

standardized manufacturing process.[34] It can be used for posterior crowns and 3-unit

FPDs in clinical implication.[35,36] Due to the enhanced mechanical properties and good

esthetic results, lithium disilicate ceramic crowns are widely used. There are several

studies to support their clinical performance with clinical success rates of 100-percent

after 2 years and 93 percent after 8 years.[36,37]

Interpenetrating Phase Ceramics

Interpenetrating phase ceramics are derived from fabricating porous matrix, which

is filled with lanthanum aluminosilicate glass. In-Ceram family (Vivadent) is the example

in this group. In-Ceram Spinel (alumina and magnesia matrix) is the most translucent

with flexural strength of 350 MPa, for anterior crowns. In-Ceram Alumina demonstrates

450 MPa of flexural strength but has a lower translucency than the former. In-Ceram

Zirconia (alumina and zirconia matrix) has higher 650 MPa flexural strength but has

poor translucency.[10] The last two are usually veneered by porcelain due to their

opacity.[38,39] The fracture strength of In-Ceram Alumina demonstrated higher fractural

resistance than IPS Empress in an *in-vitro* study.[40]

Polycrystalline solids

A polycrystalline ceramic typically exhibits 95 percent and 99 percent volume

fraction of crystallinity.[14] These ceramics are derived directly from sintering crystals

without the glass phase to create a dense, air-free polycrystalline structure. This newly

developed poly crystal tends to have higher strength and toughness than usual glass and

glass-ceramics.

These polycrystalline ceramics with CAD-CAM fabrication have been introduced

for the possible application as posterior FDPs. In addition, there are several recent

laboratory and clinical studies, which have shown the excellent results of strength,

durability, and survival rates.[41-43] However, the drawbacks of polycrystalline ceramics are

that they show insufficient bonding to a tooth and undervalued esthetic outcome. The first

outstanding disadvantage is due to the lack of a glassy phase within the polycrystalline

network, which impairs the effectiveness of conventional adhesive luting procedures.

Additionally, polycrystalline ceramic has high opacity and low translucency, resulting in

less than optimal aesthetics. Veneering could enhance the esthetic result, although using

veneering on polycrystalline ceramic core material can compromise the strength. The

limited bonding strength would be exhibited at the interfacial surfaces between

polycrystalline substrate and veneering ceramic.[44] The most popular polycrystalline

compositions are alumina and zirconia (e.g., Procera[TM] Alumina and Procera[TM] Zirconia). Procera was the first dental application for fully dense polycrystalline material.

Zirconia as core material for crowns and FPDs

Traditionally, zirconia has been used as one of the most popular core materials for all-ceramic crowns due to its high strength and toughness. Zirconia is the toughest dental ceramic available in dentistry. The particle size is 0.1 μm to 0.5 μm.[10] The higher toughness comes from the additive of compound elements to yttria (Y_2O_3), magnesia (MgO), and ceria (CeO_2).[45] Three mol% yttria is usually added over the other oxides as the transformation toughening mechanism.[46] The high temperature monoclinic phase is stabilized at room temperature. As a crack propagates through the ceramic, a tetragonal-monoclinic phase transformation occurs accompanied with 3-percent to 4-percent volume expansion, which allows the material to arrest crack propagation and increase toughness.[47] Zirconia comes in the form of porous or dense CAD/CAM blocks so it can be milled easily when it is not fully sintered. The restorations should be milled oversized by about 25 percent to compensate for the shrinkage.[10]

Clinically, zirconia has been utilized to make crowns and FPDs frameworks instead of metal because of its high strength, opacity, and the white color. The zirconia core is usually fabricated from a CAD-CAM ingot followed by veneering with a more translucent ceramic to provide a more esthetic restoration. Although clinical studies of zirconia framework have been accepted and have not displayed a serious problem,[44,48] the most common failure reported was chipping of the veneering material,[49] because veneering material is usually weaker than the core material.[50] In addition, a crown made

entirely from traditional zirconia does not provide the desirable esthetic results due to the high opacity of the traditional zirconia material.

FULL CONTOUR ZIRCONIA

The problems of conventional zirconia bring to the development of new zirconia materials with higher translucency, commonly referred to as "full contour zirconia." This material has been introduced with the aim of allowing dentists to fabricate entire all-ceramic crowns from the material with acceptable esthetic and mechanical functions without the need for veneering. Due to their high strength, the crown will also require less tooth reduction than the clinical preparation for lithium disilicate material. The optical quality has been enhanced by several procedures such as hot-isostatic pressing (HIP), a high-pressure spark plasma sintering (SPS), the adjustment of the sintering temperature at 1450°C to 1500°C, and the addition of nanoparticles powder. These procedures change the particle size and porosity of the material, through which the translucency of the material is changed.[51-55]

Hot isostatic pressing (HIP) is a usual processing technique used to increase the translucency.[52,56] The zirconia powder is heated and pressed simultaneously by a heating coil, which eliminates pores in the sintered material, but increases grain size,[52] which impairs the mechanical and optical perspectives because of the reduction in grain boundaries.[53]

Alternatively, spark plasma sintering (SPS) is used to compromise the problems of the HIP technique. In SPS, a high-density flux runs through the sample and the graphite die to provide the low sintering temperature (~1200°C) while pressure is applied. This technique also allows for reduced heating and cooling time, minimizing the amount

of grain growth and the preparation of nanoparticle ceramic[54] and thereby producing

produces dense materials of less than 20-nm grain size.[57] Reducing the porosity of the

material and creating more grain boundaries will yield a greater toughness. Several

studies displayed the ability to change the shade of zirconia with SPS technique.[54,58,59] A

vacuum and graphite die are usually used in SPS technique to reduce the temperature in

sintering condition environments, resulting in the light absorption of oxygen vacancies,

which are called color centers. As a result, the color is modified to yellow-brown. In

addition, holding time at 1200°C during sintering is responsible for the level of coloration

as well.[53]

Additionally, the increasing temperature at the final sintering process from 1350

to 1500 °C would affect the light transmittance. Yttria stabilized tetragonal zirconia

polycrystal (YTZP) would have higher light transmittance from 2 percent to 16 percent

during sintering process from 1350 to 1450 °C, after that it could maintain 17-percent- to

18-percent transmittance at the final sintering temperature of 1450°C to 1500°C.[51]

Adding titanium oxide to yttrium-stabilized tetragonal zirconia was reported to

be effective in densifying yttria-stabilized zirconia.[55] Tsukuma studied the effect of TiO_2

on the transparency of zirconia. Ten (10) mol % TiO_2 was added to 8 mol % yttria-

zirconia powder and sintered under 1430°C for 12 hours and 1630°C for 7 hours.[52] TiO_2

dissolved in ZrO and formed a solid solution and stimulated grain growth during

sintering. Furthermore, the addition of TiO_2 also provides a higher transmittance to the

zirconia, whereas the pressure associated with TiO_2 adding technique also leads to pore

migration. This phenomenon can increase the transparency and the strength of the

zirconia.

However, there is very little information currently available on the translucency

parameters of full contour zirconia. A study compared the translucency parameter among

human dentin, human enamel, and two zirconia materials, IPS E-max Zir CAD and Lava,

show that there is no statistically significant difference between Lava and human dentin

in translucency parameter.[60] In addition, Fu Wang et al.[61] mentioned that all of the

zirconia ceramics and glass ceramic showed an exponential increase in translucency with

decreasing thickness. However, a systematic study investigating the difference between

the translucency parameter of full contour zirconia and traditional zirconia at the different

thickness is still lacking.

LIGHT, COLOR, AND ESTHETIC
APPEARANCE WITH THE MEASUREMENT

The ability to distinguish the aesthetic of the material depends on surface texture,

translucency, light source, and color. The light is a form of energy and composed of

different wavelengths. When the light strikes an object, it may be reflected, refracted,

absorbed, scattered, and transmitted. The light scatters in many different directions when

the surface is rough, while a smooth surface increases the specular reflectance, in which

the angulation of light reflection is equal to the angle of the light source. The light is not

totally reflected at the surface if the material is transparent; some light passes through the

material and emerges at the other side.[62] Since 1931, the light source used as the standard

daylight was C, incandescent or tungsten lamps (2856 K), or fluorescent lamps (4000 K).

Currently, an average standard daylight is D65 and D55.[63] Illuminant D65 and D55 refer

to the correlated color temperature of 6500 K and 5000 K, respectively. D65 was defined

in 1964 as the radiation of north sky daylight on a cloudy day. According to ISO 3668

and 3664, D65 is the standard application used in the industries, while D55 is used in the printing and graphic arts industries.[62,63]

Colorimeter is a standard measurement of color perception and is based on color science, which is composed of three components: light source, object, and human vision. The color order system is the way to arrange the three-dimensional space and facilitate color description. The Munsell and CIE systems have been widely used in the dental application. The Munsell system has been introduced to define color in terms of hue, chroma, and value. Munsell Hue is related to the perception of colors such as red, orange, green, or blue. The intensity of a particular hue is described by Munsell chroma. The value represents the color lightness or darkness, from 0 (black) to 10 (white).[62,64] The CIE system is the most widely used and was developed by the Commission International de l'Echairage (CIE, International Commission on Illumination). D65 and D50 have been used as illuminants by CIE. There are two equations of this system: Tristimulus value and CIELAB. Tristimulus would be analyzed in terms of three elements: red, green and blue. The latter equation is more familiar in dentistry and based on Munsell's system. The color spectral distribution is located by L*a*b*. The L* value of 0 to 100 represents from black to white, respectively. The a*and b* value represents the position on a red/green and yellow/blue axis, respectively.[62] This equation for translucency parameter is used to compare the measurement of the reflectance of light through the specimen over a background with high reflectance (white background) to that of high absorbance (black background). It represents the color difference between two backgrounds in order to directly correspond to a common visual assessment by using the following equation: TP $= \sqrt{[(L^*_B - L^*_W)^2 + (a^*_B - a^*_W)^2 + (b^*_B - b^*_W)^2]}$ Where, L* refers to the brightness, a*

represents redness to greenness, and b* is yellowness to blueness. The subscript B refers to the color coordination on the black background and W are those on the white background.[65]

The instrumental measurement of color and translucency can be divided into 3 types according to the type of measured index: Tristimulus colorimeter, spectroradiometer, and spectrophotometer. Tristimulus colorimeters only monitor three color elements and are suitable for industrial quality control. Spectroradiometers are designed to measure radiometric quantity: irradiance (W/m^2) and radiance (m^2Sr). Their units are expressed by luminance (cd/m^2) and illuminance (lux) for spectral radiance and irradiance, respectively. The most widely usage is the spectrophotometer for measuring surface color. It is designed to measure the ratio of the light reflectance and based on the CIE system. The result is quite stable and accurate as a absolute standard.[62]

LIGHT TRANSMISSION AND TRANSLUCENCY

The definition of translucency as the relative amount of light transmission through the material[66] is incorrect. Given that transparency (also called pellucidity or diaphaneity) is the physical property of allowing light to completely pass through the material without being scattered, then translucency (also called translucidity) is a super-set of transparency. The quality of translucency allows light to partially pass through the material. The photon can be scattered at either of the two interfaces where there is a change in the index of refraction, or internally. In other words, a translucent medium allows the transport of light, while a transparent medium only allows the transport of light while allowing image formation.[61,67] Translucency depends on the wavelength, material thickness, the type of material, and the surface roughness. The higher the

wavelength, the higher the translucency value.[62,68,69] The greater the thickness of the material, the lower the translucency parameter.[62,66,68,70] In addition, the type of material could affect the translucency. For example, zirconia is significantly less translucent than glass ceramic.[61] Alumina and magnesium, which strengthen the ceramic material, could make the porcelain more opaque. The ceramic sintering process can also affect the final grain size of the ceramic and the amount of the void. The greater the grain size of the ceramic, the higher the light transmittance percentage. The higher the amount of remaining void, the more scattering and lower light transmittance.[71] Furthermore, an opacifier is added in the ceramic material, such as oxide of barium, tin, titanium and zirconia, aluminum, magnesium, which affect the light reflectance and transmittance. It would increase scattering with a result of decreased translucency.[62] Surface roughness or gloss is the other factor that affects the translucency and interferes with the correct measurement. There are two forms of light, specular and diffuse transmittance that depend on the method of measurement. The specular transmittance would exclude the proportion of scattered light that does not reach the detector in order to reduce the error from surface gloss and roughness, whereas the diffuse transmittance includes all light scattering.[72,73]

DEGREE OF CONVERSION

Light transmission can have a strong effect on the degree of conversion of resin luting cement used for crown cementing. In general, the light transmission has impacted various factors of a ceramic material such as the thickness, shade of ceramic, its microstructure, curing mode, defects and porosity.[61,74-77] Degree of conversion of material represents the conversion of double bond to single bond of methacrylate group

during polymerization after that the material becomes more rigid.[78] Degree of conversion

significantly correlates to the mechanical and biological properties such that the higher

degree of conversion, the better the mechanical properties. Network formation and cross-

linkage occur during the setting reaction.[79] The degree of conversion of resin cement was

determined by Fourier Transformation Infrared Spectroscopy (FTIR) before and after

curing with the various thickness of ceramic material. This method is the most commonly

used to detect the carbon double bond stretching. The ratio of absorbance intensity of

aliphatic carbon double bond (peak at 1638 cm^{-1}) before curing and that of aromatic

carbon double bond for Bis GMA (peak at 1608 cm^{-1}) is evaluated and calculated for the

percentage of degree of conversion. The effect of translucency on the degree of

conversion of resin luting cement was also evaluated in this study.

METHODS AND MATERIALS

MATERIALS

The specimens were divided into six groups according to the types of materials (Table 1) as follows: Group 1 Lithium disilicate glass ceramic (E-max CAD); Group 2: Traditional Zirconia (CAP QZ) as control group, and Groups 3-7: Full Contour Zirconia (CAP FZ, Zirlux, Bruxzir, KDZ Bruxer). Recently marketed zirconia products were selected to be compared with glass ceramic and traditional zirconia.

The following full contour zirconia brands were evaluated in this study:

CAP FZ has been developed with the isostatically pressed process by Custom Automated Prosthetic in Germany. The flexural strength and compressive strength of material are estimated at 1100 MPa and 3000 MPa respectively. A modulus of elasticity is 205 GPa. There are 16 shades according to Vita shade guide.

Zirlux has been introduced by Zahn Dental Laboratory Division of Henry Schein. This company claimed that the medical grade zirconia has the flexural strength above 1100 MPa with high translucency. The approximate enlargement factor is determined to be about 22 percent.

BruxZir® Solid Zirconia has been improved by Glidewell Laboratories. The smaller zirconia particles have been physically and chemically processed to improve the mechanical and esthetic aspects with 1400 MPa in flexural strength and to provide an excellent observation of esthetic satisfaction. In addition, it has been claimed that there is less wear against the opposing dentition.

KDZ Bruxer has been developed by Keating Dental Arts with a high flexural strength of 1250 MPa.

METHODS

Ceramic Disk Fabrication

One-hundred and fifty (150 square-shaped ceramic specimens were prepared from CAD-CAM material blocks using a cutting machine (Isomet 1000, Buehler, Illinois, USA) (Figure 1). The E-max CAD ingots (HT A2/ B40) were cut and polished into thicknesses of 1 mm, 1.25 mm, 1.50 mm, 1.75 mm, and 2 mm. The traditional specimens and full contour zirconia at a larger dimension were cut to account for the shrinkage factor. The sintered zirconia specimens then were polished to achieve final specimen thicknesses of 1 mm, 1.25 mm, 1.50 mm, 1.75 mm, and 2 mm. There were 5 specimens in each thickness and type of ceramic (Figure 2). The ceramic specimens were cut as square samples of 12 mm × 12 mm size at the various thicknesses. The specimens were finished, using silicon carbide sand papers at 400 grit and 600 grit (EXAKT Technologies, Oklahoma City, OK, USA) at the dental material laboratory with 10 strokes in each direction of both sides by finger pressure under water lubrication. The specimens were measured using a vernier caliper with digital readout (Mitutoyo Corp, Tokyo, Japan) of the approximate thickness. Then, IPS E-max CAD disks were sintered in Programat CS furnace (Ivoclar Vivadent, Ontario, Canada) (Figure 3) according to manufacturer's recommendations (Table 2) without glazing. All zirconia specimens were sintered using a furnace (Blue M, SPX Corp., PA, USA) (Figure 4) following the manufacturers' instructions for each material (Table 3) without glazing afterward.

Translucency Measurement

The translucency parameter (TP) developed by Johnson et al. (1995) was used.[65] This parameter is calculated from the differences between the color reflectance data of white and black in the visible range 380 nm to 780 nm, according to the following equation.

$$TP = \sqrt{[(L^*_B - L^*_W)^2 + (a^*_B - a^*_W)^2 + (b^*_B - b^*_W)^2]}$$

Where, L^* refers to the brightness, a^* represents redness to greenness, and b^* is yellowness to blueness. The subscript B refers to the color coordination on the black background and W is for those on the white background.

After sintering, sample dimensions were confirmed using a Verviers caliper with digital readout (Mitutoyo Corp, Tokyo, Japan) at the center of each sample, in each thickness group. All specimens were polished by 600-grit and 1200-grit (EXAKT Technologies, Oklahoma City, OK, USA) silicon carbide sand papers under water lubrication with 10 finger strokes on both sides before testing. After that, the color space by CIE (L, a and b) of all specimens was measured by a spectrophotometer (CM-2600D, Konica Minolta Sensing Americas, Inc., Ramsey, NJ) (Figure 5). The standard of device was controlled at a 10-percent observer angle, a 100-percent UV and standard illuminant D65 as the standard wavelength between 300 nm to 780 nm (Figure 6). The light reflected on the surface of specimens through an 8-mm target mask. Irradiance was measured with the ceramic discs of 1-mm, 1.25-mm, 1.5-mm, 1.75-mm and 2-mm thickness inserted underneath a spectrophotometer device on either a white (Figure 7) or a black background (Figure 8).

The data with specular component included and excluded (SCI and SCE) were recorded to compare the effect of surface roughness.

Light Transmission

Managing Accurate Resin Curing (MARC® Resin Calibrator, BlueLight Analytics, Inc., Halifax, NS, Canada) (Figure 9) consists a laboratory grade UV-VIS spectrometer and two laboratory grade cosine corrected sensors (top and bottom). Light captured by the sensors is transmitted to the spectrometer through a bifurcated fiber optic cable, after which dedicated software provides real-time irradiance data. The MARC® Resin Calibrator was set to monitor the curing time for 20 seconds and the sensor trigger at 50. The average irradiance and the peak wavelength for the light curing unit were determined to be 1071 mW/cm^2 and 450 nm, respectively.

Irradiance and spectra of the halogen light curing unit (DEMI LED, Kerr, CA, USA) were measured in the standard mode. Irradiance was measured at a distance of 0 mm with the ceramic disks from each sample group inserted between the light curing unit and a radiometer device, MARC® Resin Calibrator.

Degree of Conversion of the Light-Cured Resin Cement

In this infrared spectroscopic technique, the degree of conversion (DC) of a resin composite is measured with a Fourier transform infrared spectrometer (FTIR) in attenuated total reflection (ATR) mode (FT/IR 4100, JASCO Analytical Instruments, Tokyo, Japan) (Figure 10). The degree of conversion is calculated utilizing the mid-IR range peaks of 1608 cm^{-1} and 1638 cm^{-1}. The area under the peak at 1638 cm^{-1} (P1) represents the vinyl C=C groups of the resin composite, while the area under the peak at

1608 cm^{-1} (P2) represents the aromatic C=C and serves as the internal standard. In general, the resin cement consists of the aromatic double bond from Bis-GMA; therefore, the 1638 cm^{-1} peak intensity was determined according to this formula.[80]

$$\text{Degree of conversion} = \text{\# of converted C=C} / \text{Total \# of C=C}$$

$$= (\text{Total \# of C=C} - \text{Remaining \# of C=C}) / \text{Total \# of C=C}$$

$$= 1 - (\text{Remaining \# of C=C} / \text{Total \# of C=C})$$

$$= 1 - \frac{\text{Cured area under 1638 - Cured area under 1608}}{\text{Uncured area under 1638 - Uncured area under 1608}}$$

The degree of conversion of the light curing resin cement (Variolink II, Ivoclar Vivadent, Liechtenstein, Germany) was also investigated and related to light transmission in different thicknesses of various zirconia. Only the base paste of resin cement was tested, with no catalyst present. The degree of conversion was determined with infrared (IR) spectroscopic technique. First, a small quantity of uncured resin cement was placed directly on the spectrometer's diamond crystal plate (ATR-MIRacle, Pike technologies, Madison, WI, USA) under dark conditions. It was then placed in the FTIR sample holder and FTIR spectra were recorded. The measuring area of FTIR Spectrometer was 1.8 mm in diameter, and the wavelength of the FTIR spectra ranged from 4000 cm^{-1} to 1500 cm^{-1}, and spectra were recorded with 64 scans per spectrum at a resolution of 4 cm^{-1}. Measurement of degree of conversion was conducted at the room temperature (22° C). Three scans of the uncured resin were performed.

For the cured resin cement, the cement was placed between two mylar strips, with a glass slab beneath and on the top to avoid air entrapment. The film thickness was controlled by the matte-plastic mold that was 191 μm in thickness. Before testing, the light-curing unit was used to measure the irradiance. It was then cured through the

different thicknesses of ceramic disks and the controlled curing source for 40 seconds using a light curing unit (LEDemetron, SDS/Kerr, USA). The position of curing tip were verified by the curing unit holder of a MARC® Resin Calibrator (BlueLight Analytics, Inc., Halifax, NS, Canada) (Figure 11). The tip of the curing unit was placed against the disk specimens. Then, the opaque-plastic flame was inserted around the specimen in order to prevent the light transmission (Figure 12). The resin cement strips were removed from the ceramic disk and were immediately placed in a standard FTIR holder. A thin slice of resin cement was placed directly on the spectrometer's diamond crystal sample holder with the surface of the crystal. FTIR Spectra were collected in the same manner as for the uncured resin cement. Two scans from different areas of the surface (right and left) for each slice of resin were performed.

Statistical Methods

Translucency and degree of conversion were compared using two-way ANOVA with group, thickness, and their interaction as factors in the models. Linear regression was used to evaluate the association of thickness with translucency and the degree of conversion, and to test whether the associations vary by group.

With a sample size of five specimens from each group for each thickness, the study had 80-percent power to detect a translucency difference of 1.6 between any two groups, assuming two-sided tests each conducted at a 5-percent significance level, non-significant interaction between group and thickness, and a within-group standard deviation of 2.0 based on pilot data. This sample size was also provide 80-percent power to detect an increase of 0.10 in the R^2 when groups were allowed to have different slopes, and the underlying overall correlation is 0.30.

RESULTS

RESULTS OF TRANSLUCENCY PARAMETER,
LIGHT TRANSMISSION AND DEGREE OF CONVERSION

The translucency parameter, light intensity and degree of conversion values are listed in Table IV to Table VI.

In translucency parameter (SCI data), E-max CAD showed the greatest value of 12.44 (0.14) while QZ shows the smallest value of 0.29 (0.10) (Table IV), (Figure 13). The SCI and SCE data shows the same tendency, but the SCE data displayed a high level of scattering due to surface roughness (Figure 14). Therefore, only SCI data were statistically evaluated in this study.

In transmitted light intensity, E-max CAD also showed the highest value of 405.88 (15.44) mW/cm^2 at a layer thickness of 1.0 mm while Bruxzir shows the lowest value of 14.80 (0.84) mW/cm^2 at a layer thickness of 1.25 mm. (Table V), (Figure 15).

In a degree of conversion, E-max CAD showed the greatest value of 60.95 percent (0.45) at a layer thickness of 1.0 mm while Bruxzir shows the smallest value of 3.08 percent (0.34) at a layer of 2 mm (Table VI) (Figure 16).

Results for Translucency Parameter

The translucency parameter of E-max CAD ranged from 7.36 to 12.56, those of QZ ranged from 0.18 to 0.99 and those of full contour zirconia from 0.20 to 6.04. There was an increase in TP with a decrease in thickness (Table IV), (Figure 13). The general ranking of TP was E-max CAD > FZ = Zirlux = KDZ Bruxer > Bruxzir and QZ.

Results for Light Transmission

The data of 1.5 to 2 mm of Bruxzir were excluded because the irradiance could not be detected.

After the irradiance passed through 1 to 2 mm disc of E-Max CAD, the light intensity displayed 204-426 mW/cm^2 which correspond to a reduction of 60 percent to 81 percent of the original light intensity. While those of QZ showed 59 maw/cm^2 to 201 maw/cm^2 which correspond to about 81-percent to 96-percent reduction and those of full contour zirconium showed 0 maw/cm^2 to 385 maw/cm^2 or a 64-percent to 100-percent reduction (Table V), (Figure 15).

Results for Degree of Conversion

The degree of conversion of resin cement after curing through 1 mm to 2 mm of E-max CAD ranged from 54 percent to 61 percent, those through QZ ranged from 50 percent to 53 percent and those of full contour zirconia ranged from 3 percent to 59 percent. There was an increase in DC with a decrease in thickness. (Table VI) (Figure 16).

STATISTICAL ANALYSIS OF MATERIAL COMPARISONS

At 1-mm thickness, the materials were significantly different from each other except for FZ and KDZ Bruxer in degree of conversion.

At 1.25-mm thickness, the materials were significantly different from each other except for Bruxzir and QZ in degree of conversion, FZ and KDZ Bruxer in degree of conversion, FZ and Zirlux in degree of conversion, and KDZ Bruxer and Zirlux in light intensity.

At 1.5-mm thickness, the materials were significantly different from each other except for FZ, KDZ Bruxer, and Zirlux in degree of conversion; KDZ Bruxer and Zirlux light intensity; Bruxzir and QZ in TP (SCE); and KDZ Bruxer and Zirlux in TP (SCE).

At 1.75-mm thickness, the materials were significantly different from each other except for QZ and Zirlux in degree of conversion, KDZ Bruxer, and Zirlux in light intensity, and Bruxzir and QZ in TP (SCE) and TP (SCI).

At 2-mm thickness, the materials were significantly different from each other except for KDZ Bruxer and Zirlux in light intensity, and for Bruxzir and QZ in TP (SCE) and TP (SCI).

THICKNESS COMPARISONS

For all outcomes, most measurements within each material showed statistically significant differences at different thicknesses. Following are the exceptions that were not significantly different: 1 and 1.25 for Bruxzir in degree of conversion, TP (SCE), and TP (SCI); 1.25 and 1.5 for Bruxzir in TP (SCI); 1 and 1.25 for QZ in degree of conversion; 1 and 1.5 for QZ in degree of conversion; 1.5 and 1.75 for CAP QZ in degree of conversion, TP (SCE), and TP (SCI); 1.75 and 2 for QZ in degree of conversion and TP (SCI); 1 and 1.25 for CAP FZ TP (SCI); 1.25 and 1.5 for FZ in degree of conversion and TP (SCE); 1.5 and 1.75 and 2 for CAP FZ in degree of conversion; 1.5 and 1.75 for KDZ Bruxer TP (SCE); 1.25 and 1.5 for Zirlux in degree of conversion and TP (SCE); 1.5 and 1.75 for Zirlux TP (SCE) and TP (SCI); and 1.75 and 2 for Zirlux in degree of conversion.

SLOPE

There was not a significant association between thickness and degree of conversion for QZ (p = 0.20) or FZ (p = 0.08) (i.e. the slopes were not significantly different from zero). All other slopes were statistically significant with negative associations between thickness and the outcomes.

STATISTICAL ANALYSIS OF SLOPE COMPARISONS

For TP (SCE), E-max CAD had a stronger negative slope than all other materials; KDZ Bruxer, Zirlux, and FZ had stronger negative slopes than Bruxzir and QZ; and Bruxzir had a stronger negative slope than QZ.

For TP (SCI), E-max CAD had a stronger negative slope than all other materials; Zirlux and KDZ Bruxer had stronger negative slopes than FZ, Bruxzir and QZ; FZ had a stronger negative slope than Bruxzir and QZ; and Bruxir had a stronger negative slope than QZ.

For light intensity, FZ had a stronger negative slope than all other materials; E-max CAD had a stronger negative slope than QZ, Zirlux, KDZ Bruxer, and Bruxzir; and QZ had a stronger negative slope than Bruxzir.

For degree of conversion, Bruxzir had a stronger negative slope than all other materials, and Zirlux had a stronger negative slope than CAP QZ.

In summary, E-max CAD has a significantly greater translucency parameter and light transmission than all zirconia brands. Bruxzir showed significantly lower light transmission and degree of conversion than other zirconia groups. The light could not penetrate after curing through more than 1.5 mm of Bruxzir.

TABLES AND FIGURES

TABLE I

The material used in this study

Group	Brands (Manufacturers, Batch No.)	Materials
1 (control group)	IPS E-max CAD (IvoclarVivadent, Liechtenstein, Germany, 634587)	Lithium Disilicate
2 (control group)	CAP QZ (Custom Automated Prosthetics, Stoneham, MA, USA, 2000QZ1012)	Traditional Zirconia
3	CAP FZ (Custom Automated Prosthetics, Stoneham, MA, USA, 3000FZ1012)	Full Contour Zirconia
4	Zirlux (Ardent Inc., Amherst, NY, USA, 64773)	Full Contour Zirconia
5	Bruxzir (Glidewell Dental Labs, Newport Beach, CA, USA, 70-1138-BSA0657)	Full Contour Zirconia
6	KDZ Bruxer (Keating Dental Arts, Irvine, CA, USA, PSZ174MB2014-02)	Full Contour Zirconia

TABLE II

The sintering cycle in degree Celsius for IPS E-max CAD

Stand by temp	Closing time (mm:ss)		Temperature increase	Holding temp. (°C)	Holding time (mm:ss)	Vacuum on temp (°C).	Vacuum off temp (°C)	Long-term cooling (°C)
403	06:00	90/30	820/840	00:10/07:00	550/820	820/840	700	

TABLE III

The sintering cycle in degree Celsius for the other Zirconia groups

Materials	Temperature 1			Temperature 2		
	Rate (°C/min)	Temp (°C)	Hold (mins)	Rate (°C/min)	Temp (°C)	Hold (mins)
CAP QZ		25		20	980	1
CAP FZ		25		8	980	1
Zirlux	10	700		2-5	1500	120
Bruxzir	10	25			1530	120
KDZ		25		8	1000	

Materials	Temperature 3			Temperature 4			Cooling
	Rate (°C/min)	Temp (°C)	Hold (mins)	Rate (°C/min)	Temp (°C)	Hold (mins)	Rate (°C/min)
CAP QZ	10	1530	120		1530		13
CAP FZ	6	1550	120		1550		10
Zirlux							Natural
Bruxzir					1530		4
KDZ	2	1590			1590	180	Natural

TABLE IV

The mean, standard deviation, standard error, minimum
and maximum for translucency parameter

Group	Thickness (mm)	N	Mean	Standard Deviation	Standard Error	Min	Max
E-max CAD	1	5	12.44	0.14	0.06	12.29	12.56
	1.25	5	11.24	0.15	0.07	11.47	11.08
	1.5	5	9.72	0.18	0.08	9.96	9.46
	1.75	5	8.67	0.04	0.02	8.61	8.73
	2	5	7.85	0.58	0.26	8.85	7.32
QZ	1	5	0.86	0.12	0.05	0.99	0.66
	1.25	5	0.65	0.07	0.03	0.75	0.58
	1.5	5	0.46	0.05	0.02	0.54	0.41
	1.75	5	0.42	0.12	0.05	0.55	0.28
	2	5	0.29	0.10	0.05	0.42	0.18
FZ	1	5	4.14	0.68	0.30	5.08	3.34
	1.25	5	3.54	0.54	0.24	4.45	3.14
	1.5	5	2.31	0.19	0.08	2.58	2.11
	1.75	5	1.80	0.24	0.11	1.94	1.38
	2	5	1.44	0.16	0.07	1.53	1.2
Zirlux	1	5	5.37	0.09	0.04	5.5	5.26
	1.25	5	4.37	0.49	0.22	5.04	3.66
	1.5	5	3.06	0.47	0.21	3.35	2.23
	1.75	5	2.76	0.55	0.25	3.73	2.39
	2	5	2.03	0.26	0.12	2.48	1.81
KDZ	1	5	5.84	0.21	0.09	6.04	5.54
	1.25	5	4.98	0.13	0.06	5.13	4.77
	1.5	5	3.90	0.14	0.06	4.05	3.73
	1.75	5	3.58	0.32	0.14	3.94	3.26
	2	5	2.62	0.06	0.03	2.72	2.58
Bruxzir	1	5	1.70	0.10	0.04	1.87	1.62
	1.25	5	1.43	0.76	0.34	2.78	0.98
	1.5	5	0.82	0.38	0.17	1.39	0.44
	1.75	5	0.42	0.09	0.04	0.57	0.36
	2	5	0.28	0.06	0.03	0.34	0.20

TABLE V

The mean, standard deviation, standard error, minimum
and maximum for light intensity (mW/cm^2)

Group	Thickness (mm)	N	Mean	Standard Deviation	Standard Error	Min	Max
E-max CAD	1	5	405.88	15.4	6.90	386	426
	1.25	5	351.77	7.7	3.44	342	362
	1.5	5	292.42	5.8	2.59	286	298
	1.75	5	266.27	11.6	5.21	253	284
	2	5	218.31	11.2	5.02	204	232
QZ	1	5	189.76	13.1	13.07	168	201
	1.25	5	156.72	8.6	8.56	146	169
	1.5	5	110.98	7.8	7.80	101	119
	1.75	5	92.14	5.2	5.23	86	98
	2	5	64.57	3.5	3.53	59	69
FZ	1	5	363.84	16.1	7.18	341	385
	1.25	5	312.54	2.5	1.11	309	315
	1.5	5	242.27	15.6	7.29	218	263
	1.75	5	201.21	5.2	7.09	187	228
	2	5	160.92	7.5	3.36	151	172
Zirlux	1	5	146.40	5.9	2.64	137	152
	1.25	5	93.79	4.5	2.01	87	99
	1.5	5	60.77	2.5	1.11	57	63
	1.75	5	43.60	2.7	1.21	41	48
	2	5	26.60	1.1	0.51	25	28
KDZ	1	5	137.87	6.6	2.97	128	143
	1.25	5	93.28	1.4	0.64	91	95
	1.5	5	64.07	3.4	1.51	59	68
	1.75	5	46.04	1.2	0.54	45	48
	2	5	28	1.6	0.71	26	30
Bruxzir	1	5	28.20	2.9	1.28	26	33
	1.25	5	14.80	0.8	0.37	14	16
	1.5	5	None	None	None	None	None
	1.75	5	None	None	None	None	None
	2	5	None	None	None	None	None

TABLE VI

The mean, standard deviation, standard error, minimum
and maximum for degree of conversion (percentage)

Group	Thickness (mm)	N	Mean	Standard Deviation	Standard Error	Min	Max
E-max CAD	1	5	60.95	0.45	0.20	60.39	61.59
	1.25	5	59.99	0.53	0.23	59.46	60.77
	1.5	5	58.21	0.35	0.16	57.94	58.74
	1.75	5	57.30	0.69	0.31	56.16	58.02
	2	5	55.12	0.58	0.26	54.33	55.95
QZ	1	5	52.74	0.73	0.33	51.51	53.41
	1.25	5	53.11	0.28	0.12	52.80	53.44
	1.5	5	52.05	0.69	0.31	51.40	52.60
	1.75	5	51.48	0.56	0.25	50.74	52.03
	2	5	51.10	0.55	0.24	50.25	51.72
FZ	1	5	59.20	0.15	0.06	59.05	59.37
	1.25	5	57.59	0.95	0.73	56.23	58.66
	1.5	5	56.68	0.66	0.27	56.22	57.97
	1.75	5	56.30	0.15	0.08	56.15	56.50
	2	5	56.55	0.66	0.29	55.47	57.26
Zirlux	1	5	57.95	0.87	0.39	57.37	59.49
	1.25	5	56.82	0.35	0.16	56.22	57.10
	1.5	5	56.90	0.53	0.24	56.33	57.52
	1.75	5	51.94	0.76	0.34	50.63	52.02
	2	5	52.23	0.32	0.14	51.86	52.67
KDZ	1	5	58.85	0.40	0.18	58.28	59.15
	1.25	5	57.26	0.16	0.07	57.09	57.50
	1.5	5	56.58	0.24	0.11	56.33	56.96
	1.75	5	54.93	0.18	0.08	54.72	55.18
	2	5	53.90	0.09	0.04	53.82	54.01
Bruxzir	1	5	54.49	0.24	0.11	54.28	54.81
	1.25	5	53.24	1.79	0.80	50.10	54.35
	1.5	5	32.04	3.74	1.67	26.23	35.23
	1.75	5	4.71	0.42	0.19	4.04	5.16
	2	5	3.08	0.34	0.15	2.79	3.67

FIGURE 1. Isomet 1000, a cutting machine.

FIGURE 2. Diagram of the amount of specimens.

FIGURE 3. Programmat S1 for IPS E-max CAD sintering.

FIGURE 4. Blue M for zirconia sintering.

FIGURE 5. CM-2600 D, a spectrophotometer, used to evaluate light reflectance.

FIGURE 6. The standard wavelength of D65 is between 300 nm to 780 nm.

FIGURE 7. White background for translucency parameter testing.

FIGURE 8. Black background for translucency parameter testing.

FIGURE 9. MARC resin calibrator, a blue light analytics for light transmission testing.

FIGURE 10.   FTIR (at right) used to measure the degree of conversion. The
diamond crystal and mounting assembly (at left).

FIGURE 11.   The curing unit holder used to verify the position of curing unit tip at the center of the matte-plastic mold.

FIGURE 12.   The opaque-plastic frame used to prevent
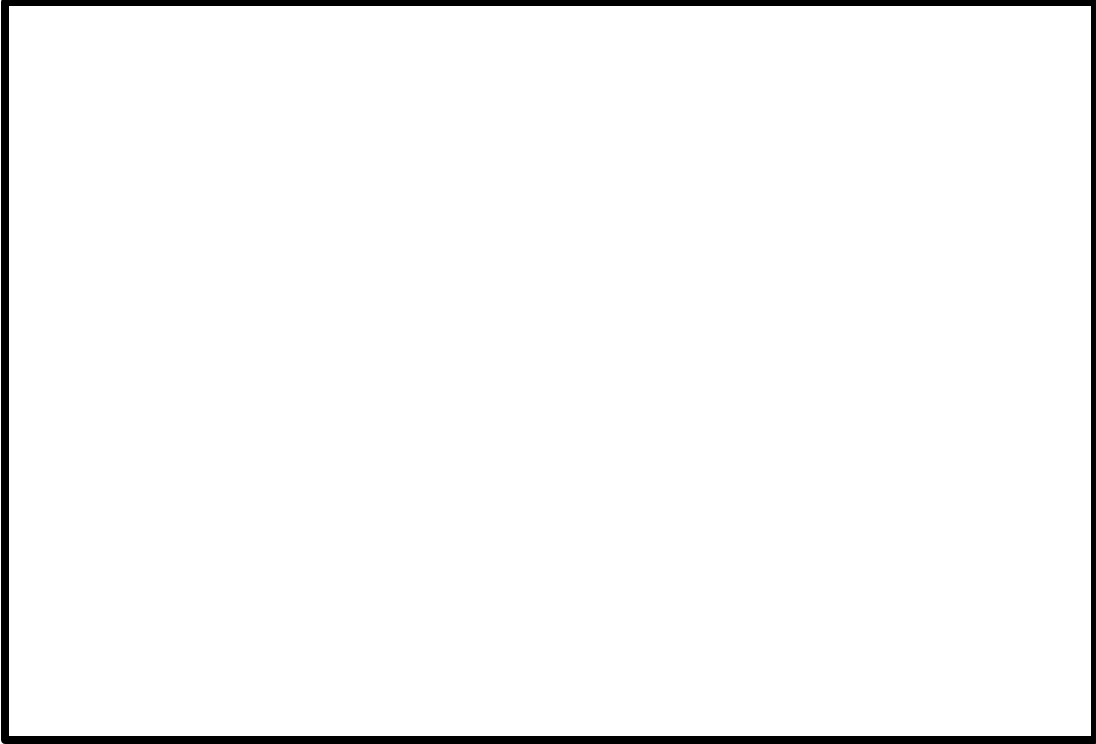the light transmission around the specimen.

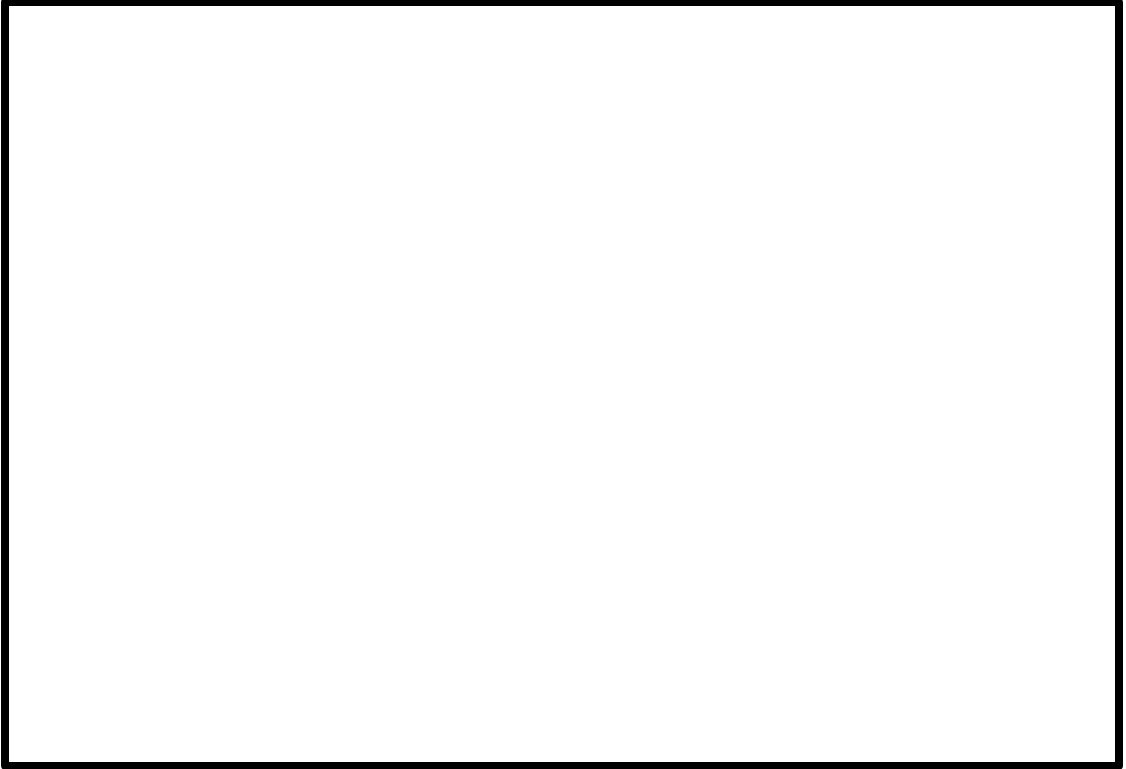FIGURE 13. The translucency parameter (SCI data)in each thickness.

FIGURE 14. The translucency parameter (SCE data) in each thickness.
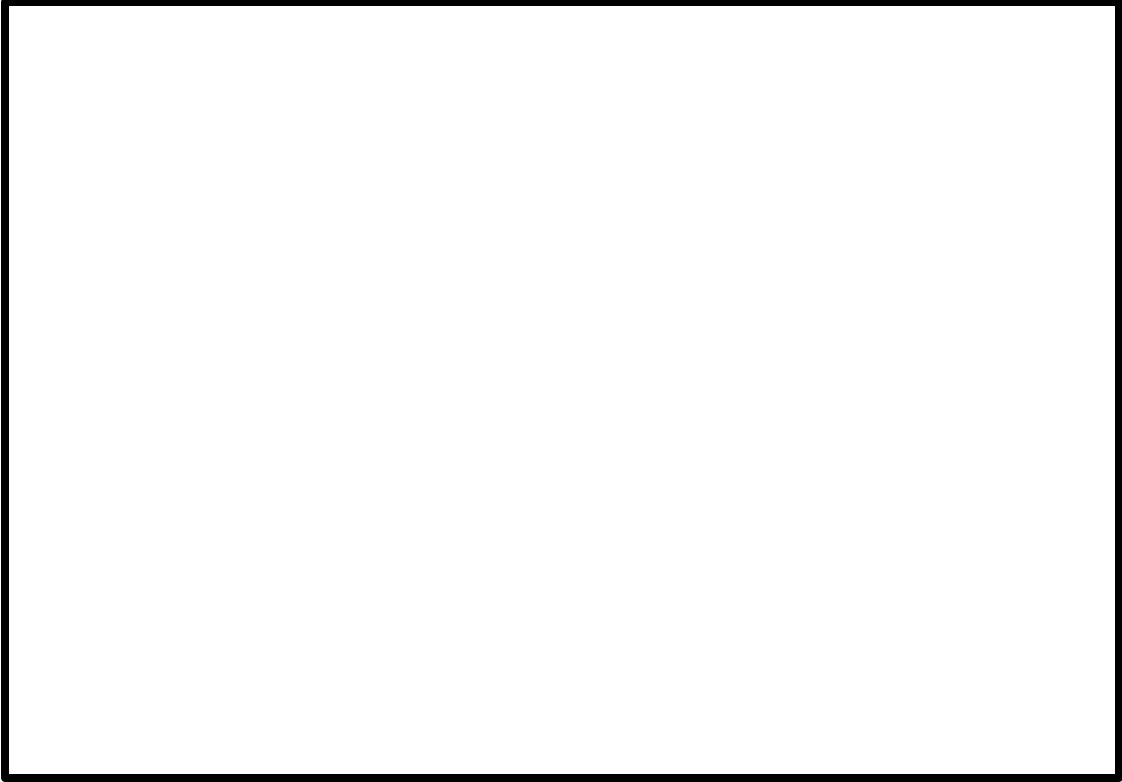
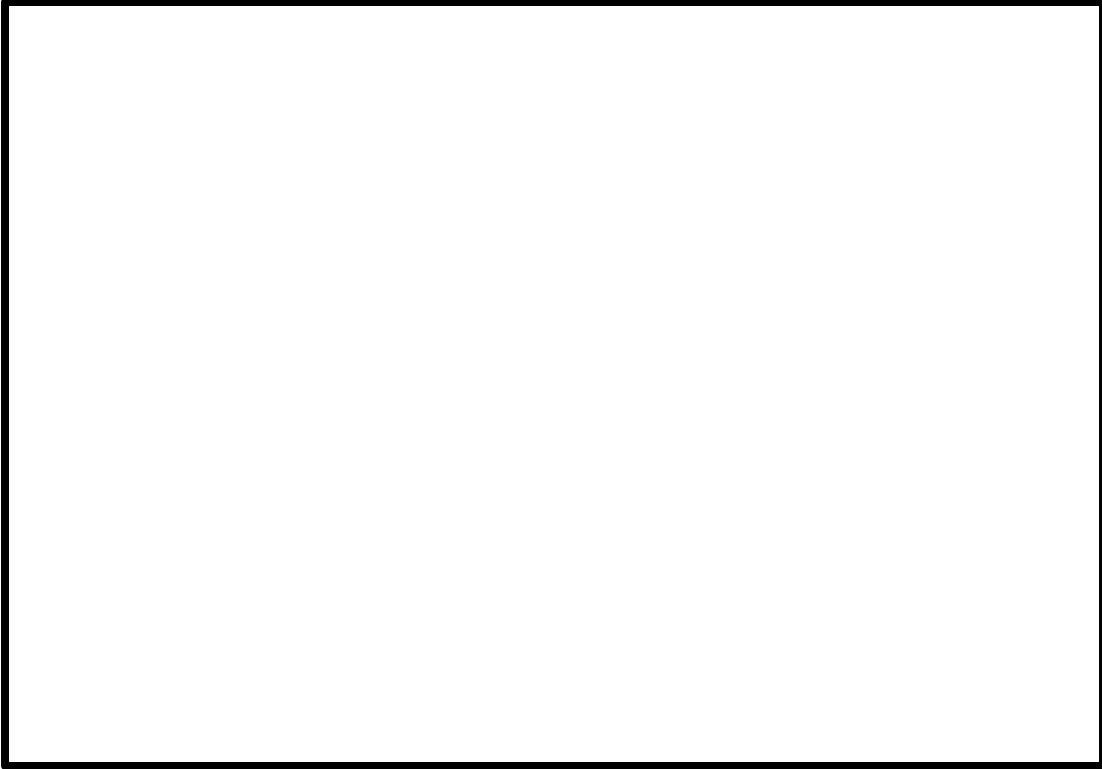FIGURE 15. The light intensity (mW/cm^2) in each thickness.

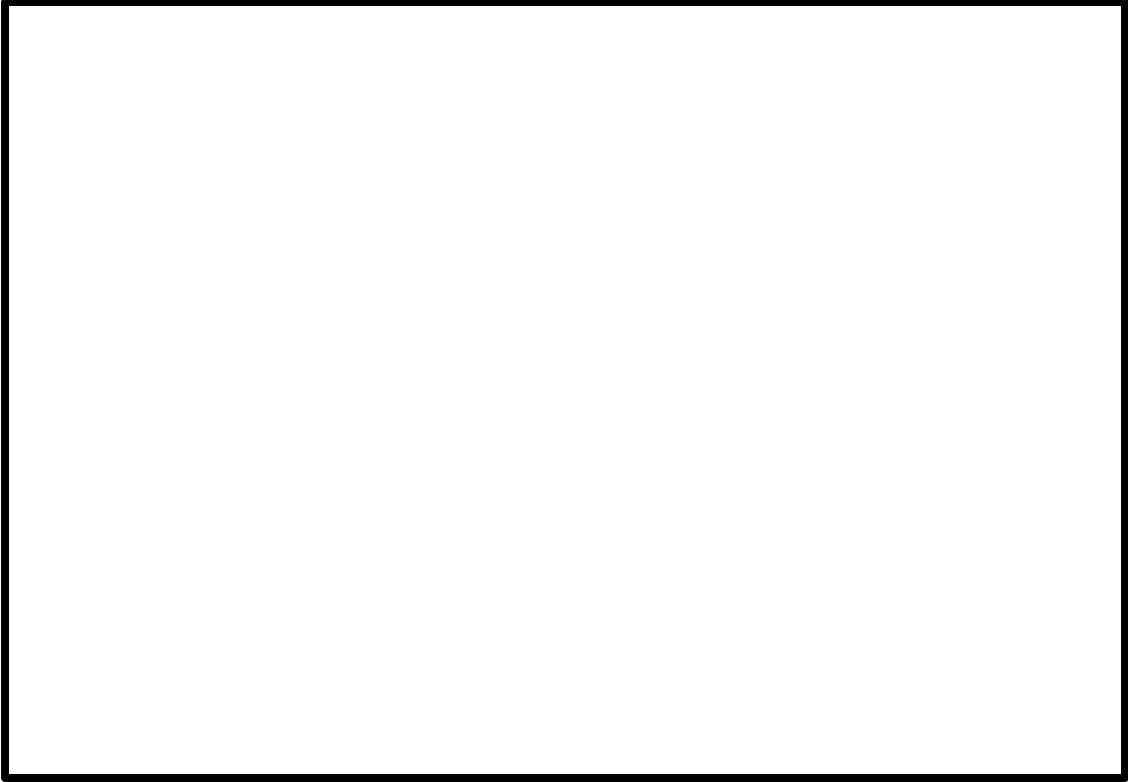FIGURE 16. The degree of conversion (percentage) in each thickness.

FIGURE 17. The relationship between light transmittance and translucency parameter.
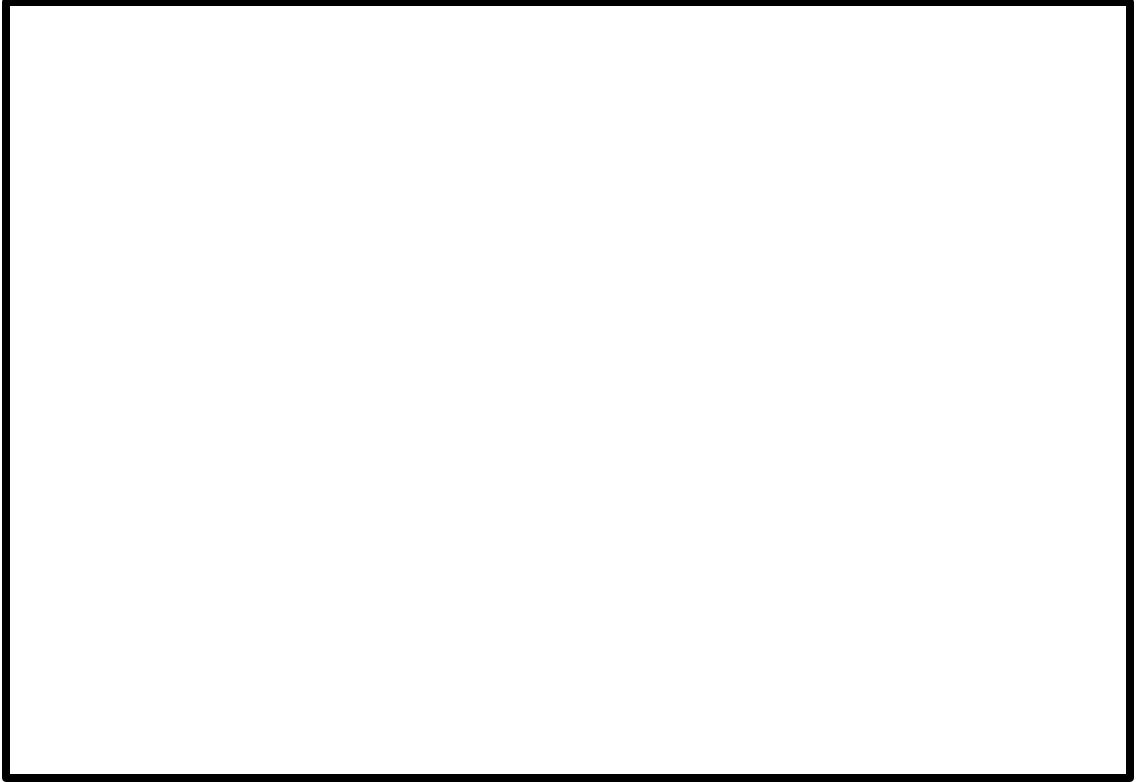
FIGURE 18. The relationship between degree of conversion and light intensity.

DISCUSSION

Translucent full contour zirconia has been accepted for use, because the material's mechanical properties require less tooth reduction and allow good optical quality. This development can save time and laboratory costs, and prevent the common problem of chipping in the veneering layer associated with veneered restoration.[50] However, after full contour zirconia was launched into the dental market, there was no adequate information about its esthetic and optical properties.

In the first hypothesis, we validated that the translucency parameter of all full contour zirconia was significantly smaller than that of E-max CAD, but only some full contour zirconia demonstrated significantly greater parameters than those of traditional zirconia. When comparing several studies about translucency parameters, our current study showed that the translucency parameter of E-max CAD was about 12.44 at 1 mm ceramic thickness whereas Fu Wang et al.[61] reported a translucency parameter of E-max CAD that was approximately 19 mm at the same thickness, which is a greater parameter than the result in this study. Yu [68] used a 3-mm aperture and found the TP of human dentine and enamel to be 16.4 and 18.7, respectively while Ryan et al. showed the translucency parameter of enamel to be 11.6 using an 8-mm aperture.[81] In this study, an 8- mm aperture was used for the measurement. It is known that TP is influenced by the diameter of the aperture, the ceramic thickness, and the testing machine.[61,68] Our results confirmed that the translucency parameter of E-max CAD is obviously more similar to the natural tooth structure than those of traditional zirconia (CAP QZ) which is 0.86. This data are comparable to Haffernan's result, about 0.7-1.[70] Baldissara et al also found that

the translucency of zirconia is significantly lower than that of lithium disilicate glass ceramic.[82] The results of the translucency parameter of full contour zirconia have not been reported in any other article. In the present study, only some recently marketed full contour zirconia (CAP FZ, Zirlux and KDZ Bruxer) had a higher translucency than traditional zirconia. The translucency difference was assumed to be the result of the various amounts of crystal, the sizes of particles and porosity of these materials, including the sintering temperature. They determined the amount of the light that is absorbed, reflected, and transmitted. All of the above need to be further investigated for full contour zirconia.

It has been demonstrated that the type of ceramic also influences the translucency parameter of ceramic material. Not only the type of ceramic is important, but also the thickness is a crucial factor affecting the translucency parameter as well. The higher the thickness of any type of ceramic disk, the smaller the translucency parameter becomes. In addition, it is known that an increase in thickness will exponentially reduce the light transmitting from the light source. Several studies displayed agreement that 60 percent to more than 80 percent of the light intensity reduction is seen after an increase in the thickness of the lithium disilicate glass ceramic from 1 mm to 1.5 mm.[83-85] In the present study, E-max CAD demonstrated a 60 percent to 80 percent light reduction, while both traditional and full contour zirconia eliminated more than 80 percent of the light intensity with 1 mm to 2 mm of ceramic disk.

Following the result of the light transmission underneath the ceramic disk, the degree of conversion was reduced to 54 percent to 61 percent of the original degree of conversion. The degree of conversion is approximately equal to the result of Flury,[84] 45

percent to 65 percent after curing through a 1.5-mm E-max CAD. There is no significant difference among the types of ceramic systems. However, it could be distinguished among full contour zirconia, especially Bruxzir. In this group, the degree of conversion was quite low after a 1.5-mm disk so that neither the light transmission nor the degree of conversion could be measured. As Flury et al.[84] demonstrated, the light curing through a 1.5-mm ceramic disk (ProCAD and E-max CAD) did not lead to a significant decrease in the degree of conversion, but a 3-mm ceramic disk resulted in a significantly lower degree of conversion.

Additionally, the curing condition and polymerization mechanisms of resin cement are vital factors affecting the mechanical properties of the resin cement underneath the ceramic material. In this study, we used a curing of 1,071 mW/cm^2 in 40 seconds and only the photo-polymerizable portion of the resin cement was investigated. The main purpose in the present study was to demonstrate the genuine result of the translucency of the ceramic disk to the degree of conversion without the influences of setting time and the effect of the catalyst. The result showed a low percentage of degree of conversion in several groups. In the Bruxzir group, the degree of conversion in the resin cement was too low to be detected. The curing mode of resin cement could be the reason for the undetectable degree of conversion. However, the degree of conversion is critical to the material's mechanical properties. Therefore, Bruxzir will not be recommended as a crown material with only light-cured resin cement. It will result in an unacceptably low degree of conversion of the resin cement underneath a 1.5-mm of ceramic thickness. Ilie et al.[86] recommended that at least 15 seconds of a high-power curing unit (1600 mW/cm^2) will be necessary to properly cure both dual and light-cured

resin cement through 0.5 mm to 3 mm of ceramic disks. The catalyst of resin cement in the dual-cured mode could improve 50 percent of Vicker hardness when comparing to only a light-cured mode. In addition, Meng et al.[87] suggested that light curing remains more favorable for dual-curing cements. Therefore, the high light transmission efficiency and the use of dual-cured mode are important. Our research could partially inform how the light affects the degree of conversion of the resin cements underneath the translucency of the various ceramic disks. However the comparison of the degree of conversion and the mechanical properties between light and dual-cure resin cement under full contour zirconia needs to be further investigated as clinical application.

While the definitions of degree of absorbance and degree of light transmission has been widely accepted for decades, the definition of TP was only proposed by Johnston very recently in 1995. The measurement of TP, involves measuring the differences in reflectance when the object, is laid against a black and a white background. It is essentially measuring the change in color of the light reflected from the material with the white and the black backing. A higher TP value will indicate larger changes in reflected color and therefore a more translucent material, since the material is not able to mask the effect of the black backing. On the other hand, a lower TP will indicate less color change and therefore a better ability of the material to block or mask the color underneath, hence a lower translucency. As expected, LDGC shows the highest TP, while traditional zirconia shows the lowest TP, and all full contour zirconia show intermediate TP values. (Figure 13).

One would expect that the intensity behind the disks would have a positive correlation to the transparency parameter. Surprisingly, the light intensity under the disk

follows a different trend, where the two materials with universal shades show higher
levels of light intensity (Figure 15). A further analysis by plotting light intensity and
translucency parameter indeed suggests that the shade has a strong effect on the light
intensity behind the disk but not on the translucency parameter (Figure 15). To test this
hypothesis, the translucency parameter of ceramic disks of differently external stained-
shades will be further measured at different thicknesses in the future.

In addition, by plotting of light transmittance and translucency parameter (Figure
17), we found a linear correlation between translucency parameter and light
transmittance, a relation that has never been documented before. These are two different
parameters. Translucency parameter is related to the reflectance of the light through the
medium and back with reflectance, internal scattering, and absorption. Transmittance
related to the attenuation of light after it passes through and exits the medium, a process
also involves reflectance, internal scattering, and absorption. The strong correlation
between the two parameters is shown between LDGC and full contour zirconia in this
study, two classes of very different materials that are only common in their shades. The
trend line of CAP QZ and CAP FZ without a specific shade is totally different from that
of the others with shade A2. The result indicated that CAP QZ and CAP FZ would have
higher transmittance than the first group of materials (E-max CAD, Zirlux, Bruxzir, and
KDZ Bruxer) at the same translucency value. Similarly, CAP QZ and CAP FZ will have
a greater translucency parameter at the same light transmittance.

According to the plotting of light intensity and degree of conversion (Figure 18),
we found that the general trend of correlation between the light intensity and degree of

conversion holds partially constant, except 1.5 mm of Bruxzir could not be detected for both light intensity and degree of conversion.

Therefore, in summary, the first part of the null hypothesis was rejected, and the second part was partially rejected.

SUMMARY AND CONCLUSIONS

- The translucency parameter of ceramic material has been influenced by the type of ceramic and its thickness.

- All of translucent zirconia has still lower translucency parameters and light transmissions than lithium disilicate glass ceramic.

- The higher thickness of any type of ceramic disk, the lower the translucency parameter. Also, the higher thickness reduced exponentially the light transmitting from the light source.

- There is no significant difference of the degree of conversion of light-cured mode of resin cement among the type of ceramic disks.

- The degree of conversion of resin cement has been reduced exponentially after the ceramic disk was increased from 1 mm to 2 mm in thickness.

- Only one full contour zirconia brand has still been inappropriately prepared for a clinical crown with only light-cured resin cement underneath a 1.5-mm ceramic thickness.

This study had limitations in its initial experiment and the ability to simulate oral environmental changes. The dual-cured resin, moisture condition and thermal cycling were not applied to stimulate the clinical situation. Further studies using the application of dual-cure resin cement as the clinical use need to be investigated.

REFERENCES

1.   McLean JW. Evolution of dental ceramics in the twentieth century. J Prosthetic Dent 2001;85:61-6.

2.   Conrad HJ, Seong WJ, Pesun IJ. Current ceramic materials and systems with clinical recommendations: a systematic review. J Prosthetic Dent 2007;98:389-404.

3.   Griggs JA. Recent advances in materials for all-ceramic restorations. Dent Clin North Am 2007;51:713-27, viii.

4.   Della Bona A, Kelly JR. The clinical success of all-ceramic restorations. J Am Dent Assoc 2008;139 Suppl:8s-13s.

5.   Goodacre CJ, Bernal G, Rungcharassaeng K, Kan JY. Clinical complications in fixed prosthodontics. J Prosthetic Dent 2003;90:31-41.

6.   Deany IL. Recent advances in ceramics for dentistry. Crit Rev Oral Biol Med 1996;7:134-43.

7.   Antonson SA, Anusavice KJ. Contrast ratio of veneering and core ceramics as a function of thickness. Int J Prosthodont 2001;14:316-20.

8.   Anusavice KJ, Phillips RW. Phillips' science of dental materials: Saunders; 2003.

9.   Carter B, Norton G. Ceramic materials:science and engineering. [S.l.]: Springer; 2012.

10.   Giordano R, McLaren EA. Ceramics overview: classification by microstructure and processing methods. Compendium Continuing Educ Dent 2010;31:682-4, 686, 688 passim; quiz 698, 700.

11.   Reiss B, Walther W. Clinical long-term results and 10-year Kaplan-Meier analysis of Cerec restorations. Int J Computerized Dent 2000;3:9-23.

12.   McLaren EA, Terry DA. CAD/CAM systems, materials, and clinical guidelines for all-ceramic crowns and fixed partial dentures. Compendium Continuing Educ Dent 2002;23:637-41, 644, 646 passim; quiz 654.

13.   Elsevier/Mosby. Craig's restorative dental materials. Available at: http://www.sciencedirect.com/science/book/9780323081085.

14.   Denry I, Holloway JA. Effect of heat treatment after accelerated aging on phase transformation in 3Y-TZP. J Biomed Mater Res B Appl Biomater 2010;93(1):236-243.

15.   Griggs JA. Recent advances in materials for all-ceramic restorations. Dent Clin N Am 2007;51:713-27.

16.   Kelly J. Dental ceramics: What is this stuff anyway? J Am Dent Assoc 2008;139:4S-7S.

17.   Borom MP, Turkalo AM, Doremus RH. Strength and microstructure in lithium disilicate glass-ceramics. J Am Ceramic Soc 1975;58:385-91.

18.   Albakry M, Guazzato M, Swain MV. Biaxial flexural strength, elastic moduli, and x-ray diffraction characterization of three pressable all-ceramic materials. J Prosthet Dent 2003;89:374-80.

19.   Albakry M, Guazzato M, Swain MV. Fracture toughness and hardness evaluation of three pressable all-ceramic dental materials. J Dent 2003;31:181-8.

20.   Gao J, Chen J-h, Wang F, Deng Z-x, Li F, Wu D. Effect of heat-pressing on the microstructure and properties of a novel lithium disilicate glass-ceramic. Advanced Mater Res 2011;177:441-6.

21.   Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of all-ceramic materials. (Pt 1). Pressable and alumina glass-infiltrated ceramics. Dent Mater 2004;20:441-8.

22.   Tinschert J, Natt G, Mautsch W, Augthun M, Spiekermann H. Fracture resistance of lithium disilicate-, alumina-, and zirconia-based three-unit fixed partial dentures: a laboratory study. Int J Prosthodont 2001;14:231-8.

23.   Beall G. Design and properties of glass-ceramics. Annual Review Mater Sci 1992;22:91-119.

24.   Kelly J. Dental ceramics: current thinking and trends. Dent Clin North Am 2004;48:513-30.

25.   Dong JK, Luthy H, Wohlwend A, Scharer P. Heat-pressed ceramics: technology and strength. Int J Prosthodont 1992;5:9-16.

26.   Rosenblum MA, Schulman A. A review of all-ceramic restorations. J Am Dent Assoc 1997;128:297-307.

27.   Mackert JR, Jr., Twiggs SW, Russell CM, Williams AL. Evidence of a critical leucite particle size for microcracking in dental porcelains. J Dent Res 2001;80:1574-9.

28.   Cattell MJ, Knowles JC, Clarke RL, Lynch E. The biaxial flexural strength of two pressable ceramic systems. J Dent 1999;27:183-96.

29.   El-Mowafy O, Brochu JF. Longevity and clinical performance of IPS-Empress ceramic restorations--a literature review. J Can Dent Assoc 2002;68:233-7.

30.   Della Bona A, Mecholsky JJ, Jr., Anusavice KJ. Fracture behavior of lithia disilicate- and leucite-based ceramics. Dent Mater 2004;20:956-62.

31.   Stappert CF, Att W, Gerds T, Strub JR. Fracture resistance of different partial-coverage ceramic molar restorations: An in vitro investigation. J Am Dent Assoc 2006;137:514-22.

32.   Oh SC, Dong JK, Luthy H, Scharer P. Strength and microstructure of IPS Empress 2 glass-ceramic after different treatments. Int J Prosthodont 2000;13:468-72.

33.   Esquivel-Upshaw JF, Chai J, Sansano S, Shonberg D. Resistance to staining, flexural strength, and chemical solubility of core porcelains for all-ceramic crowns. Int J Prosthodont 2001;14:284-8.

34.   Fasbinder DJ. Clinical performance of chairside CAD/CAM restorations. J Am Dent Assoc 2006;137 Suppl:22S-31S.

35.   Sorensen JA, Cruz M, Mito WT, Raffeiner O, Meredith HR, Foser HP. A clinical investigation on three-unit fixed partial dentures fabricated with a lithium disilicate glass-ceramic. PPAD 1999;11:95-106; quiz 108.

36.   Taskonak B, Sertgoz A. Two-year clinical evaluation of lithia-disilicate-based all-ceramic crowns and fixed partial dentures. Dent Mater 2006;22:1008-13.

37.   Wolfart S, Eschbach S, Scherrer S, Kern M. Clinical outcome of three-unit lithium-disilicate glass-ceramic fixed dental prostheses: up to 8 years results. Dent Mater 2009;25:e63-71.

38.   Heffernan MJ, Aquilino SA, Diaz-Arnold AM, Haselton DR, Stanford CM, Vargas MA. Relative translucency of six all-ceramic systems. (Pt 2) Core and veneer materials. J Prosthet Dent 2002;88:10-5.

39.   Sundh A, Sjogren G. A comparison of fracture strength of yttrium-oxide-partially-stabilized zirconia ceramic crowns with varying core thickness, shapes and veneer ceramics. J Oral Rehab 2004;31:682-8.

40.   Al-Wahadni AM, Hussey DL, Grey N, Hatamleh MM. Fracture resistance of aluminium oxide and lithium disilicate-based crowns using different luting cements: an in vitro study. J Contemporary Dent Pract 2009;10:51-8.

41.   Tinschert J, Zwez D, Marx R, Anusavice KJ. Structural reliability of alumina-, feldspar-, leucite-, mica- and zirconia-based ceramics. J Dent 2000;28:529-35.

42.   Conrad HJ, Seong W-J, Pesun GJ. Current ceramic materials and systems with clinical recommendations: a systematic review. J Prosthet Dent 2007;98:389-404.

43.   Odén A, Andersson M, Krystek-Ondracek I, Magnusson D. Five-year clinical evaluation of Procera AllCeram crowns. J Prosthet Dent 1998;80:450-6.

44.   Raigrodski AJ, Chiche GJ, Potiket N, et al. The efficacy of posterior three-unit zirconium-oxide-based ceramic fixed partial dental prostheses: a prospective clinical pilot study. J Prosthet Dent 2006;96:237-44.

45.   Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. Biomaterials 1999;20:1-25.

46.   Cales B. Zirconia as a sliding material: histologic, laboratory, and clinical data. Clin Orthopaed Related Res 2000:94-112.

47.   Lazar DR, Bottino MC, Ozcan M, et al. Y-TZP ceramic processing from coprecipitated powders: a comparative study with three commercial dental ceramics. Dent Mater 2008;24:1676-85.

48.   Sailer I, Feher A, Filser F, Gauckler LJ, Luthy H, Hammerle CH. Five-year clinical results of zirconia frameworks for posterior fixed partial dentures. Int J Prosthodont 2007;20:383-8.

49.   Silva NR, Thompson VP, Valverde GB, et al. Comparative reliability analyses of zirconium oxide and lithium disilicate restorations in vitro and in vivo. J Am Dent Assoc 2011;142 Suppl 2:4S-9S.

50.   Larsson C, Vult von Steyern P, Sunzel B, Nilner K. All-ceramic two- to five-unit implant-supported reconstructions. A randomized, prospective clinical trial. Swed Dent J 2006;30:45-53.

51.   Jiang L, Liao Y, Wan Q, Li W. Effects of sintering temperature and particle size on the translucency of zirconium dioxide dental ceramic. J Mater Sci Mater Med 2011;22:2429-35.

52.   Tsukuma K. Transparent titania-yttria-zirconia ceramics. J Mater Sci Letter 1986:1143-4.

53.   Zhang H KB, Morita K, Hiraga H, Sakka Y. Effect of sintering temperature on optical properties and microstructure of translucent zirconia prepared by high pressure sparl plasma sintering. Sci Technol Adv Mater 2011;12.

54.   Alaniz J, Perez-Gutierrez F, Aguilar G, Garay J. Optical properties of transparent nanocrystalline yttria stabilized zirconia. Optic Mater 2009;32:62-68.

55.   Noguchi T. Advance in high temperature chemistry 2.1967:274.

56.   Tsukuma K YI, Kusunose T. J Am Ceram Soc 2008:813.

57.   Anselmi-Tamburini U, Garay JE, Munir ZA. Fast low-temperature consolidation of bulk nanometric ceramic materials. Scripta Materialia 2006;54:823-8.

58.   Casolco SR, Xu J, Garay JE. Transparent/translucent polycrystalline nanostructured yttria stabilized zirconia with varying colors. Scripta Materialia 2008;58:516-9.

59.   Anselmi-Tamburini U, Woolman JN, Munir ZA. Transparent nanometric cubic and tetragonal zirconia obtained by high-pressure Pulsed electric current sintering. Advance Function Mater 2007;17:3267-73.

60.   Pecho OE, Ghinea R, Ionescu AM, Cardona Jde L, Paravina RD, Perez Mdel M. Color and translucency of zirconia ceramics, human dentine and bovine dentine. J Dent 2012;40 Suppl 2:e34-40.

61.   Wang F, Takahashi H, Iwasaki N. Translucency of dental ceramics with different thicknesses. J Prosthet Dent 2013;110:14-20.

62.   Paravina RD MISSING NAME? PJ. Esthetic color training in dentistry. St. Louis, Mo: Elsevier Mosby, 2004.

63.   http://www.konicaminolta.eu/en/measuring-instruments/learning-centre/light-measurement/light/light-sources-and-illuminants.html.

64.   Moser JB WW, Muller TP, Moore BK. Use of the munsell system to compute color differences in composite resins. J Dental Res 1978;57:958-63.

65.   Johnston WM, Ma T, Kienle BH. Translucency parameter of colorants for maxillofacial prostheses. Int J Prosthodont 1995;8:79-86.

66.   Brodbelt RH, O'Brien WJ, Fan PL, Frazer-Dib JG, Yu R. Translucency of human dental enamel. J Dent Res 1981;60:1749-53.

67.   http://en.wikipedia.org/wiki/Transparency_and_translucency.

68.   Yu B, Ahn JS, Lee YK. Measurement of translucency of tooth enamel and dentin. Acta Odontologica Scandinavica 2009;67:57-64.

69.   O'Brien WJ. Double layer effect and other optical phenomena related to esthetics. Dent Clin North Am 1985;29:667-72.

70.   Heffernan MJ, Aquilino SA, Diaz-Arnold AM, Haselton DR, Stanford CM, Vargas MA. Relative translucency of six all-ceramic systems.(Pt 1). Core materials. J Prosthet Dent 2002;88:4-9.

71.   Rosentiel SF PS. Color measurements of all ceramic crown system. J Oral Rehab 1989;1989:491-501.

72.   O'Keefe KL, Pease PL, Herrin HK. Variables affecting the spectral transmittance of light through porcelain veneer samples. J Prosthet Dent 1991;66:434-8.

73.   http://sensing.konicaminolta.us/2014/02/specular-component-included-vs-specular-component-excluded/.

74.   Calgaro PA, Furuse AY, Correr GM, Ornaghi BP, Gonzaga CC. Influence of the interposition of ceramic spacers on the degree of conversion and the hardness of resin cements. Braz Oral Res 2013;27:403-9.

75.   Lee IB, An W, Chang J, Um CM. Influence of ceramic thickness and curing mode on the polymerization shrinkage kinetics of dual-cured resin cements. Dent Mater 2008;24:1141-7.

76.   Soares CJ, da Silva NR, Fonseca RB. Influence of the feldspathic ceramic thickness and shade on the microhardness of dual resin cement. Oper Dent 2006;31:384-9.

77.   Passos SP, Kimpara ET, Bottino MA, Santos GC, Jr., Rizkalla AS. Effect of ceramic shade on the degree of conversion of a dual-cure resin cement analyzed by FTIR. Dent Mater 2013;29:317-23.

78.   Dewaele M, Truffier-Boutry D, Devaux J, Leloup G. Volume contraction in photocured dental resins: the shrinkage-conversion relationship revisited. Dent Mater 2006;22:359-65.

79.   Calheiros FC, Braga RR, Kawano Y, Ballester RY. Relationship between contraction stress and degree of conversion in restorative composites. Dent Mater 2004;20:939-46.

80.   Atai M, Nekoomanesh M, Hashemi SA, Amani S. Physical and mechanical properties of an experimental dental composite based on a new monomer. Dent Mater 2004;20:663-8.

81.   Ryan EA, Tam LE, McComb D. Comparative translucency of esthetic composite resin restorative materials. J Can Dent Assoc 2010;76:a84.

82.   Baldissara P, Llukacej A, Ciocca L, Valandro FL, Scotti R. Translucency of zirconia copings made with different CAD/CAM systems. J Prosthet Dent 2010;104:6-12.

83.   Zhang X, Wang F. Hardness of resin cement cured under different thickness of lithium disilicate-based ceramic. Chinese Med J 2011;124:3762-7.

84.   Flury S, Lussi A, Hickel R, Ilie N. Light curing through glass ceramics with a second- and a third-generation LED curing unit: effect of curing mode on the degree of conversion of dual-curing resin cements. Clin Oral Investigation 2013;17:2127-37.

85.   Kim MJ, Kim KH, Kim YK, Kwon TY. Degree of conversion of two dual-cured resin cements light-irradiated through zirconia ceramic disks. J Advance Prosthodont 2013;5:464-70.

86.   Ilie N, Hickel R. Correlation between ceramics translucency and polymerization efficiency through ceramics. Dent Mater 2008;24:908-14.

87.   Meng X, Yoshida K, Atsuta M. Influence of ceramic thickness on mechanical properties and polymer structure of dual-cured resin luting agents. Dent Mater 2008;24:594-9.

ABSTRACT

TRANSLUCENCY AND DEGREE OF CONVERSION OF

RESIN CEMENT WITH DIFFERENT THICKNESS OF

FULL CONTOUR ZIRCONIA

by

Noppamath Supornpun

Indiana University School of Dentistry
Indianapolis, Indiana

Background: Traditionally, zirconia has been used as a core material for all-ceramic crowns that are later covered by a more esthetic veneering layer. Recently, new zirconia materials with higher translucency commonly referred to as the "full contour zirconia" have been introduced with the aim to allow dentist to fabricate entire all-ceramic crown from the material with acceptable esthetic and mechanical functions without the need for veneering. However, there is little information in the literature regarding the translucency of full contour zirconia and the degree of conversion of resin

cement underneath the full contour zirconia. Objectives: 1) To investigate the translucency parameter (TP) of recently marketed full contour zirconia and compare that to traditional zirconia and lithium disilicate glass ceramic (LDGC) at different thicknesses. 2) To evaluate the degree of conversion (DC) of the resin cement through different thicknesses of the full contour zirconia, traditional zirconia and LDGC. Alternative hypothesis: The new generation zirconia at the clinically recommended thickness has lower translucency than that of LDGC and higher than that of non-veneered traditional zirconia. In addition, DC of resin cement under full contour zirconia is lower than that of LDGC and higher than that of traditional zirconia. Methods: 150 ceramic specimens (12 x12 mm with thickness of 1-2 mm for LDGC and Zirconia) were divided into 6 groups according to the type of material, as follow: LDGC (IPS e-max CAD), Traditional Zirconia (CAP QZ), full contour zirconia (CAP FZ, Zirlux, Bruxzir, KDZ Bruxer). The TP for materials at various thicknesses were measured by a spectrophotometer (CM-2600D). The DC of the light curing resin cement (Variolink II) underneath the ceramic disks was measured by FTIR. Result: All full contour zirconia has lower translucency parameter and light transmission than LDGC. The translucency parameter decreases with increasing thickness of any type of ceramic. There were no significant differences in the degree of conversion of resin cement among the type of ceramic disc, except Bruxzir. The correlation of TP between various thicknesses and the types of ceramic materials was established by a regression analysis.

CURRICULUM VITAE

Noppamath Supornpun

May 2003	DDS (second class honors) Mahidol University, Thailand
Apr 2003 to Dec 2004	General Practitioner The Dental Public Health Department of Khunhan community hospital, Ministry of Public Health, Srisakate, Thailand
May 2007	MS, Operative Dentistry Mahidol University, Thailand
May 2007 to 2012	Teaching assistant Operative Department of Mahidol University
Apr 2008 to Present	General Practitioner, Specialized in Operative Dentistry, and Instructor, Department of Medical Service, Ministry of Public Health, Thailand
Nov 2010 to 2012	Teaching assistant Operative Department of Rangsit University
July 2015	MSD, Prosthodontics, Indiana University School of Dentistry, Indianapolis, IN, USA

Professional Organizations

Thai Dental Council
Thai Dental Association
Thai Prosthodontics Association
Thai Operative Dentistry Society
Endodontic Society of Thailand
American College of Prosthodontics (ACP)
The John F. Johnston Society