EFFECT OF COPING DESIGN ON FRACTURE RESISTANCE OF ZIRCONIA CORE CERAMICS

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DEDICATION

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CHAPTER I

1. Background and Significance

The substantial increase in esthetic consciousness and patient demand for esthetic dental restoration has led to a rapid development in the art and science of restorative dental materials. Superior esthetic requirements are no longer a luxury. It is the everyday basic need that has pushed dental materials to the edge of its limitations. Newer materials must be developed to accommodate the ever demanding self-conscious patients that expect restorative work to mimic or quite often exceed the natural look of the combination of enamel and dentin found within human teeth.

Increased application of ceramic restorations has led to development of a variety of ceramic systems. Demand for improved clinical performance pressured the dental material industry to introduce several ceramic materials that are classified by porcelain type (feldspathic porcelain, leucite reinforced, aluminous, glass-infiltrated, glass-infiltrated spinell, glass-infiltrated zirconia and glass-ceramic), by porcelain use (denture teeth, metal ceramics, veneers, onlays, inlays, crowns, permanent prosthetic restorations), by porcelain processing method (casting, sintering, machining), by temperature fusion (high fusing - from 1315 ºC to 1370ºC, medium fusing - from 1090 ºC to 1260 ºC, low fusing - from 870 ºC to 1065 ºC) and by porcelain substructure material (cast metal, glass-ceramic, CAD/CAM (computer assisted design/computer assisted machining), sintered ceramic core).[1] Ceramic porcelain is essentially composed of silica, feldspar, kaolin and metallic pigments like opacifiers and color modifiers. Only the purest ingredients used in the dental field provide biocompatibility, low thermal expansion, insolubility, durability due to fracture and abrasion resistance as well as color stability and optical quality.

In many situations where the use of alloy or gold supported restorations was previously indicated, they are replaced today by tooth-colored materials. Any dental material requires sufficient physical properties to achieve good esthetic results, marginal integrity and high strength to withstand occlusal load. However, elimination of the metal substructure has raised concerns in resistance to fracture.

In an effort to improve strength, various core substrates were developed such as spinell (In-Ceram Spinell), alumina (eg. In-Ceram Alumina) and zirconia cores. These materials are becoming increasingly popular due to improved biocompatibility, physical, mechanical and esthetic properties.

Zirconium ceramics have current application in fabrication of endodontic posts, implant abutments, crowns, fixed partial dentures and orthodontic brackets as well as in other medical fields. Zirconia is composed of fine particles of ZrO_2 (zirconium oxide) and Y_2O_3 (yttrium oxide), having at room temperature monoclonic symmetry, forming, after the process of sintering (in the 1000-1100°C) a tetragonal structure that later transforms to a cubic phase that is characterized by high strength. Pure zirconia (unstabilized) is monoclinic at room temperature and transforms to a denser tetragonal form. This process promotes volume changes, and thereafter fractures.

The evolution of ceramics changed not only chemical and physical properties of materials but also quality and method of preparation and luting. Manufacturers have suggested different concepts of tooth preparation as well as specifications regarding coping design. The aim of these guidelines is stability and resistance to any occlusal stress and good esthetic results.

Currently the recommended width of zirconium coping supporting the veneered ceramic restoration is dependent on the manufacturer and ranges from 0.5-0.8 mm, with the shoulder covering the margins of the tooth preparation. Unfortunately the application of the recommended coping design contributes to an opaque and unnatural appearance, particularly at the cervical third of the restoration contour.

2. Purpose and Hypotheses

2.1. Purpose

The purpose of this study was to determine the effect of modified zirconium copings on the fracture resistance of Procera® All-Zircon.

The main features of the preparation design that will be evaluated in the research are:

Design 1 (control group) the coping material was fabricated according to the manufacturer recommendations - zirconia coping, extended to cover the complete shoulder.

Design 2 the zirconia coping was cut back to the axial wall of the preparation so that only veneering porcelain is on the shoulder.

Design 3 the zirconia coping on the lingual was fabricated according to the manufacturer recommendations, while the coping on the facial will be cut-back to the middle of the axial wall.

2.2 Hypotheses

2.2.1 Primary Hypothesis

Ho: The modification of the coping design does not significantly affect the fracture resistance of the zirconia core ceramic crowns.

Ha: The modification of the coping design significantly affects the fracture resistance of the zirconia core ceramic crowns.

2.2.2 Secondary Hypothesis

Ho: The modification of the coping design is not associated with the fracture location in the zirconia core ceramic crowns.

Ha: The modification of the coping design is associated with the fracture location in the zirconia core ceramic crowns.

2.3 Specific Aims

- To evaluate the effect of a shoulder-free zirconia coping design on the fracture strength of ceramic restorations.
- To evaluate the effect of shoulder-free zirconia coping design on the fracture location.

3. Literature Review

3.1 History of Ceramics

Porcelain history, named keramos meaning pottery in Greek started in early human civilization. About 100 B.C. China developed stoneware that was fired at higher temperatures improving its characteristics of strength and ability to withstand water penetration. It took another thousand years for porcelain to evolve to its current characteristics of high strength and translucency. Porcelain was introduced to dentistry in 1774 when Alexis Duchateau and Nicholas Dubois de Chemant fabricated the first successful all-porcelain dentures. In 1808, in Paris, Giuseppangelo Fonzi introduced individually-formed porcelain teeth containing embedded pins.[2] Early porcelain crowns were developed by Elias Wildman in 1838 and later improved by Richmond and Logan that made Davis crowns, and further, Land developed porcelain jacket crowns.[3] Initially, porcelain use was restricted to anterior dentition until recent advances modified the material to endure occlusal forces observed in posterior dentition. Various components were combined in precise proportions and a controlled firing process was developed to achieve biocompatibility, durability, stability, appropriate thermal expansion as well as translucency. A high-strength alumina coping, introduced in 1985 by Sadoun, contained over 85 % alumina and was intended for anterior and posterior single crowns as well as anterior three-unit bridges.[4] In 1993 Andersson manufactured a densely sintered alumina core for porcelain restorations, currently known as Procera All-Ceram Alumina.[5] Both of these exemplify the possible modifications that porcelain materials have undergone. In the 1988, Duret introduced CAD/CAM, an industrial machining process, into dentistry where numerous modifications over the following ten years brought about the Cerec system.[6, 7] The system unites computer technology, die-cutting tools and a prepable porcelain block. Further improvement in the strength of ceramic materials was achieved utilizing the zirconium particles. The terminology zircon, as the literature reports, comes from Arabic word zargon (golden in color) which in turn comes from the two Persian words zar (gold) and gun (color) that express the color of one of the gemstones. In the 1789 German chemist Martin Klaproth analyzed the gemstone named jargon and identified the metal dioxide in the reaction product obtained after heating some gems.[8] Zirconia is an alloy having numerous applications in daily basis in industry. The properties of the zirconia are extensively utilized in manufacturing vacuum tubes, surgical appliances, thermal isolation and many other daily used items.

3.2 Composition

Ceramics, from the finest porcelain to china composed of metallic oxides, have been used for centuries due to their stability, durability, biocompatibility as well as low thermal conductivity and good optical properties. Conventional dental porcelain is a vitreous ceramic based on silica (quartz (SiO_2)) and kaolin (clay) (Al2O3 $\cdot 2SiO_2 \cdot 2H_2O$), metallic pigments, potash feldspar ($K_2O \cdot Al_2O_3 \cdot 6SiO_2$), soda feldspar ($Na_2O \cdot Al_2O_3 \cdot 6SiO_2$) or both. The principal quality and difference is dependant on correct proportion of primary elements and proper firing procedure. The manufacture of the dental ceramics utilizes the purest ingredients to meet the requirements of abrasion and physical and mechanical properties in adjunct to color stability and translucency. Two phases are being distinguished during the procedure of blending the porcelain. [9] The vitreous phase (glass), formed during the firing process, presents typical glass properties as brittleness, high surface tension and non-directional fracture pattern. The crystalline phase (mineral) often named feldspar includes quartz or silica and metallic oxides. Chemically composed of potassium, aluminum silicate feldspar provides opaque color ranging from gray to pink. When heated, it retains the form and contours of the restorations and at 1290°C it fuses and becomes glassy. The form stability provides an important property for the fabrication of porcelain restorations. In the process of creation of fine particles for dental porcelain, feldspar undergoes several intricate processes that give homogenous light-colored pieces of the feldspar. Further grinding of the particles in a ball mill ultimately formats fine powder followed by the process of coarse particle elimination. To improve the color it is important to remove the iron, commonly found in feldspar. During that final step of slow vibration of the powder iron is picked up by narrow ledges formed by induction magnets.[10]

Pure quartz is ground to finer than feldspar powder particles and remains unchanged during the process of porcelain firing which contributes stability to the mass during heating by providing a framework for the other ingredients.[1] The pigments called "color frits" are added in small quantities to obtain slight shades for natural tooth color imitation. The process involves grinding together metallic oxides with fine glass and feldspar, fusing the mixture in a furnace, and regrinding to a powder. The metallic pigments used in that process are titanium oxide (yellow-brown), manganese oxide (lavender), iron or nickel oxide (brown), cobalt oxide (blue), copper or chromium oxide (green), tin oxide to obtain opacity and uranium oxide to increase fluorescence.[10] The exact composition of porcelain is commonly not provided by manufacturers while the published literature states that porcelain is a mixture of 75-85% feldspar, 12-22% quartz, 3-5% kaolin and a small percentage of pigments.[10] Progressive compositional changes have been made to bring firing temperature from 1290°C to current level of 900-980°C as well as to provide the higher strength and resistance to oral environment.

3.3 Classification of High-Strength All-Ceramics

High strength core ceramics is classified based upon chemical structure into main three groups; glass ceramics, glass-infiltrated ceramics and polycrystalline ceramics. [11]

3.3.1 Glass Ceramics

Glass ceramics are partially crystallized, amorphous glasses that are produced by enucleation and growth of crystals in the glass matrix phase.[12]

Albakry et al. tested the fracture toughness and hardness of three pressable all-ceramic materials: IPS-Empress, Empress 2 and an experimental ceramic material. Fifteen discs and 15 bars per material were prepared and fracture toughness was measured with two different techniques: indentation fracture and indentation strength.

During the indentation fracture tests the hardness of each material was also measured. The results in MPa showed mean fracture toughness using the indentation strength technique (with three-point bending and biaxial flexure tests): IPS-Empress (1.39 ± 0.3 and 1.32 ± 0.3); Empress 2 (3.14 ± 0.5 and 2.50 ± 0.3) MPa x $m^{1/2}$; and the experimental glass-ceramic (3.32 ± 0.6 and 2.43 ± 0.3) MPa x $m^{1/2}$. The indentation fracture technique generated orthogonal cracks of different lengths for Empress 2 and the experimental ceramic, whether perpendicular or parallel to the lithium disilicate elongated crystals. Thus, two values were reported: Empress 2 (1.5 ± 0.2 and 1.16 ± 0.2) MPa x $m^{1/2}$ and the experimental ceramic (1.67 ± 0.3 and 1.15 ± 0.15) MPa x $m^{1/2}$. The IPS-Empress indentation fracture result was 1.26 ± 0.1 MPa x $m^{1/2}$. The hardness results were: 6.6, 5.3 and 5.5 GPa for IPS-Empress, Empress 2 and the experimental ceramic, respectively. They concluded that there was no significant differences in fracture toughness and hardness between Empress 2 and the experimental ceramic ($P<0.05$ ANOVA). Both materials exhibited fracture toughness anisotropy following pressing and demonstrated improved fracture toughness and reduced hardness compared with IPS-Empress.[13]

3.3.1.1 Lithium Disilicate Glass Ceramics

The framework of lithium disilicate ceramics can be made by either lost-wax, heat-pressed technique, or milled out from prefabricated blanks. One of the ceramic materials belonging to that category is Empress II (Ivoclar) and it’s fracture toughness was evaluated in the study done by Quinn as 2.8 MPa [14]

In another study, Taskonak evaluated the site of crack initiation and the causes of fracture of clinically failed ceramic fixed partial dentures. Six Empress 2 lithium-disilicate veneered bridges and 7 lithium-disilicate non-veneered ceramic bridges were retrieved and analyzed using fractography and fracture mechanics methods. The analysis included failure in 6 bridges (50%) whose fracture initiated from the occlusal surface of the connectors and fracture of 1 non-veneered bridge (8%) initiated within the gingival surface of the connector. Three veneered bridges fractured within the veneer layers. Failure stresses of the all-core fixed partial dentures ranged from 107 to 161 MPa. Failure stresses of the veneered fixed partial dentures ranged from 19 to 68 MPa. It has been stated that fracture initiation sites were controlled primarily by contact damage.[15]
The application of lithium disilicate material includes not only single unit restorations but also short span three-unit fixed partial dentures extending up to the second premolar.[12]

In a two-year clinical evaluation Taskonak fabricated twenty anterior or posterior all-ceramic (Empress 2) crowns and twenty anterior or posterior, three-unit fixed partial dentures for 15 patients. Evaluations of the restorations were performed at baseline and once a year during the 2-year follow-up period that examined the marginal adaptation, color match, secondary caries and visible fractures in the restorations. Criteria showed 100% Alpha scores concerning recurrent caries for both crowns and FPDs and no crown fractures were observed during the 2-year follow-up, however, 10 (50%) catastrophic failures of FPDs occurred. Five (25%) failures occurred within the 1-year clinical period and the others (25%) within the second year.[16]

3.3.1.2 Leucite Reinforced Glass Ceramics

The IPS-Empress was introduced in the United States in 1991 and is a fine grained pressed ceramic material where leucite crystals are formed through various temperature cycles. These core materials can be fabricated by either heat-pressing procedure or via CAD/CAM technology. In heat-press method the system requires the restoration to be waxed to full contour invested and burnt-out in furnace at 800ºC. A ceramic ingot plunger is used in different shades that is heated at 1100ºC and pressed into the investment mold under 0.3-0.4 MPa pressure. The temperature is maintained for 20 minutes in a designed automatic press furnace.[17] The final IPS Empress microstructure contains 40% volume of 1-5µm leucite crystals dispersed in a glassy matrix.[17] Two finishing techniques can be applied including a staining or layering technique that lead both to comparable mean flexure strength values for the final restoration.[18] Dong et.al reported that flexural strength values range between 160 and 182 MPa and was found to be significantly improved after additional firings.[17] The reasonable strength and superior light transmission made IPS Empress a successful ceramic system for inlays, veneers and anterior crowns.[17]

3.3.2 Glass Infiltrated Ceramics

Glass infiltrated core ceramics is porcelain consisting of glass infiltrated to partially sintered oxides. That group is mainly represented by In-Ceram Alumina, In-Ceram Spinell and In-Ceram Zirconia. In-Ceram Spinell was marketed more recently to improve the esthetic potential. In vivo evaluation of specific esthetic parameters inherent to different types of cores was made and revealed the relative opacity of alumina while spinell was found to have the ability to blend in with the underlying substrate. Nevertheless both materials demonstrated a general lack of fluorescence.[19]

3.3.2.1 In-Ceram Spinell

The In-Ceram Spinell (Vita, D-Bad Sackingen) is glass infiltrated $MgAl_2O_4$. This ceramic is characterized with lower fracture and flexural strength (687.90 ± 90.26 N) than In-Ceram Alumina (876.19 ± 92.2 N), yet the translucency is higher therefore this type of ceramic is recommended in anterior areas, where higher esthetic result is required.[20, 21] The strength of that particular material depends on a successful restoration treatment and bond to the tooth structure and the survival rate in anterior region was shown to be 95% after 11 years.[22]

3.3.2.2 In-Ceram Alumina

The original In-Ceram material (In-Ceram Alumina) composed of sintered aluminum oxide subsequently infused with a glass, features interesting mechanical properties. Pure aluminum is a silvery-white metal possessing nonmagnetic, nonsparking property and has multiple applications: kitchen utensils, building décor, and other industrial purposes. A high purity alumina (85%) requires a firing temperature of 1750ºC, while 60% alumina may be fused at 1300 ºC. In 1964 Sandhaus first used alumina materials for tooth replacement and since then alumina has been proposed as a biomaterial and applied for socket and ball in hip replacement as well as used in the dental field. In-Ceram Alumina is a highly sintered alumina core ceramic, glass infiltrated and layered with veneering porcelain. Unfortunately, the alumina has limited esthetics due to semiopaque core therefore application is restricted and needs to be considered accordingly. [23]

3.3.2.3 In-Ceram Zirconia

In-Ceram Zirconia contains 35 % partially sintered stabilized zirconia with glass-infiltrated alumina. Its strength, according to Guazzato study, is much higher than In-Ceram Alumina. In-Ceram Zirconia mean flexural strength and fracture toughness are 580 ± 60 MPa and 4.0 MPa x $m^{1/2}$ respectively. In-Ceram Alumina characterized with values 520 ± 55 MPa and 3.2 MPa x $m^{1/2}$. [24] The process of fabrication is achieved either by milling or slip-cast technique. Due to higher material strength and fracture toughness the material is recommended in anterior, posterior single replacement as well as FPDs.[20, 25, 26]

Evaluation of biaxial flexural strength (piston on three ball), Weibull modulus, hardness, and fracture toughness of In-Ceram Zirconia and In-Ceram Alumina using indentation fracture and indentation strength methodology revealed that mean biaxial flexure strengths of In-Ceram Alumina and In-Ceram Zirconia were 600 MPa (SD 60) and 620 MPa (SD 61), respectively. Ninety-four disks and six bars were prepared with the slip-casting technique. The disks were used to assess biaxial flexural strength (piston on three balls), Weibull modulus, hardness, and fracture toughness with two methods: indentation fracture and indentation strength. The bars were used to measure elastic moduli (Young's modulus and Poisson's ratio). Mean fracture toughness measured according to indentation strength was 3.2 MPa.m1/2 (SD 0.34) for in-Ceram Alumina and 4.0 MPa.m1/2 (SD 0.43) for In-Ceram Zirconia, while mean fracture toughnesses of In-Ceram Alumina and In-Ceram Zirconia measured according to indentation fracture were 2.7 MPa.m1/2 (SD 0.34) and 3.0 MPa.m1/2 (SD 0.48), respectively. [25]

A study where Guzzato tested the strength, fracture toughness and microstructure of DC Zirkon, In-Ceram Zirconia slip, an experimental yttria partially stabilized zirconia, and In-Ceram Zirconia dry-pressed presented means of strength (MPa) and fracture toughness (MPa m(1/2)) values and their standard deviation: In-Ceram Zirconia dry-pressed 476 (50)1, 4.9 (0.36)1; In-Ceram Zirconia slip 630 (58)2, 4.8 (0.50)1; the experimental yttria partially stabilized zirconia 680 (130)2, 5.5 (0.34)2; DC-Zirkon 840 (140)3, 7.4 (0.62)3. Strength was appraised on ten bar-shaped specimens for each material (20 x 4 x 1.2 mm) with the three-point bending method. The fracture toughness (Indentation Strength) was measured on twenty specimens (20 x 4 x 2 mm) for each ceramic. The volume fraction of each phase, the dimensions and shapes of the grains and the crack pattern were investigated with SEM. The author postulated that the zirconia-based dental ceramics are stronger and tougher materials than the conventional glass-ceramics and better properties can have positive influence on the clinical performance of all-ceramic restorations. [27]

The purpose of Suarez's study was to evaluate the clinical performance of In-Ceram Zirconia posterior fixed partial dentures (FPD) after 3 years in service. He fabricated eighteen In-Ceram Zirconia FPDs for sixteen patients. The CDA quality evaluation system was used for assessment of surface and color, anatomic form, and marginal integrity and bleeding on probing was also recorded. The results report that only one of the 18 posterior FPDs was lost because of a root fracture and all remaining FPDs were rated as either excellent or acceptable after the observation period.[28]

3.3.3 Polycrystalline Ceramics

Polycrystalline ceramic material is composed of densely sintered particles with no glassy components and is solely processed by CAD/CAM technology. CAD/CAM stands for "Computer-Aided-Design/Computer-Aided-Manufacturing", and designates the three-dimensional planning of a workpiece on the screen of a computer with subsequent automated production by a computer controlled machine tool. [29] CAD/CAM processing was introduced to dentistry by Francois Duret in 1971 and has received considerable clinical and research interest from modern dental practices as a means of delivering all-ceramic restorations. Up to now the CAD/CAM system with zirconia has the highest fracture strength of all all-ceramic materials, and consistently produced the most esthetic, lifelike reproduction of natural dentition. They have been widely received by both dentists and patients.[30, 31] The contemporary CAD/CAM systems consist of three components: the scanner, software, and hardware. The material used in fabrication of restorations can be different: silicate ceramics; glass-infiltrated aluminum oxide; densely sintered aluminum oxide; densely sintered zirconium dioxide, manufactured at green stage, presintered stage and completely sintered stage; hipped zirconium dioxide; titanium; precious alloys; nonprecious alloys. [32]

Procera All-Ceram (Nobel Biocare, S-Goteborg) is a polycrystalline ceramic made of densely sintered high-purity (99.9%) aluminum oxide core with 500ppm MgO and was developed in 1993 by Matts Andersson and colleagues. [33, 34] With the Procera milling machine changes to the configuration of the preparation can be made, copies with a positive or negative offset of the surface can be produced, and the stone die can be replicated in a suitable material. During the coping fabrication linear expansion ranging between 12 to 20% occurs allowing the gap width between the crown and prepared tooth to be controlled and compensate for the shrinkage during the sintering process. [5] The milling process is started with alumina powder compaction using the industrial pressing technique against the enlarged replica. The compacted alumina is pre-sintered to a “green stage” and subjected to sintering process at 1550ºC followed by cooling and grinding procedure to achieve predetermined dimension. Procera All-Ceram has flexural strength between 500 and 650 MPa, fracture toughness of 4.48-6MPa x m $^{1/2}$ and mean grain size 4μm.[35, 36]

Yttrium tetragonal zirconia polycrystals (Y-TZP) is a glass-free, high polycrystalline ceramic material containing about 3% mol Y_2O_3 with a flexural strength from 900 to 1200MPa and fracture toughness of 9 to 10MPa x m $^{1/2}$. [37]

Castellon in his case report tested in vitro ceramic copings for fatigue and compression of six tooth shaped copings and several luting agents and found out that crown endurance limits for fatigue compression were 70% higher and 46% higher, respectively, than the established minimum fatigue endurance limits in those categories, The study confirms that the material performs well and produced excellent results.[38]

The majority of the Y-TZP –based CAD/CAM systems use CAM of partially sintered Y-TZP blanks: Lava (3M ESPE Dental AG, Seefeld); Cercon (DeguDent, Hanau); Cerec InLab (Sirona Dental Systems, Bensheim); Procera All-Zircon (Nobel Biocare, S-Goteborg). The milling of these blanks is faster and results in less wear and tear to the hardware .[39] With fully sintered blanks, such as DC-Zircon (DCS-Precident, DCS Dental AG, CH-Allschwill), there is no shrinkage involved in the milling process, but microcracks may be introduced to the infrastructure. [40]

According to Razzoog et al. zirconia coping can be fabricated by either waxing the crown to full contour and then cutting back to the desired thickness or by creating a suitable resin pattern. After the abutment is placed on an analog that is secured in a holder a sapphire probe contacts the abutment and records the data. The probe ascends 200 μm per revolution until the highest point of the abutment is reached. After the image is created by merging abutment and coping files together, the files are sent to a manufacturing center to have the coping finalized.[41]

3.4 Fracture Resistance Studies

The physical properties of any new dental ceramic must be tested in-vitro prior any clinical application. Numerous studies have been carried out to evaluate the fracture resistance of different dental ceramic materials by using various testing methods, sample dimensions, and testing conditions. Ceramic hardness similar to that of enamel is desirable to minimize the abrasiveness and the wear of resulting ceramic restorations, and reduce the wear damage that can be produced on enamel by ceramic restoration. Porcelain demonstrates excellent insulating properties, such as low thermal conductivity, low electrical conductivity and low thermal diffusivity.[1] On the other hand, the brittleness, particularly when flaws and tensile stresses coexist in the same region of the restoration, is a commonly known drawback. When tensile stress is applied, small flaws tend to open up and propagate cracks. The flaw could be a microcrack on the surface that is created by a diamond bur while adjusting the ceramic, corrosion and surface diversification or it can be a subsurface porosity from the processing flaw and error during firing cycles.[42]

Discontinuities or any irregularities in the body of the porcelain, or abrupt changes in the shape of the restoration promote stress and serve as a stress inducer. The amount of that increased stress depends on the shape of the irregularity. The main cause of such flaws, according to Griffith’s fracture theory, are stress concentrations formed around small flaws and are high around cracks since the ceramics lack the ductility to deform and reduce sharp angles.[43] The stress concentration as surface defects results in ceramics that fails at stress levels much lower than theoretical. While the metal yields the stress by deformation due to plasticity of the material, ceramics lacks that property and results with a fracture.[42]

Stress is termed as the reaction to externally applied forces and is equal to intensity, but opposite in direction, to the external force. Stress may occur with compression, tension, or shearing forces and are dispersed over a given area. Ceramics have weaker properties in tension or transverse loading than in compression. Furthermore, the largely covalent and/or ionic bonded structure of ceramics results in their resistance to chemical degradation in the oral environment, but also imparts brittleness. Dental porcelain also has a limited capacity to withstand the stress at a nominal temperatures.[44] Tensile or bending stresses promote the crack extension whereas compressive stress tends to inhibit crack propagation.[45] Porcelain failure intraorally occurs by a combination of tension and bending forces on the crown. These involve tensile stresses, upon light occlusal loading on the intaglio surfaces of the restoration mainly at the cervical third.[46]

A wide range of ceramic materials have a critical strain fracture that ranges from 0.05 to 0.2%, therefore to improve the strength of ceramics the elastic modulus needs to be ameliorated. [47] Batchelor and Dinsdale and Binns demonstrated that after introduction into glass of the crystalline grains of high strength and elasticity, the strength and modulus of elasticity of the mixtures increased gradually with the proportion of the crystalline phase.[48, 49] Studies also demonstrated that in that type of system, crack propagation was present through both glass and crystal phases, therefore the energy evolving the crack had to be higher than one required to fracture the glass phase alone. The strength of dental ceramics may also be influenced by the presence of residual stresses existing in the porcelain as a result of uneven cooling of the fused porcelain or difference in coefficients of thermal expansion among different layers of porcelain fused together. Residual stresses existing on either the outer layer of porcelain or in the porcelain along the ceramic/metal interface will inhibit crack initiation and increase strength.[50]

While evaluating the strength of ceramics it is important to consider the mechanical fatigue of the material. Mechanical fatigue has been defined by the American Society of Testing Materials (1979) as “The process of progressive localized permanent structural change occurring in a material to conditions which produce fluctuating stresses and strains at some point or points and which may culminate in cracks or complete fracture after sufficient number of fluctuation”.[51]

Fracture strength can be described as a stress at which material tends to fracture. The most critical factors restricting the resistance to fracture are size and distribution of load and fracture toughness. Nevertheless, fracture strength is a helpful parameter in evaluation of the fracture resistance of ceramic materials.

The methods which have been used for the measurements of strength of dental ceramics are varied and diverse. Different test pieces, including bars, discs, rods, cylinders and crown-shaped specimens have been utilized. The following tests are the methods which have been described in the dental literature for strength measurements of ceramic materials.

3.4.1 Three-point Bending Test

Three-point bending test is most commonly used strength test due to the fact that it is the most sensitive and reliable laboratory test for dental ceramic materials. The results of the fracture strength are described as the transverse strength, modulus of rupture or flexure strength and are presented using MPa. [52, 53]

Shimizu et al. tested 2.5-3.0 mol % Y_2O_3 partially stabilized zirconia by implanting them in seventy-eight rabbits. The study was conducted to examine time dependent changes in the phase-transformation rate and bending strength of new zirconia ceramics *in vivo* as well as in various *in vitro* environments. The material was obtained by sintering at 1300-1400°C using a material with an addition of 2.5-3.0 mol % Y_2O_3 to stabilize the tetragonal phase. The bulk density and average grain size range was from 5.95-6.0 and from 0.6-0.25, respectively. Four pieces of ceramic were placed in medullary cavity of the upper end of the bilateral tibia, two on each side separately by making a drill hole. Four other pieces of each ceramic were placed subcutaneously in the backs of rabbits by stable incisions. Three zirconia test pieces were obtained from a rabbit 30 months after the operation and subjected to mechanical tests series. Three-point flexion method was used to measure the bending strength of an 8 mm span at a crosshead speed of 0.5 mm/s. The initial strength was above 1000 MPa in vitro and values of more than 700 MPa were determined for all probes after a period of 3 years in vivo.[54]

In the same bending test, Ichikawa et al. demonstrated measured values of approximately 1300 MPa 12 months after implanting cylindrical zirconia ceramics subcutaneously in rats. Zirconia was made from 97 mol% of zirconium oxide and 3 mol% of yttrium oxide and a small amount of alumina and silicon dioxide and molded into a cylinder shape at 1500º C by casting. Its crystal size was 0.4 µm, and crystallinity was approximately 100 %. Alumina was used as a control group. Each specimen of zirconia and polycrystalline alumina was cylinder shaped, 2.0 mm in diameter, and 10.0 mm in length, without any sharp edges. Zirconia ceramic cylinder was implanted in each of the right subcutaneously prepared pockets, and a polycrystalline alumina cylinder as a control was implanted in each of the left pockets. The site of incision was sutured. Some specimens were kept in the physiologic solution of sodium chloride (pH 5.0 to 7.0, 5.0 ml) without periodic changes (37º C) and in the air as a control in the room at a controlled 37º C during the experiment to evaluate the change of weight and mechanical properties in vivo. Animals were sacrificed at 3, 6 and 12 months after implantation; excised specimens were stained and examined under the light microscopy. Five blocks without fixation at 12 months after implantation were used for evaluation using three-point flexion using a bending strength of a 7.0 mm span at a cross-head speed of 1.0 mm/s. The results suggested that zirconia ceramic specimens are tissue compatible and no signs of degradation were observed and which was shown to be twice that of polycrystalline alumina.[55]

3.4.2 Four-point Bending Test

Similarly to three-point test transverse testing is performed using a four-point loading jig. Due to the lack of the shear stresses in the central position of the beam the latter test is generally preferable.[56]

A study by Tinschert et al., featuring different industry and laboratory-developed ceramic materials, demonstrated that zirconium TZP achieved the best results in the four-point flexural strength with 913.0 ± 50.2 MPa. The study material consisted of eight ceramic materials, six core materials and two veneering ceramics (Cerec Mark II, Dicor, In-Ceram Alumina, IPS Empress, Vitadur Alpha Core, Vitadur Alpha Dentin, Vita VMK 68, Zirconia-TZP). Thirty bar specimens per material were prepared and tested. All bar-shaped specimens were fabricated to predetermined dimensions, polished using abrasive papers and flexure strength was determined using four-point bending test. For each type of ceramic material, the fracture stress was evaluated for a total of 30 specimens per group. A computer program was used to calculate the Weibull modulus and the strength at failure probabilities of 1 and 5%. Two-parameter Weibull distribution was used to analyze the fracture stress values of the ceramic materials. The mean strength and standard deviation values for these ceramics (MPa+/-SD) were as follows: Cerec Mark II, 86.3+/-4.3; Dicor, 70.3+/-12.2; In-Ceram Alumina, 429. 3+/-87.2; IPS Empress, 83.9+/-11.3; Vitadur Alpha Core, 131.0+/-9.5; Vitadur Alpha Dentin, 60.7+/-6.8; Vita VMK 68, 82.7+/-10.0; and Zirconia-TZP, 913.0+/-50.2. There was no statistically significant difference among the flexure strength of Cerec Mark II, Dicor, IPS Empress, Vitadur

Alpha Dentin, and Vita VMK 68 ceramics ($p>0.05$). The highest Weibull moduli were associated with Cerec Mark II and Zirconia-TZP ceramics (23.6 and 18.4). Dicor glass-ceramic and In-Ceram Alumina had the lowest Weibull modulus m values (5.5 and 5.7), whereas intermediate values were observed for IPS-Empress, Vita VMK 68, Vitadur Alpha Dentin and Vitadur Alpha Core ceramics (8.6, 8.9, 10.0 and 13.0, respectively). It has been affirmed that except for In-Ceram Alumina, Vitadur Alpha and Zirconia-TZP core ceramics, most of the investigated ceramic materials fabricated under the condition of a dental laboratory were not stronger or more structurally reliable than Vita VMK 68 veneering porcelain. Only Cerec Mark II and Zirconia-TZP specimens, which were prepared from an industrially optimized ceramic material, exhibited m values greater than 18. [57]

Jung et al. evaluated the decrease in strength and fatigue properties in water of a feldspathic ceramic, a glass-infiltrated aluminum oxide ceramic and a tetragonal zirconia ceramic stabilized with approximately 3 mol% yttrium. Loaded to fracture in a four-point bending test, yttrium-stabilized zirconia yielded the best results. Bar specimens 3 x 4 x 25 mm of different ceramic materials (Vita Mark II, MGC, Vita Celay In-Ceram, Y-TZP) were cut from blocks and polished with diamond paste. Samples were centrally aligned along the load axis and subjected to indentation test with a tungsten carbide sphere with a radius of $r=3.18$ mm mounted to a universal testing machine. Cyclic test was carried out at frequency $f = 10$ Hz, in haversinusoidal wave form. The load was cycled between a specified maximum (200N to 3000N) and small but non-zero minimum (<20 N). Some static tests over a prescribed hold time at maximum load P-500N were conducted for comparative purposes. A minimum of five specimens were indented at each given load and number of cycles. Selected porcelain, MGC, and alumina specimens were sectioned from the back surface to a final thickness = 0.5 mm with a 10-μm-grit diamond wheel followed by 1-μm diamond paste. The thinned, translucent specimens were then viewed in transmission optical microscopy, which highlighted any subsurface cracks. The latter, observational procedure was not useful for the Y-TZP, owing to the relatively high opacity of this material. Indented specimens were then placed in a four-point bend fixture (inner span, 10 mm; outer span, 20 mm) with the damage site centrally located on the tensile side. Indentation sites were dried and covered with a drop of silicone oil and the specimens were then broken in fast fracture (time to fracture < 40 ms). Optical microscopy revealed that multi-cycle-sphere contact loading on the surface promotes the cumulative damage even at loads considerably lower than those needed to produce single-cycle degradation thus limiting the useful life of the structure. Described deterioration occurs in all tested materials, most rapidly at lower contact loads in the esthetic ceramics (porcelain and MGC) but even to some degree in stronger materials like glass-infiltrated alumina and Y-TZP. Comparative fatigue tests on the porcelain confirm that cyclic loading is much more deleterious than static loading under conditions of equivalent hold time at the same maximum load. Degradation occurs in the porcelain and MGC after 10^4 cycles at loads as low as 200 N; comparable degradation in the alumina and Y-TZP requires loads higher than 500 N, well above the clinically significant range.[58]

3.4.3 Shell Test

The shell test is a modified bending test, first introduced by Sced et al. in 1977.On prepared platinum foils adapted to an end of metal cylinders Sced used disc-shaped specimens resting on a circular knife-edge support that were tested to failure by center loading with a spherical indenter.[59] Disc samples that are tested can possess a surface to volume ratio which is closer to that of actual crowns than conventional three-point bending test, therefore a shell test is sensitive to surface conditions. [60]

3.4.4 “C” Test

The “C” test, a standard test used for metallic materials was utilized by Tan in testing of industrial ceramics as well.[61] Edward and Jacobsen evaluated the effect of surface treatments on the strength of porcelain made of aluminum and found that the specimens constructed using tin-oxide-coated foil did not demonstrate greater strength that those not coated with platinum.[62]

3.4.5 Brittle Ring Test

The pioneers of the brittle test were Bortz and Lund who used that test for testing engineering ceramics. The cylindrical test pieces were constructed on metal master die with a minimal taper to facilitate removal, and after sintering specimens were loaded diametrally. [63]

3.4.6 Fracture Resistance of Crown-shaped Specimens Stressed Diametrally

Diametric compression test on crown samples were first invented by Hondrum when he machined a stainless steel die to dimensions of a porcelain crown preparation on a maxillary bicuspid. The study was divided into two experiments. In the first one, crown-shaped specimens of uniform dimensions were constructed and then loaded to fracture in a diametral fashion.The second experiment involved constructing crown specimens of uniform dimensions, but then cementing the crowns onto dies and axially loading them to fracture in a manner similar to that in oral cavity. He used sixty crown-shaped specimens and for each of them platinum foil was adapted to the die after standard technique. All foils were annealed in a flame both before and after adaptation to the die. The study tested magnesium oxide core porcelain with Ceramco Vacuum veneering porcelain, and aluminum oxide core with Vitadur veneering porcelain. Six tested groups were as follows: aluminum oxide/magnesium oxide core with foil removed before test, aluminum oxide/magnesium oxide core with foil remaining in crown during test, aluminum oxide/magnesium oxide core internally glazed. All crowns were tested for fracture strength by being compressed diametrally at a crosshead speed of 0.5 mm/min on an Instron Universal Testing Machine until fracture. Analysis of data revealed that loading crown –shaped samples had similar values to modulus of rapture test and the crown specimens glazed with magnesium oxide core were 50% more resistant to fracture stress than conventional porcelain restorations with either alumina or magnesia core alone.[64]

Due to the fact that dental porcelain is brittle and has limited tensile strength, it is subjected to time dependent stress failure. The lifetime of the ceramics may be attributed to the presence of microdefects within the material and to degradation in oral environment that results in crack propagation. Therefore, it is of increasing interest to investigate fracture resistance of dental ceramics.

In a two-year clinical study Genho et al. evaluated Procera All-Ceram crown performance. Fifty-nine single unit full coverage Procera/All-Ceram crowns were cemented with RelyX (3M) on vital molars in 54 patients by 19 practicing clinicians using standardized procedures for preparation design and cementation according to manufacturer's directions. Standardized evaluations were performed using modified Ryge scales at initial placement, 6 months, 1 year, and 2 years. Criteria graded in-vivo were: color match, interproximal contact, caries, post-operative sensitivity, and gingival health. Criteria graded in-vitro were: surface smoothness, presence of pitting, presence of occlusal adjustments, breakage, and quantitative wear (CRA Measurement System. JDR 69:126 #140 ‘90). Analyzed across time, significant criteria were: breakage and pitting, while non-significant criteria were: sensitivity, caries, surface smoothness, gingival health, color match, interproximal contacts and occlusal adjustment. The results showed that 3% of Procera/All-Ceram exhibited bulk fractures; 23% exhibited small chips and the majority of small chips were found on marginal ridges and 85% exhibited surface pitting. At the 2-year recall, 100% of the copings were intact and 100% of the crowns remained cemented. The analysis revealed that 86% required occlusal adjustments after cementation, which affected surface smoothness adversely. Mean 2-year cumulative wear was 45 mm. These data indicated Procera/All-Ceram full crowns cemented with RelyX can provide a viable treatment option for patients with metal allergies and/or concerned with metal use in their treatment. [65]

Neiva et al. compared in vitro the load to fracture using three bonded all-ceramic systems: IPS-Empress, In-Ceram and Procera All-Ceram. Thirty dies were replicated from a master die, simulating the preparation on a maxillary premolar, using high filler resin with a modulus of elasticity similar to dentin. Ten cores each of In-Ceram and Procera were fabricated to a thickness of 0.5 mm. The remaining porcelain was applied using a sculpting device to produce a crown with a final thickness of 1.0 mm axially and 2.5 mm occlusally. Ten IPS Empress crowns were made to the same dimensions and pressed by the manufacturer. The internal surfaces of all the crowns were subjected to etching and silanization procedure followed by cementation with resin cement (Panavia 21). The cemented samples were stored in 100% humidity for 24 hours and then loaded in an Instron machine at a crosshead speed of 0.5 mm/minute until fracture occurred. A hard stainless steel ball bearing, 4mm in diameter, was centered on the occlusal surface of each specimen and stabilized with utility wax. The mean fracture loads were: IPS Empress, 222.45 (+/- 49) kg; In-Ceram, 218.8 (+/- 36) kg; Procera AllCeram, 194.20 (+/- 37) kg. The optical microscope verifying the thickness of the crowns demonstrated larger gap size for Procera AllCeram, especially at the marginal opening and axial wall. The statistical analysis of fracture strength showed no differences among the three all-ceramic systems at $p < 0.05$. [66]

Quinn et al. measured fracture toughness for several dental ceramic groups to determine if chemistry and microstructure affects ceramic properties. A fully-articulating four-point flexure fixture, self-aligning in three dimensions, was used with a 40 mm outer (support) span and 20 mm inner (loading) span for the 3 mm×4 mm×50 mm machinable glass ceramic (MGC) specimens. The broken, 3 mm×4 mm×25 mm MGC specimen pieces were then precracked and retested using shorter spans. The same fixture was used with blocks positioned behind the roller stops to decrease the outer span size to 20 mm and the inner span to 10 mm. These shorter spans (20 mm outer, 10 mm inner) were also used to test the 3 mm×4 mm×25 mm zirconia, glass-infused alumina, and Mark II porcelain specimens. K_{Ic} was calculated from the fracture load, specimen size and measured precrack size. The data revealed that large increases to fracture toughness were largely associated with material crystallinity, large grain size and high aspect ratios. Fracture toughness (KIc) values were obtained using Single Edge Precracked Beam (SEPB) and Single Edge V-Notch Beam (SEVNB) methods. Dynamic Young's modulus, which often scales with strength and has been used in explaining the microstructure/toughness relationship on a theoretical basis, was also obtained for the three groups of materials comprising this study. The first group, consisting of micaceous glass ceramics, included model materials that varied systematically in microstructure but not in chemistry. The second group, the feldspathic porcelains, varied significantly in microstructure, but little in chemistry. The ceramics comprising the third group were significantly different in both chemistry and microstructure. Upper toughness limits for the micaceous glass-ceramics and feldspathic porcelains were significantly raised compared to the base glasses, but remained under 2 MPa m(1/2). The highest toughnesses were associated with high percent crystallinity, large grains and high aspect ratios. The third group KIc values were 2.8 MPa m $^{1/2}$ for a lithium disilicate glass-ceramic, 3.1 MPa m (1/2) for a glass-infused alumina, and 4.9 MPa m $^{1/2}$ for zirconia. From a practical standpoint, microstructure effects were found to be important, but only within a limited range; the chemistry apparently defined a band of achievable property values. This suggests very large increases in fracture toughness are unlikely to be attained by changes in microstructure alone. A functional relationship determined for the micaceous glass-ceramics enables quantitative predictions of fracture toughness based on the microstructure. [14]

Webber et al. conducted research investigating the effect of different thicknesses of veneer porcelain on the compressive load at fracture of Procera AllCeram crowns. They fabricated sixty brass dies with a crown-like preparation and a chamfer margin. Sixty crowns were made for prepared dies with a 0.6-mm-thick core: Procera crowns with either a 0.4 mm- or 0.9 mm-thick veneers of AllCeram (Groups 1 and 2 respectively) and In-Ceram crowns with a 0.9 mm-thick veneer of Vitadur Alpha porcelain (Group 3). Each group consisted of 20 crowns. In-Ceram crowns were used as the control group. All crowns were measured at 4 axial and 1 occlusal random locations before autoglazing, air abrading and adhesively bonding onto the appropriate brass die using Clearfil Newbond

Bonding Agent, Clearfil Porcelain Bond Activator, and Panavia 21 TC Dental Adhesive as the luting agent. After storage in distilled water at 37°C for 24 hours, the specimens were placed in a compressive test rig within an Instron universal testing machine and loaded in the center of the occlusal surface with a 4-mm diameter stainless steel ball. An axial preload of 20 N was applied before compressive testing at a crosshead speed of 0.1 mm/minute until fracture occurred and an analysis of variance revealed no significant difference in the load at fracture between the 3 groups ($P < 0 .05$). The mean load at fracture for Group 1 was 2197.6 N (SD = 776.4); Group 2, 2401.4 N (SD = 699.1); and Group 3, 2581.0 N (SD = 715.6). The authors concluded that the axial thickness of veneer porcelain did not have a significant effect on the compressive load at fracture of tested Procera AllCeram crowns. [67]

3.5 Preparation Design

Although there is no standard preparation design for all-ceramic restorations due to the variety of materials available on the market, there is a necessity of occlusal reduction of 2 mm and finish line with deep rounded chamfer ranging in thickness from 1.0 to 1.5 mm. Nevertheless, smoothed tooth preparation margins seem to be mandatory and recommended in all-ceramic FPD. The classic ideal convergence angle as described by Shillingburg is very hard to achieve and, if achieved, parallel walls make it difficult to seat the restoration.

Sato studied the preparation performed by students in Tokyo Medical and Dental University who were trained to prepare teeth with a wall taper of 2º to 5º. He examined the abutment taper of final-year students and traced shadowgraphs of 63 working study dies. The results show that 12.7 % fell within the ideal range, and that the average convergence taper was 9.5º which means that they had a convergence angle of up to 19º. However, it was concluded that the taper was clinically acceptable due to the high difficulty in intra-oral environment to accomplish the ideal taper.[68]

Christensen suggested that crown preparations should have minimal divergence from parallelism, approximately 10º, while Dodge at al. concluded that a 16º convergence angle is acceptable.[69, 70]

3.6 Bonding Veneering Porcelain to Ceramic Core

A ceramic-metal bond may fail in several locations: metal-metal oxide, metal-oxide ceramic or ceramic-ceramic surface. Good adhesion of porcelain to metal depends on proper wetting, adherent oxide and mechanical retention. For metal-ceramic restorations a technique where a thin layer of opaque porcelain is applied allowing creating a glass surface that bonds with a metal. The fracture mechanism of ceramics and metals are different due to diverse structure and bonding. Ceramic restorations present with covalent and ionic bonds hence resistant to plastic deformation.[71] In core ceramic techniques, the core is lightly abraded using aluminum oxide air to break the glaze followed by application of porcelain that will wet the abraded surface.[72] Due to the fact that both core and veneering porcelain are brittle materials it is inevitable to match both in mechanical and physical properties to prevent from delamination. Strength, coefficient of thermal expansion, and oxidation time play an important role in fracture behavior of prosthesis. [73]

An increasing demand for esthetic restorations has resulted in the development of new ceramic systems, but the fracture of veneering ceramics still remains the primary cause of failure.

The aim of the Aboushelib et al. study was evaluation of the core-veneer bond strength and the cohesive strength of the components of Cercon, Vita Mark II and Empress 2 ceramics as well as the effect of an optional liner material between the core and veneer where applicable. Bilayered zirconia veneer discs were fabricated from five layering and two pressable veneer ceramics and additionally, discs from each veneer ceramic were prepared. The discs were cut into microbars of 6mm in length and 1mm in cross-section followed by the microtensile bond strength test in a universal testing machine. The fracture surfaces of the microbars were examined with scanning electron microscopy (SEM) and EDAX. The microtensile strength of Rondo Dentine and Lava Dentine veneer ceramics were significantly higher than the other tested veneer ceramics. Furthermore, the layered systems Rondo Dentine and Ceram Express were significantly stronger than the other tested core-veneer ceramics. The application of liner material dramatically affected the bond strength and failure mode, which was also material dependent. SEM analysis showed that two pressable veneers and one type of layering veneer ceramic failed entirely cohesively in the veneer while the remaining test groups had higher percentage of interfacial failure. It has been concluded that selection of stronger veneer ceramics which have good bond strength with zirconia can reduce the chances of chipping and delamination under function. It was also stated that the liner material should only be used with some layering veneers but not in combination with pressable veneers as it will result in weakening of the microtensile bond strength. Scanning Electron Microscopy and finite element analysis demonstrated that the core materials were significantly stronger than veneering materials and the core-veneer bond strength is one of the weakest links of layered all-ceramic restorations hence having a critical role in their success.[74]

Al-Dohan et al. investigated the strength of the core-veneer interface in bi-layered all ceramic systems (IPS-Empress2, Procera AllCeram, Procera All-Zircon, and DC-Zircon) and porcelain fused to metal as a control group, using the shear bond testing methodology. A sixth group was included where DC-Zircon coping was bonded with Eris veneering porcelain. A total of seventy-two samples, twelve of each system were made from one master die. A 2.378 mm diameter cylinder of the veneering porcelain was applied using a specially designed aluminum split mold and after firing, all specimens were placed in a mounting jig and subjected to a shear load using an Instron Testing Machine at a crosshead speed of 0.5 mm/min until failure. Microscopic examination at 20x showed that complete adhesive failure did not occur between compatible ceramic core and veneer materials. The mean values for shear strengths in MPa were: porcelain fused to metal 30.16 ± 5.88; IPS-Empress2 bonded to Eris 30.86 ± 6.47; Procera All-Zircon bonded to Cerabien CZR 28.03 ± 5.03; DC-Zircon bonded to Vita D 27.90 ± 4.79;

Procera AllCeram bonded to AllCeram veneering porcelain 22.40 ± 2.40; DC-Zircon bonded to Eris 2.028 ± 3.12. The study concluded that the bond of veneering porcelain to a ceramic core was similar to the bond of porcelain fused to metal.[75]

3.7 Fracture Location Analysis

The design of the restorations and the actual distribution of the tensile stresses must be taken into account; otherwise, the significant contribution of stronger and tougher core materials to the performance of all-ceramic restorations may be offset by the weaker veneering porcelain. [24]

An investigation conducted by Kelly et al. on clinically failed all-ceramic FPDs with a glass-infiltrated alumina core and porcelain veneer have shown that fractures originated at the area of the connector (where the thick core material was veneered with a thin layer of porcelain) and at the interface between the core ceramic and the veneering porcelain.[76] They explained the fracture at the connector showing with FEA the tensile stress concentration in this area, as also shown by Proos et al. [77] The fracture at the interface commonly has been associated with a stress enhancement arising from large differences in elastic modulus between the veneer and the core ceramic. Other investigators have shown that the fracture origin and the fracture mode are greatly influenced by the test methodology and by the core thickness/veneer thickness ratio. [78]
In a recent study, conducted by Guazzato, the mode of failure of disks with either In-Ceram Alumina or In-Ceram Zirconia and porcelain (on the bottom surface) has been

described as cracks with a star like configuration initiated on the bottom surface and propagating laterally and radially across the bottom tensile surface and towards the interface. [24]

Microscopic observations conducted in Guazzato's investigation did not allow drawing a definitive conclusion on which site the strength-controlling crack was initiated, since two clearly distinguishable origins were always seen. However, on the basis of the results of the mechanical strength tests, FEA and microscopy, it was surmised that the critical crack initiated on the bottom surface of the porcelain (where there is a peak of tensile stress) and propagated radially and laterally. In vicinity of the interface, the crack was deflected and propagated along the interface until another crack initiated from a flaw on the bottom surface of the core material.[79] Eighty discs 14mm in diameter were made from conventional dental porcelain and DC-Zircon core ceramic, and equally divided into four groups of twenty specimens in each as follows: (VD) monolithic specimens of porcelain; (DZ) monolithic specimens of core material; (VD/DZ) bilayered specimens with the porcelain on top (facing the loading piston during testing); (DZ/VD) bilayered specimens with core material on top. The load was applied (crosshead speed of 0.5mm/min) at the center of the surface through the flat tip of a piston (1.5mm of diameter) mounted on a universal testing machine. The maximum load at fracture was calculated with a biaxial flexural test, finite element analysis was used to estimate the maximum tensile stress at fracture and results were analyzed with one-way Anova and Tukey HSD tests. The statistical analysis showed that monolithic core specimens and the bilayered sample with the core material on the bottom were significantly stronger than monolithic porcelain disks and bilayered samples with the porcelain on the bottom. SEM was utilized to identify the initial crack and characterize the fracture mode. The present study supports this hypothesis by showing not only crushing of the porcelain without fracture of the core material, but also that the strength and strength variability of VD/DZ specimens were dictated by the stronger Y-TZP core material. Since Y-TZP is stronger than any other ceramic core material previously investigated, an improvement of the resistance of the crown and therefore its clinical performance is expected when this ceramic is used as core material. Ultimately, it was observed that when cracks met the core normal to the interface, they generally propagated through the whole specimen without any delamination. Some of the other star cracks did not approach the core normal to the interface and were deflected to run along the porcelain/core interface. This behavior has been related to the elastic modulus and fracture toughness mismatch between core and porcelain. In the paper delamination was consistently seen even for those cracks that apparently approached the core perpendicularly to the interface. [79]

Potiket in his study examined fracture mode of the crown and classified them according to a classification proposed by Burke; Class I, minimal fracture or crack in crown; Class II, less than half of crown lost; Class III, crown fracture through midline, half of crown displaced or lost; Class IV, more than half of crown lost; Class V, severe fracture of tooth and/or crown. Visual analysis of the fractured specimens showed that all the specimens (100%) in every group exhibited a Class V mode of fracture. On forty intact, extracted maxillary central incisors, prepared with 1.0-mm deep shoulder finish line with a rounded internal line angle, restorations were fabricated and subdivided into 4 groups (n=10): Group MCC (control), metal-ceramic crown (JRVT High Noble Alloy); Group AC4, crown with 0.4-mm aluminum oxide coping (Procera AllCeram); Group AC6, crown with 0.6-mm aluminum oxide coping (Procera AllCeram); and Group ZC6, crown with 0.6-mm zirconia ceramic coping (Procera AllZirkon). All restorations were treated with bonding agent and cemented with Panavia 21. Restorations were stored in 100% relative humidity of a normal saline solution for 7 days and fracture strength was assessed with a universal testing machine at a crosshead speed of 2 mm per minute with an angle of 30 degrees to the long axis of the tooth. Visual examination revealed that mode of failure for all specimens was fracture of the natural tooth, no crowns dislodged from the prepared tooth, and there were no fractures of the all-ceramic or metal-ceramic crowns.[80]

Pallis et al. evaluated the fracture resistance and origin of failure of simulated first molar crowns fabricated using 3 all-ceramic systems, IPS Empress 2, Procera AllCeram, and In-Ceram Zirconia. IPS Empress, Procera All Ceram and In-Ceram Zirconia and loaded with universal testing machine. A stainless steel definitive die was machined to be axisymmetric with a profile identical to an all-ceramic crown preparation on a maxillary first molar with a1-mm modified shoulder and 1.5- to 2.0-mm occlusal reduction (1.5-mm reduction at the center of the occlusal table and 2.0-mm reduction at the cusps). Sixty duplicate dies were fabricated in a high filler content resin material (Viade Products Inc) to replicate the definitive die. The Procera AllCeram cores were presintered, milled, and sintered by the manufacturer and twenty In-Ceram Zirconia cores were fabricated using a CAD-CAM system. All 40 cores were fabricated to a target thickness of 0.5 mm on all surfaces. Twenty IPS Empress 2 cores (Ivoclar Vivadent) were fabricated to a target thickness of 0.7 mm on the axial wall and 1.0 mm on the occlusal table using vacuum forming sheets. Vitadur Alpha porcelain (Vident, Brea Calif) was used to complete Procera AllCeram and In-Ceram Zirconia crowns, while Eris porcelain (Ivoclar Vivadent) was used to complete the IPS Empress 2 crowns with centrifugal sculpturing device to provide contour at a consistent thickness followed by firing cycles. The Procera AllCeram and In-Ceram Zirconia crowns were prepared for luting by airborne-particle abrasion of the internal surfaces with 50 µm aluminum oxide at 80 psi. The internal surfaces of the IPS Empress 2 crowns were acid-etched with 9.5% hydrofluoric acid for 2.5 minutes in preparation for luting. The surfaces of all 60 dies were airborne-particle abraded with 50 µm aluminum oxide at 40 psi for 5 seconds. The surfaces of all 60 crowns were cleaned in distilled water for 10 minutes and air dried. All crowns were silanated (Clearfil Porcelain Bond Activator), luted to the dies with a RelyX resin luting agent and immediately placed under a static load of 20 N for 5 minutes. The center of the occlusal surface on each of 15 specimens per ceramic system was axially loaded to fracture in a universal testing machine using a stainless steel ball bearing (6.35 mm in diameter) at a crosshead speed of 5 mm/min, and the maximum load (N) was recorded. Fractured surfaces were examined using optical and electron microscopy to determine the most prevalent origin of failure in each ceramic system. Five crowns per system were sectioned, and thickness of the luting agent, core material, and veneer porcelain layers were measured. The 95% confidence intervals of the Weibull modulus was used to compare failure load between the 3 systems. Two-way multivariate analysis of variance was used to analyze the thickness of the luting agent, ceramic core, and veneer porcelain layers (α=.05). The 95% confidence intervals for Weibull modulus were 1.8 to 2.3 (IPS Empress 2), 2.8 to 3.6 (Procera AllCeram), and 3.9 to 4.9 (In-Ceram Zirconia). The 95% confidence intervals for characteristic failure load were 771 to 1115 N (IPS Empress 2), 859 to 1086 N (Procera AllCeram), and 998 to 1183 (In-Ceram Zirconia). The microscopy examination revealed that the origin of failure was most commonly found at the interface between the ceramic core and veneer porcelain for IPS Empress 2 and between the ceramic core and luting agent layer for the other systems. According to data, failure loads of all-ceramic crowns were influenced not only by the fracture resistance of the component materials but also by prosthesis geometry and size and location of flaws.[81]

3.8 Marginal Adaptation

The accuracy of fit of all-ceramic restorations is important for the integrity of dental and periodontal tissues, dissolution of luting agents and the fracture resistance of the restorations. Poorly fitted dental prostheses are believed to be associated with the development of secondary caries and periodontitis.[82, 83] The aspect determining the accuracy of restorations, marginal fit, has been evaluated in numerous studies using various systems and materials. American Dental Association specification no.8 states that the luting agent film thickness for a crown restoration should be no more than 25 µm when using a Type I luting agent, or 40 µm with Type II cement. Marginal fit of cemented prostheses in the 25 to 40 µm range has been suggested as a clinical goal. [84]

Neiva at al. in her study evaluated three ceramic systems (IPS Empress, Procera AllCeram and InCeram) in the aspects of their fracture and marginal fit. The study reported that the largest gap size, especially at the marginal opening and axial wall was evident in Procera AllCeram (225 and 105 respectively) ceramics compared to IPS Empress (90, 60) and In-Ceram (135, 445) ceramic systems. The presence of marginal discrepancy was explained by the scanning error, the diameter of sapphire probe was slightly larger than the rounded internal angle of the preparation. The second possibility was that the presenter original coping is 20% larger than the die, and after sintering it shrinks about 20%. However, variations that compromise the fit of the crown to the original die may occur. [66]

The clinical success of the CAD/CAM restorations highly depends on the mechanical properties as well as design of the restoration and accuracy of the CAD/CAM process.

Hertlein et al. performed a study to assess marginal fit of zirconia restorations with three/four abutment teeth. Restorations of five different clinical situations of 4-unit bridges and four splinted crowns, respectively, were manufactured from Lava™ Zirconia by means of the Lava System (3M ESPE AG). After manufacturing, each framework was cemented on the scanned dies, embedded into acrylic and subsequently sectioned faciolingually and mesiodistally. A stereomicroscope and a special analyzing software (analysis, Soft Imaging -System GmbH) were used for the determination of marginal opening (MO) and absolute marginal opening (AMO) of the cross-sections. The data were compared to previous results of 3-unit bridges with two abutments. There was a significant difference of the MO values between the three indications with a higher MO of four splinted crowns (One-way ANOVA, $p<0.05$ Tukey Test). The results indicated that marginal opening was higher in sites with four crowns and no pontics while no difference in absolute marginal opening was observed among the five studied prostheses.[85]

In a study, on marginal adaptation of a single anterior restoration, Yeo et al. compared three ceramic systems: Celay In-Ceram, conventional In-Ceram and IPS Empress 2 with metal ceramic restoration as a control group. The in vitro marginal discrepancies of 3 different all-ceramic crown systems (Celay In-Ceram, conventional In-Ceram, and IPS Empress 2 layering technique), and a control group of metal ceramic restorations were evaluated and compared by measuring the gap dimension between the crowns and the prepared tooth at the marginal opening. The crowns were made for 1 extracted maxillary central incisor prepared with a 1-mm shoulder margin and 6-degree tapered walls by milling. Thirty crowns per system were fabricated and crown measurements were recorded with an optical microscope, with an accuracy of +/-0.1 micron, at 50 points spaced approximately 400 microns along the circumferential margin. The criterion of 120 micron was used as the maximum clinically acceptable marginal gap. Mean gap dimensions and standard deviations were calculated for marginal opening. The authors found that mean gap dimensions and standard deviations at the marginal opening for the incisor crowns were 87 +/- 34 µm for control, 83 +/- 33 µm for Celay In-Ceram, 112 +/- 55 µm for conventional In-Ceram, and 46 +/- 16 µm for the IPS Empress

2 layering technique. Compared with the control group, the IPS Empress 2 group had significantly smaller marginal discrepancies ($P<.05$), and the conventional In-Ceram group exhibited significantly greater marginal discrepancies ($P<.05$). There was no significant difference between the Celay In-Ceram and the control group, and marginal discrepancies were all within the clinically acceptable standard set at 120 µm. As a standard criterion for the marginal discrepancy the maximum 120 µm gap of acceptance was used where the IPS Empress 2 system showed the smallest (46 ± 16 µm) and most homogenous gap dimension, whereas the conventional In-Ceram system presented the largest (112 ± 55 µm) and most variable gap dimension compared with the metal ceramic (control) restoration. [86]

Different research performed by Hilgert et al. evaluated the marginal adaptation of ceramic crowns (In-Ceram, Vita) by varying two cervical endings of copings which received one ceramic (Vitadur Alfa, Vita). Two master steel dies were milled with all-ceramic crowns preparation, one with a round shoulder margin design, and the other with a deep chamfer. Fifteen copings were made of each die, and the marginal adaptation evaluated in measuring microscope. The ceramic addition was accomplished with the aid of a silicon matrix for standardization of the samples and was baked in three firing cycles: building of the crown, correction and glazing. The data were submitted to the analysis of variance (ANOVA). Mean marginal discrepancy observed was 44.0 µm for deep chamfer and 24.0 µm for round shoulder in the initial measures. Marginal gaps of 53.3 µm for deep chamfer and 27.4 µm for the round shoulder were observed after addition of ceramic. Statistical differences were observed between the two types of margin designs, as well as before and after the addition of ceramic. The data analysis reported that round shoulder margin design presents smaller values of marginal gaps and that the addition of ceramic influences the final values of marginal adaptation. [87]

Highly important factor affecting the marginal adaptation hence the longevity of the restoration is a coping thickness. The study undertaken by Al-Ghazzawi et al. evaluated the effect of various coping thicknesses and laboratory processes on marginal adaptation of alumina and zirconia crown copings utilizing CAD-CAM technology. A metal abutment was fabricated with a 1 mm marginal width, 4 mm abutment height, and 12° convergence angle. A Cerec inLab system (Sirona) was used to scan the abutment, design, and mill the test samples. Alumina and zirconia copings were fabricated with thicknesses of 0.6 and 0.8 mm using Cerec inLab system and cement space thickness of 20, 30, and 40µm. Epoxy resin replicas of the abutment and copings were fabricated and the vertical interfacial gaps were measured with a Micro-Vu optical microscope at three stages: post milling (PM), post trimming (PT), and post infiltration with the glass phase (PI). Statistical analysis (Anova, $p < 0.05$) showed no significant difference among the 3 groups except for the zirconia PM and PI groups. The marginal gap was not affected by different coping thickness (n=156-180): 0.6 mm coping measured 38.3 µm whereas 0.8mm coping had 38.8 µm, and cement space thickness (n=56-120): 20 µm = 36.5, 30 µm = 39.5, 40 µm = 39.9. It has been suggested that glass infiltration significantly improved marginal adaptation of CAD/CAM zirconia copings allowing these restorations to meet the clinical standards. [88]

Rosentritt et al. compared the fracture strength and marginal adaptation of all-ceramic three-unit fixed partial dentures (FPDs) in computer aided manufactured zirconia (Cercon Base/Cercon Ceram Kiss, DeguDent, G), an electro-layered chip-bridge-system (Inceram Alumina- Wolceram, G). Human molars were inserted in resin to create a three-unit (10mm) oral situation. Eight bridges of each series were made and a veneered with Al_2O_3 (Inceram Alumina, Vita, G; - control). All FPDs were fixed with an adhesive bonding system (Syntac classic/Variolink2; Ivoclar-Vivadent, FL) and subjected to thermocycling and mechanical loading (6000 thermal cycles [5°C/55°C] and $1.2x10^6$ mastication cycles [50N]). Fracture strength (at speed of 1mm/min) was determined. Failure detection was set to 10% of the maximum force. Marginal adaptation was evaluated at both transitions cement-tooth (CT) and cement-FPD (CF) using scanning electron microscopy before and after thermocycling. The criteria "perfect margin" was ranked as a smooth transition without interruptions of continuity. Statistical analysis showed that zirconia CAM FPD's had statistically significant higher fracture resistance than Al_2O_3 FPD's or electro-layered FPD's. [89]

Another study presenting the clinically acceptable adaptation of CAD/CAM ceramics was carried out by Komine at al. where the marginal and internal adaptations of partially sintered zirconium dioxide ceramic copings with three different finish line designs were evaluated. Three steel dies were prepared for maxillary central incisor crowns with the following finish line designs: a shoulder (S), a rounded shoulder (RS), and a chamfer (C) preparation. Twenty-four standardized partially sintered zirconia ceramic copings were fabricated using a CAD/CAM system (Cercon smart ceramics) for the three tested groups (n=8). The marginal adaptation was assessed by measuring the vertical discrepancy, which measured parallel to the abutment axis the distance between the outer restoration margin and the preparation line. The measurements were performed at 60 different points across the entire circumference of each coping. Internal adaptation was measured with the cement space replica technique using two different silicone pastes, while marginal and internal discrepancy was measured in a laser microscope at a magnification of x250. The data presented the following median values obtained from marginal/internal adaptation: S, 84.8/116.8; RS, 61.2/71.8; and C, 64.0/49.4 (μm). There were no significant differences among the three groups for marginal adaptation. On the other hand, significant differences in the internal adaptation were found among all groups. They concluded that internal adaptation of partially sintered zirconia ceramic copings is affected by finish line design. [90]

3.9 Bond Strength

The clinical application or resin-bonded fixed partial restorations require a strong and stable bond to the ceramic. The luting agent and conditioning methods, as well as the proper treatment of the prosthodontic application under the conditions of oral cavity, play significant roles in obtaining the successful clinical outcome for non-etchable ceramics, unless a retentive type of tooth preparation is used. There are reports on bond strengths of various types of ceramics and composites tested with a variety of adhesive systems.

A very important aspect having an influence on the ceramic restoration longevity is the bonding cement. A strong, durable resin bond provides high retention, improves marginal adaptation preventing microleakage and increases fracture resistance of ceramic restorations. Adhesive bonding techniques currently available offer a range of highly esthetic options yielding to a predictable and durable final result. The compositions and properties of zirconia and alumina-based ceramic restorations differ significantly from silica-based ceramics and therefore require substantially different treatment procedures. Various articles contemplating that subject were published and one written by Markus Blatz et al. presents a literature review on the resin bond to dental ceramics. According to Blatz preferable treatment for the glass-infiltrated aluminum oxide ceramic are either Al_2O_3 abrasion and phosphate-modified resin cement, or tribochemical surface treatment with Rocatec system in combination with Bis-GMA resin cement. [91]

Successful ceramic-resin bonding is achieved by the chemical bond and micromechanical lock between the interface of ceramic and luting agent. Conventional silica-based ceramic materials achieve the bond strength by applying acid-etching and silane coupling solution. However, high strength alumina and zirconia ceramics have more unique physical properties hence requiring different treatments. A few studies demonstrated the superior bonding properties of adhesive phosphate monomer cement providing long-term durability of zirconia-based ceramic restorations.[92-94] The few available studies on the resin bond to zirconium-oxide ceramic recommend airborne-particle abrasion and modified resin luting agents containing adhesive monomers for superior and long-term durable bond strengths.[95] It is uncertain whether this regimen can also be successfully applied to the intaglio surface of a commercial zirconia-based all-ceramic system.

Blatz et al. evaluated and compared bond strengths of different bonding/silane coupling agents and resin luting agents to zirconia ceramic before and after artificial aging. Composite cylinders (2.9 mm x 3.0 mm) were bonded to air-borne –abraded internal surface of Procera AllZirkon specimens (n=80) with either Panavia F (PAN) or RelyX ARC (REL) luting cements after placement of the bonding agent Clearfil SE Bond/ Porcelain Bond Activator (Group SE). Other groups of specimens were cemented with RelyX ARC and pretreated with bonding/silane coupling agent (Single Bond/Ceramic Primer, Group SB). Panavia with no bonding at all was used a control group (Group NO). Subgroups of 10 specimens were stored in distilled water for either 3 or 180 days before shear bond strength testing. One hundred eighty-day-old specimens were repeatedly thermal cycled for 12,000 cycles between 5 and 60 degrees C with a 15-second dwell time. Data were analyzed with 1- and 2-way analysis of variance and the Tukey multiple comparisons test (α=.05) and failure modes were examined under 25x magnification. After 3 days, SE-REL (25.15 +/- 3.48 MPa) and SE-PAN (20.14 +/- 2.59 MPa) groups had significantly superior mean shear bond strengths (P=.0007) compared with either NO-PAN (17.36 +/- 3.05 MPa) or SB-REL (16.90 +/- 7.22 MPa). SE-PAN, NO-PAN, and SB-REL groups were not significantly different. Artificial aging significantly reduced bond strengths. After 180-day storage, SE-PAN (16.85 +/- 3.72 MPa), and SE-REL (15.45 +/- 3.79 MPa) groups demonstrated significantly higher shear bond strengths than NO-PAN (9.45 +/- 5.06 MPa) or SB-REL (1.08 +/- 1.85 MPa) groups. The modes of failure varied among 3-day groups but were 100% adhesive at the ceramic surfaces after artificial aging. It has been confirmed that artificial aging significantly reduced bond strength and bonding/silane coupling agent containing an adhesive phosphate monomer achieved superior long-term shear bond strength to airborne-particle- abraded Procera AllZirkon restorations with either one of the 2 resin luting agents tested. [96]

CAD/CAM systems using zirconium-oxide ceramics for the fabrication of copings and frameworks (e.g., Procera AllZirkon, Nobel Biocare) became very popular and the quest for adequate bonding techniques, material selection, and the compatibility of different bonding systems is of high interest. The phosphate-containing resin cement Panavia is the preferred cement for a strong resin bond to zirconia. However, research on the influence of silanization on the resin bond of this cement and the performance of conventional resin cements is controversial.

A study on comparative shear bond strength to Procera AllZirkon was carried out to inspect shear bond strength of the resin cements Panavia F (Kuraray) and RelyX ARC (3M ESPE) in combination with their corresponding ceramic primers/bonding agents Clearfil SE Bond/Clearfil Porcleain Bond Activator(CSB/PBA, Kuraray) and Ceramic Primer/Single Bond (CP/SB, 3M ESPE) to sandblasted Procera AllZirkon intaglio surface. Forty Procera AllZirkon specimens were sandblasted and composite-resin cylinders were bonded to the ceramic surface with the following material combinations: Group A (Panavia/no bond), Group B (Panavia/CSB/PBA), Group C (RelyX

ARC/CP/SB), and Group D (RelyX ARC/CSB/PBA). Shear bond strength was performed in an Instron universal testing machine after storage of the specimens in distilled water for 3 days. One-way ANOVA and Tukey's HSD post-hoc tests were used to compare and separate the groups with respect to mean shear bond strength (α=0.05) and the mean shear bond strengths were shown to be different significantly among the four groups. Group D (25.1 MPa) was superior to groups C (16.9 MPa) and A (17.4 MPa). Group B (20.1 MPa) ranked just below Group D but produced bond strengths that were not statistically different from the other groups. The bonding agent/ceramic primer Clearfil SE Bond/Clearfil Porcelain Bond Activator seemed to provide a superior bond with the resin cements Panavia and especially RelyX ARC.[97]

Luthy in his research tested shear bond strength of several cements (Ketac-Cem, Nexus, RelyX Unicem, Superbond C&B, Panavia F, and Panavia 21) to densely sintered tetragonal zirconia. Groups of thirty test specimens were prepared by bonding stainless steel cylinders tribochemically silica-coated with the Rocatec-system to sandblasted ZrO2-TZP ceramic disks (Cercon smart ceramics). Prior to testing all bonded specimens were stored in distilled water at 37ºC for 48 h and half of them (n=15) were additionally aged by thermocycling (10,000 times). The analysis illustrated that none of the fractures occurred at the interface of the metallic rods and furthermore the assemblies failed either at the interface between the ceramic surface and the cements or within the cements. It was shown that thermocycling affected the bond strength of all luting cements studied except for both Panavia materials and RelyX Unicem. Nexus in combination with tribochemical silica-coating of ceramic surface produced higher bond strength.. Superior results were achieved with RelyX Unicem, Superbond C&B, Panavia F, and Panavia 21, and the strongest bond to zirconia was obtained with Panavia 21.[92]

In the study where Lava system was tested with different luting agents, Ernst et al. demonstrated the superior retentive strength of Superbond C&B, Panavia F, Dyract Cem Plus, RelyX Luting and RelyX Unicem. The aim of this in vitro study was to determine the retentive strength of four resin-cement systems: a compomer, a glass-ionomer cement, a resin-modified glass-ionomer cement, and a self-adhesive resin for luting zirconium oxide ceramic crowns. One-hundred-twenty extracted human teeth were randomly divided into 12 groups (n = 10) and prepared in a standardized manner (5-degree taper, 3-mm occlusogingival height) followed by fabrication of all-ceramic crowns and cementation with: CO, Compolute/EBS Multi; CO/RT, Compolute/EBS Multi/Rocatec; CB, Superbond C and B; CB/RT, Superbond C and B/Rocatec; CB/PL, Superbond C&B/Porcelain Liner M; PA, Panavia F; DC, Dyract Cem Plus/Xeno III; CH/PL, Chemiace II/Porcelain Liner M; RL, RelyX Luting, K/C, Ketac Cem/Ketac Conditioner; K, Ketac Cem; and RU, RelyX Unicem. After thermal cycling (5000 cycles, 5 degrees C-55 degrees C), the outer surfaces of the cemented zirconium oxide ceramic crowns were treated (Rocatec) to improve bonding and then placed into a low-shrinkage epoxy resin block (Paladur). The block/crown and tooth components for each specimen were connected to opposing ends of a universal testing machine so that crown retention could be measured. Crowns were removed from teeth along their path of insertion and the retentive surface area of 2 mm was determined individually for each tooth. Statistical analyses were performed using the Wilcoxon exact test (α=.05) and a Bonferroni correction (μ =.001). The median retentive strength values (MPa) were as follows: CO, 1.7 (0.6/4.3); CO/RT, 3.0 (1.3/5.4); CB, 4.8 (3.7/7.9); CB/RT, 8.1 (4.2/12.7); CB/PL, 5.3 (3.7/10.2); PA, 4.0 (3.3/5.1); DC, 3.3 (2.1/5.6); CH/PL, 4.0 (1.3/6.3); RL, 4.7 (2.8/6.6); K/C, 1.8 (0.6/2.3); K, 1.9 (0.2/4.5); and RU, 4.8 (2.5/6.7). Superbond C&B + Rocatec specimens showed the highest median retentive strength, but were not significantly different from Superbond C&B without Rocatec pretreatment. Compolute specimens also did not benefit significantly from the Rocatec pretreatment, whereas Superbond C&B, Panavia, Dyract Cem Plus, RelyX Luting, and RelyX Unicem showed the highest median retentive strength values and were not significantly different. The study gave evidence that Rocatec pretreatment of the ceramic surface did not improve the retentive strengths of Compolute and Superbond C&B.[94]

Escribano et al. study tested microtensile bond strength of self-adhesive luting cements to ceramic IPS Empress II discs and dentin of perfused teeth. In that study Panavia F obtained the highest bonding values, followed by Multilink System. RelyX Unicem., the only material that does not require pretreatment of dentin, achieved the lowest microtensile bond strength. Occlusal enamel and the roots of 9 human third molars were removed and crown segments connected to a perfusion system (30 cm H2O). Nine ceramic disks made of IPS Empress II were prepared, conditioned with 5% HF (20 s), rinsed with water, and air dried. A primer silane agent was applied (Monobond-S) and teeth were bonded to disks using one of three materials: Multilink System, RelyX Unicem, or Panavia F light. Specimens were vertically sectioned to obtain square bars and each bar was fixed to a rigid custom-made tensile device and subjected to tensile force until debonding. Microtensile bond strength was given in MPa. ANOVA was performed to examine the statistical significance of differences between means of the groups, and the Tamhane's post hoc test was used to locate eventual differences. ANOVA showed statistically significant differences between the groups' means ($p < 0.00001$), while Tamhane's post hoc test ($p < 0.05$) showed that different group means were RelyX < Multilink < Panavia F. Materials showed different tensile bond strengths where Panavia F obtained highest bonding values, followed by Multilink System and RelyX Unicem. The only material that does not require pretreatment of dentin, achieved the lowest TBS.[93]

The coupling agent or more precisely triakoxysilanes that are hybrid inorganic-organic bifunctional molecules have an influence on bonding. Typical silane coupling agent contains an organofuncional part and three hydrolyzable aloxy groups. Prior to becoming adhesion promoters and activated, the hydrolyzation reaction is promoted in slightly acidic ethanol-water solvent to transform trialkoxy groups to silanols. The methacrylate group can then be polymerized with the monomers of resin composite systems. The presence of coupling agent lowers the surface tension of a substrate, wet it and make its surface energy higher thus accessible for effective bonding. The market offers two ways of application. Tribochemical silica-coating, Rocatec system, used in laboratories contains the silica-coated alumina particles that are blasted onto intaglio surface of the porcelain. CoJet, having similar properties, is applied intra-orally. The studies confirmed that silane coupling agents improved the bond of resin composite to ceramics by 25%. [98] The study comparing the retention of ceramics treated with both airborne particle abrasion and silane solution with retention of ceramics treated solely with abrasion showed that air-borne particle conditioning did not yield enough retention without silane solution.[99]

The fact that zirconia is different than alumina has an influence on bonding the surface to the tooth by cement. The adhesive performance of air-borne alumina particle treated and silica-coated zirconia was tested in the study with commercially used resin cement RelyX ARC and Bis-GMA resin. Samples made of Procera AllZircon were sandblasted with 110 μm Al_2O_3, silica-coated with Rocatec Plus and treated with three different silane agents followed by cementation with either Bis-GMA resin or RelyX ARC. Two of the tested silane agents performed well for both cements.[100]

Kim at al evaluated, in-vitro, the tensile bond strength of composite resin to all-ceramic coping materials made of lithium-disilicate (Empress2), alumina ceramic (In-Ceram Alumina), zirconia ceramic (Zi-Ceram) and feldspathic ceramic (Duceram Plus [F]) as the control. Ceramic specimens of each coping material were fabricated at dimensions of 10 x 10 x 2 mm and grouped into three different surface treatments: airborne-particle abrasion with 50 μm alumina particles (Ab); airborne-particle abrasion with 50 μm alumina particles and acid etching with 4% hydrofluoric acid (Ae); or airborne-particle abrasion with 30 μm alumina particles modified with silica acid (Si). After surface treatment of ceramic specimens, composite resin cylinders (5 mm diameter x 10 mm height) were light polymerized onto the ceramic specimens. Each specimen was subjected to a tensile load at a crosshead speed of 2 mm/min until fracture and fracture sites were examined with scanning electron microscopy to determine the location of failure during debonding and to evaluate the surface treatment effects. Two-way analysis of variance and the Duncan multiple comparison test (α=.05) were used to analyze the bond strength values. Significant differences were found in the bond strengths for both ceramics (P<.001) and surface treatments (P<.001) and the interaction (P<.001). The Duncan analysis yielded the following statistical subsets of the bond strength values: (FAe, ISi, EAe, ZSi) > FAb > (FSi, EAb, ESi) (IAb, IAe) > (ZAe, ZAb). The results illustrated no differences within the parentheses but statistically significant differences among the groups. Alumina and zirconia ceramic specimens treated with a silica coating technique, and lithium disilicate ceramic specimens treated with airborne-particle abrasion and acid etching yielded the highest tensile bond strength values to a composite resin for the materials tested.[95]

Derand at al assessed in-vitro bond strength of dental resin agent to zirconia ceramic after surface pre-treatment with different techniques. Blocks made of yttrium-oxide-partially-stabilized zirconia (ZF) (Procera All-Zircon) were compared to glossy dense zirconia blocks (ZG). Four groups of specimens with different surface treatment were prepared. Group I: ZF (n = 5) and ZG (n = 5) without any pre-treatment, Group II: ZF-s (n = 5) and ZG-s (n = 5) treated with silane solution, Group III: ZF-P (n = 10) and ZG-P (n = 10) treated with RF plasma spraying (hexamethyldisiloxane) using a reactor (Plasma Electronic, Germany), Group IV: ZF-p (n = 10) and ZG-p (n = 10) treated with micro pearls of low fusing porcelain (720 degrees C) on the surfaces. Composite cylinders (Charisma, Heraeus Kulzer, Dormagen, Germany) were luted with Variolink II

(Ivoclar-Vivadent, Schaan, Liechtenstein) to the test specimens. After 1 h storage in the air the specimens were subjected to shear load in a universal testing machine (LRX, Lloyd Instruments, Farnham, England) until failure. No statistical difference was found between the untreated ZF and ZG specimens (Group I) nor between the specimens treated with silane (Group II), yet plasma spraying treatment improved bond strength by a factor of three ($p < 0.001$). Treatment with low fusing porcelain micro pearls increased the bond strength by a factor of 10 compared to untreated surfaces ($p < 0.001$), while no significant difference was seen between the surfaces treated ZF-p and ZG-p specimens. The thickness of the glass pearls layer did not exceed 5 μm and SEM showed dense grain borders of ZF and a flat glossy texture of ZG. It was concluded that treatment of zirconia ceramic surfaces with plasma spraying or a low fusing porcelain pearl layer significantly increased the bond strength of resin cement to the ceramic surface. [101]

3.10 Esthetics

Ceramics have been advocated as a material of choice for matching the natural dentition. The structure of the tooth influences its color: dentin is more opaque than enamel and reflects light, whereas enamel is a crystalline layer over the dentin, composed of many prisms or rods implemented together in an organic structure. The light ray is scattered by reflection and refraction to produce a translucent effect, and as it reaches the tooth structure, part of it is reflected, while the remainder penetrates the enamel and is scattered. Any light contacting the dentin is either absorbed or reflected and further scattered within enamel. In the absence of dentin, the area of that particular tooth structure appears to be translucent, where the light is transmitted and absorbed in the oral cavity. Metamerism is a common phenomenon that occurs as an effect of different optical tooth appearance based upon the light. Dental ceramic materials are pigmented by the inclusion of oxides. The color of pigment is determined by absorption and reflection of light.[1] All ceramic restorations offer improved esthetic results, compared to metal-ceramic restorations, due to the natural translucent and reflection effect.[21, 23] Natural translucency is needed to achieve an appearance similar to that of human teeth. Fabrication of all-ceramic dental crowns is challenging because exceptional skills of a technician are required to provide minimal stress concentration areas using proper occlusal design and accurate marginal fit. In addition, ceramic crowns must be translucent and resistant to fracture even in clinical situations where inadequate thickness precludes optimal design. The ability to blend a porcelain crown with its natural counterpart involves consideration of size, shape, surface texture, translucency, and color. The light reflection can be affected by many factors, including ceramic thickness[21], crystalline structure, and number of firings.[102] Increased crystalline content results in greater opacity. The ceramic core of ceramic restorations may be fabricated from feldspathic porcelain, aluminous porcelain, lithia-disilicate-based ceramic, glass infiltrated magnesia aluminate spinel, glass-infiltrated alumina, glass-infiltrated zirconia and mica-based glass–ceramics. However, poor resistance to fracture has been a limiting factor in their use, especially for multi-unit fixed partial dentures.

Certain all-ceramic restorative materials have been recommended based on the translucency of the teeth to be matched. The coping design, its crystalline structure, and length or thickness influences the strength, adaptation but more importantly the esthetics.

Haffernan et al. compared the translucency of six all-ceramic materials where specimens (n = 5 per group) of Empress dentin, Empress 2 dentin, In-Ceram Alumina, In-Ceram Spinell, In-Ceram Zirconia, and Procera AllCeram were fabricated and veneered with their corresponding dentin porcelain to a final thickness of 1.47 +/- 0.01 mm. These specimens were compared with veneered Vitadur Alpha opaque dentin (as a standard), a clear glass disc and a high-noble metal-ceramic alloy veneered with Vitadur Omega dentin. Specimen reflectance was measured with an integrating sphere attached to a spectrophotometer across the visible spectrum (380 to 700 nm); 0-degree illumination and diffuse viewing geometry were used. Measurements were repeated after a glazing cycle and contrast ratios were calculated. One-way analysis of variance and Tukey's multiple-comparison test showed a significant differences in contrast ratios among ceramic systems tested when they were veneered (P<.0001) and after the glazing cycle (P<.0001). Significant changes in contrast ratios (P<.0001) also were identified when the veneered specimens were glazed. The glazing cycle resulted in decreased opacity for all test materials except the completely opaque In-Ceram Zirconia and metal-ceramic specimens.[21]

Haffernan et al. in another study disputed whether translucency differs when all-ceramic materials are fabricated similarly to the clinical restoration with a veneered core material. He compared the translucency of all-ceramic materials veneered and glazed at clinically appropriate thicknesses. Core specimens (n = 5 per group) of Empress dentin, Empress 2 dentin, In-Ceram Alumina, In-Ceram Spinell, In-Ceram Zirconia, and Procera AllCeram were fabricated and veneered with their corresponding dentin porcelain to a final thickness of 1.47 +/- 0.01 mm and compared with veneered Vitadur Alpha opaque dentin (as a standard), a clear glass disc (positive control), and a high-noble metal-ceramic alloy (Porc. 52 SF) veneered with Vitadur Omega dentin (negative control). Specimen reflectance was measured with an integrating sphere attached to a spectrophotometer across the visible spectrum (380 to 700 nm); 0-degree illumination and diffuse viewing geometry were used and all measurements were repeated after a glazing cycle. Contrast ratios were calculated from the luminous reflectance (Y) of the specimens with a black (Yb) and a white backing (Yw) to give Yb/Yw with CIE illuminant D65 and a 2-degree observer function (0.0 = transparent, 1.0 = opaque). One-way analysis of variance and Tukey's multiple-comparison test ($P<.05$ affirmed significant differences in contrast ratios among the ceramic systems tested when they were veneered ($P<.0001$) and after the glazing cycle ($P<.0001$). Significant changes in contrast ratios ($P<.0001$) also were identified when the veneered specimens were glazed. The presented work illustrated that glazing cycle resulted in decreased opacity for all test materials except the completely opaque In-Ceram Zirconia and metal-ceramic specimens.[23]

3.11 Zirconia Porcelain

3.11.1 Historical Perspective and Applications

A new method for fabrication of crowns and fixed partial dentures was developed involving the combination of computerized machining and copy milling. A pioneer Procera All-Ceram was developed by Andersson and Oden through cooperative effort between Nobel Biocare AB, Gothenburg, and Sandvik Hard Material AB, Stockholm, Sweden. The Procera system provides industrial sintering of an alumina coping using highly purified alumina. That system requires the following steps for fabrication of the restoration: magnification of the die, sintering of high-purity aluminous oxide powder at 1550°C on a die to form the coping, and addition of the overlay, low-fusing porcelain to achieve natural anatomy and shade of the tooth. For the past years extensive research has focused on improving the physical strength of dental ceramics. Among the various procedures, the mechanism of addition of zirconia and the process of addition of yttrium oxide particles have demonstrated the potential interest in the stabilized zirconia as a candidate biomaterial. The biocompatibility and high strength has led the zirconia material to be utilized in many potentially suitable situations by chemical industry where corrosive agents are employed, medical field in hip replacement or as an alloying agent in surgical appliances. The impure oxide of zirconia is utilized for laboratory crucibles that will withstand heat shock, for linings of metallurgical furnaces, and by the glass and ceramic industries as a refractory material.[75]

The name zircon probably originates from the Persian word zargun, which describes the color of the gemstone now known as zircon, jargon, hyacinth, or ligure. German chemist, Martin Heinrich Klaproth in 1789 identified zirconia as a reaction product in the process after heating some gems. The impure metal was first isolated by Berzelius in 1824 by heating a mixture of potassium and potassium zirconium fluoride. Zirconium is found in S-type stars, and has been identified in the sun and meteorites.[103] The implementation in biomedical field was started by Helmer and Driskell in 1969, while the first use of zirconia in orthopedics was initiated by Christel to manufacture ball heads for total hip replacements..[104, 105] Further expansion in dentistry led the zirconia be applicable in orthodontics, post and core systems, all-ceramic restorations, and ceramic implant/implant abutments offering improved esthetic alternatives.[106-110]

3.11.2 Chemical and Mechanical Properties

The preparation of zirconia starts from zircon ($ZrSiO_4$) which due to the process of melting and adding HCl (hydrochloric acid) produces zirconyl chloride. The next step is to produce zirconia powder ZrO_2 from zirconyl chloride by either precipitation or thermal decomposition. The final product is achieved through the addition of stabilizers (magnesium, calcium and yttria oxides) followed by sintering to allow conversion phase to occur. Three forms of zirconium can be distinguished: pure zirconia, partially stabilized zirconia (PSZ) and fully stabilized zirconia. Zirconia is a polymorph that occurs in three crystal phases: monoclinic (M), tetragonal (T) and cubic (C) (Figure 1). Pure zirconia in a room temperature presents in monoclinic phase. This phase is stable up to 1170ºC, while above that temperature zirconia transforms to tetragonal and then into cubic phase at 2370 ºC. The transformation of monoclinic into tetragonal phase causes 5% volumetric shrinkage. Reversible transformation is associated with 3% volume expansion. These phase transformations, however, induce stresses and propagate crack formation.[37, 111] When stress develops in the tetragonal structure causing a crack to occur, the tetragonal grains transform to monoclinic grains. That process is associated with volumetric changes thus results in compressive stresses at the edge of the crack front and additional energy is required for the crack to propagate further (Figure 2). The phenomenon of phase transition of zirconia is known as stress-inducing. Several approaches that are based on impeding the propagation of flaws have been used to strengthen dental porcelains, including bonding to metals, adding microcrystalline phases, and surface treatments (i.e., polishing, ion exchange and hydration).[112] The inhibition of negative transformations can be achieved by addition of stabilizing oxides (CaO, MgO, Y_2O_3), which allows the existence of tetragonal structure at room temperature. During the process of adding small particles of stabilizer (yttrium oxide) the tetragonal phase is produced along with a mixture of monoclonic and cubic phase. The process allows the formation of partially stabilized zirconia (PSZ), maintained in tetragonal phase (TZP), and is most commonly used in dental systems. The addition of a greater percentage of oxide particles promotes, during heating and cooling, a transformation to fully stabilized zirconia (fully yttria stabilized zirconia) which is characterized by solid cubic solution. Through that transformation, zirconia has been shown to have relatively high mechanical strength and the mechanism is called transformation toughness. [113]

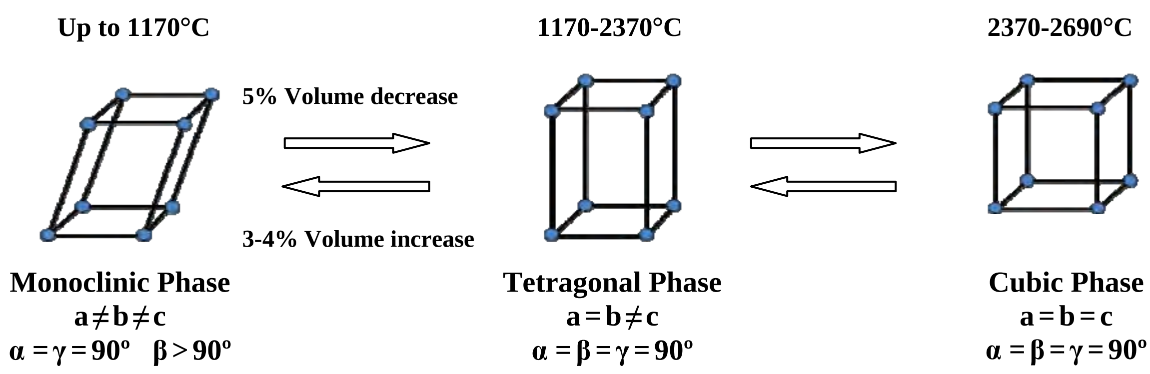


Figure 1. Zirconia phase transformation[113]

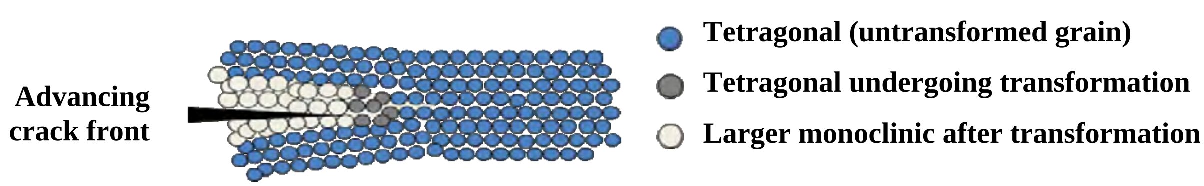


Figure 2. Crack propagation[113]

Yttrium stabilized tetragonal zirconia (YSTZ) is known to be the strongest porcelain available to the dental field. Physical properties of the zirconia depend on particle size, surface area, crystallite size and the presence of impurities.[114] The polycrystalline zirconia is characterized by high flexural strength of over 1000 MPa. Takagi and co-workers investigated the properties of densely sintered zirconia with medium-sized grains of approximately 0.8 μm. The zirconia was partially stabilized with

3.5 mol-percent Y_2O_3 and exhibited a fracture toughness K_{IC} of 8.4 MPa * $m^{1/2}$, an average elastic modulus of 200 GPa and flexural strength of 1000 MPa, which is twice as high as pure Al_2O_3. Measured values for the flexural strength with a content of 5 mol-percent Y_2O_3 were lower than that of 3 mol-percent, since with an yttria content of more than 3-4%, the transformation toughening loses its effectiveness.[115]

To fabricate denser sintered zirconium restorations different methods of manufacture have been developed. In-Ceram advertises zirconia-infiltrated glass as the material of choice for posterior bridges, and claims having the flexural strength of 700MPa as opposed to In-Ceram alumina with a flexural strength 500MPa. Different systems presently are available on the market that mills the tetragonal stabilized zirconia from either a partially (eg. Cercon Dentsply Ceramco; LAVA, 3MESPE,; Procera All-Zircon, Noble Biocare) or a fully sintered blank (DC-Zirkon, President). The phenomenon called transformation toughening is induced by stress exerted by grinding, impact, or fracture. While various systems have the capability to use a scanner to provide a copy of a die image and use computer assisted design to provide an optimal framework and marginal fit, the use of the conventional waxing technique can also be successful in designing the framework.

Studies have demonstrated that zirconia-based industrially optimized ceramic materials are more structurally reliable for dental application, comparatively superior in terms of mechanical properties [14, 116, 117] and can be used not only for prosthodontic purposes but also as endodontic posts or orthodontic brackets.[118] Guazzato et al.

investigated the strength, fracture toughness and microstructure of ceramic materials, demonstrated that zirconia-based dental porcelain is a stronger and tougher material compared to a conventional glass-ceramic.[27]

Tinschert et al. were tested the fracture resistance of three-unit fixed partial dentures (FPD) made of core ceramics. He used base metal three-unit master FPD model with a maxillary premolar and molar abutment. Tooth preparation showed 0.8-mm circumferential and 1.5-mm occlusal reduction and a chamfer margin design. FPDs were constructed with a uniform 0.8-mm-thick core ceramic and a porcelain veneer layer. In-Ceram Alumina, In-Ceram Zirconia, and DC-Zirkon core ceramics were machined by a computer-aided design/manufacturing system, whereas IPS Empress 2 core ceramic was indirectly built up using the fabrication technology of waxing and heat pressing. FPDs of IPS Empress were heat pressed as complete restorations without core material. All FPDs were cemented with ZnPO4 on the master model and loaded on a universal testing machine until failure. They found that the highest failure loads, exceeding 2,000 N, were associated with FPDs of DC-Zirkon. FPDs of IPS Empress and In-Ceram Alumina showed the lowest failure loads, below 1000 N, whereas intermediate values were observed for FPDs of IPS Empress 2 and In-Ceram Zirconia. [116]

Investigation of the load to fracture of different all-ceramic crown systems was carried out by Hogg et al. were five different coping systems were used such as: Group A - IPS Eris®, group B - Cerec® In-lab Alumina coping, group C - Cerec® In-lab Zirconia coping, Group D - Procera® AllCeram Alumina coping and group E - Procera® Zirconia coping. Ten crowns from each of the systems were fabricated, luted using G.C. Link Max resin adhesive cement to standardized, grade 2, pre-milled titanium dies, and placed under static load of 5 kg for 10 minutes followed by 100% humidity storage for one week. The samples were then loaded to fracture at the rate of 0.5mm/min. using a universal testing machine. The load to fracture (kgF) for each group was as follows; Group A=321.49 ± 113.69, Group B=288.63± 102.82, Group C=266.58± 69.17, Group D=295.49± 80.54, Group E=420.37± 82.45. The data was analyzed using an unpaired t-test which indicated that there was a statistical difference between group E and all of the others (P-value >0.039). It was concluded that the Procera Zirconia crown system had a significantly higher load to fracture value (420.37 ± 82.45 kgF) than several other all-ceramic crown systems. [119]

Pallis et al. evaluated the fracture resistance and origin of failure of simulated first molar crowns fabricated using 3 all-ceramic systems, IPS Empress 2, Procera AllCeram, and In-Ceram Zirconia. A stainless steel definitive die was machined to be axisymmetric with a profile identical to an all-ceramic crown preparation on a maxillary first molar with a1-mm modified shoulder and 1.5 to 2.0 mm occlusal reduction (1.5 mm reduction at the center of the occlusal table and 2.0 mm reduction at the cusps). Sixty duplicate dies were fabricated in a high filler content resin material (Viade Products Inc) to replicate the definitive die. The Procera AllCeram cores were presintered, milled, and sintered by the manufacturer and twenty In-Ceram Zirconia cores were fabricated using a CAD-CAM system. All forty cores were fabricated to a target thickness of 0.5 mm on all surfaces. Twenty IPS Empress 2 cores (Ivoclar Vivadent) were fabricated to a target thickness of

0.7 mm on the axial wall and 1.0 mm on the occlusal table using vacuum forming sheets. Vitadur Alpha porcelain (Vident, Brea Calif) was used to complete Procera AllCeram and In-Ceram Zirconia crowns, while Eris porcelain (Ivoclar Vivadent) was used to complete the IPS Empress 2 crowns with centrifugal sculpturing device to provide contour at a consistent thickness followed by firing cycles. The Procera AllCeram and In-Ceram Zirconia crowns were prepared for luting by airborne-particle abrasion of the internal surfaces with 50 μm aluminum oxide at 80 psi. The internal surfaces of the IPS Empress 2 crowns were acid-etched with 9.5% hydrofluoric acid for 2.5 minutes in preparation for luting. The surfaces of all 60 dies were airborne-particle abraded with 50 μm aluminum oxide at 40 psi for 5 seconds. The surfaces of all 60 crowns were cleaned in distilled water for 10 minutes and air dried. All crowns were silanated (Clearfil Porcelain Bond Activator), luted to the dies with RelyX resin luting agent and immediately placed under a static load of 20 N for 5 minutes. The center of the occlusal surface on each of 15 specimens per ceramic system was axially loaded to fracture in a universal testing machine using a stainless steel ball bearing (6.35 mm in diameter) at a crosshead speed of 5 mm/min, and the maximum load (N) was recorded. Fractured surfaces were examined using optical and electron microscopy to determine the most prevalent origin of failure in each ceramic system. Five crowns per system were sectioned, and thickness of the luting agent, core material, and veneer porcelain layers were measured. The 95% confidence intervals of the Weibull modulus was used to compare failure load between the 3 systems. Two-way multivariate analysis of variance was used to analyze the thickness of the luting agent, ceramic core, and veneer porcelain layers (α=.05). The 95% confidence intervals for Weibull modulus were 1.8 to 2.3 (IPS

Empress 2), 2.8 to 3.6 (Procera AllCeram), and 3.9 to 4.9 (In-Ceram Zirconia). The 95% confidence intervals for characteristic failure load were 771 to 1115 N (IPS Empress 2), 859 to 1086 N (Procera AllCeram), and 998 to 1183 (In-Ceram Zirconia). The microscopy examination revealed that the origin of failure was most commonly found at the interface between the ceramic core and veneer porcelain for IPS Empress 2 and between the ceramic core and luting agent layer for the other systems. According to data, failure loads of all-ceramic crowns were influenced not only by the fracture resistance of the component materials but also by prosthesis geometry and size and location of flaws.[81]

In 2004, Guazzato at al. investigated the biaxial flexural strength, reliability and the mode of fracture of eighty discs (diameter of 14 mm) made of monolithic and bilayered DC-Zirkon, an isostatically hot-pressed and fully sintered 5 wt% Y_2O_3 TZP, (DCS Dental AG, Allschwil, Switzerland) and Vita D, a feldspathic porcelain specifically developed to veneer dental zirconia. The specimens were prepared and divided in four groups of 20 specimens as follows: (VD) monolithic samples of Vita D; (DZ) monolithic samples of DC-Zirkon; (VD/DZ) bilayered samples of Vita D (top surface, facing the loading piston during testing) and DC-Zirkon (bottom surface, facing the supporting jig during testing); (DZ/VD) bilayered samples of DC-Zirkon (top surface, facing the loading piston during testing) and Vita D (bottom surface, facing the supporting jig during testing). After firing and sintering, all specimen surfaces were ground using diamond discs to the thickness of approximately 1.6 mm (±0.02) and then fired at the temperature of 920°C×1 min. The DZ, VD/DZ and DZ/VD specimens were prepared by initially cutting a number of cylinders from a block of DC-Zirkon with a 600 grit size diamond wheel. After cutting, 20 disks (for DZ) were ground and polished using a diamond wheel up to the nominal grit size of 1 μm and to the thickness of approximately 1.6 mm (±0.01). For the bilayered specimens (VD/DZ and DZ/VD), 40 discs were cut, as mentioned above, and ground to thickness of approximately 0.90 mm. The disks were then mounted on a silicon mold and slurry of Vita D porcelain was packed on top of the mold and fired as described above for the VD specimens. After sintering both surfaces were polished with diamond discs to the final thickness of 1.6 mm (±0.02), where the thickness of each layer was approximately 0.8 mm. Finally all specimens were glazed and biaxial flexural test (piston on three-ball) was used to calculate the maximum load at failure in the center of the other surface through the flat tip of a piston (diameter 1.5 mm) mounted on a universal testing machine at crosshead speed of 0.5 mm/min. One-way ANOVA was used to appraise whether there was any statistical difference among groups and a series of Tukey HSD post hoc tests were used to identify which pairs of groups were different. The variability of strength was estimated by calculating the Weibull modulus (*m*) from the Weibull distribution. Upon fracture, all specimens were gold coated and observed with an SEM to identify the initial crack and characterize the fracture mode. Finite Element Analysis was executed in order to examine the stresses that resulted from biaxial loading. The VD and DZ/VD specimens showed to be significantly weaker (p value < 0.000) than those groups where the core material was facing the jig (DZ and VD/DZ). Furthermore, there was no statistical significant difference between VD and DZ/VD (p 0.641), while VD/DZ specimens were significantly stronger that DZ (p<0000). The majority of the VD/DZ samples (80%) show multiple peaks and crushing of the porcelain. Only those that broke at lower loads did not show this behavior. Flexural strength has been recalculated by

considering the load at fracture of those specimens where bulk fracture occurred without crushing of the porcelain and, for the other specimens, the load which caused crushing of the porcelain rather than bulk fracture of the sample. This group is indicated as VD/DZ and shows no significant statistical difference with DZ (*p* 0.873). Monolithic core specimens and bilayered sample with the core material on the bottom were statistically significantly stronger than monolithic porcelain disks and bilayered samples with the porcelain on the bottom.[79]

In his other study, Guazatto et al. demonstrated significantly higher strength of dense zirconia based ceramics, where DC Zirkon, an experimental yttria partially stabilized zirconia, In-Ceram Zirconia slip and In-Ceram Zirconia dry-pressed were compared. Means of strength (MPa) and fracture toughness (MPa m(1/2)) values and their standard deviation were: In-Ceram Zirconia dry-pressed 476 (50)1, 4.9 (0.36)1; In-Ceram Zirconia slip 630 (58)2, 4.8 (0.50)1; the experimental yttria partially stabilized zirconia 680 (130)2, 5.5 (0.34)2; DC-Zirkon 840 (140)3, 7.4 (0.62)3. Consequently the study reported that zirconia-based dental ceramics were stronger and tougher materials than the conventional glass-ceramics. [27]

The investigation was conducted by Snyder et al. to assess the ultimate compressive strength of Procera sintered Zirconia Y-TZP copings using Procera Aluminum Oxide copings as a control. The study consisted of four groups, two contained six samples and two contained five samples: Group A was Aluminum Oxide copings luted with 3M Vitremer cement, Group B was Aluminum Oxide copings luted with G.C. Fuji Plus cement, Group C was Zirconium copings luted with 3M Vitremer cement, Group D was Zirconium copings luted with G.C Fuji Plus cement. All copings were luted to standardized, grade 2, pre-milled titanium dies and stored in 100% humidity for 10 weeks and inspected prior to testing for any detectable fractures. The samples were then loaded to fracture at the rate of 0.5mm/min. using a universal testing machine providing the following results. Group A=78.99 +23.32Kg (S.D.), Group B=105.70 +21.01Kg (S.D.), Group C=89.97 +34.92Kg (S.D.), Group D=111.26 +13.38Kg (S.D.). According to the authors, the ultimate compressive strength of the experimental Procera sintered Zirconia Y-TZP copings were not significantly different from that of Procera Aluminum oxide copings (p= 0.6066). [120]

The normal physiological chewing forces on posterior teeth range between 2-150N, while maximum biting forces can increase up to the range between 300 to 880 N. It has been shown that female biting forces are lighter than in males.[121] When bruxing occurs the biting forces are significantly higher and can rise up to 500N. [122] Due to crack growth, when dental ceramics is subjected to cyclic load it displays only 50% of initial fracture resistance. That process needs to be considered when decision is made on the material selection.[123]

In 2005, Curtis at al conducted an in-vitro investigation of the effect of simulated masticatory loading regimes on 5 mol% Y-TZP ceramic (LAVA). Ten sets of 30 Y-TZP ceramic discs measured 13 mm diameter where six groups were loaded for 2000 cycles at 500N (383-420MPa), 700N (536-588MPa) and 800N (613-672MPa) with three groups maintained dry and the remaining three groups loaded while immersed in water at 37±1ºC. A further two groups underwent extended simulated masticatory loading regimes at 80N (61-67MPa) for 10(4) and 10(5) cycles under dry conditions. The latter group presented with the significant increase in Weibull moduli (8.6±1.6, 8.5±1.6 and 10.3±1.9, respectively), compared with the unloaded specimens (7.1±1.3). The increased Weibull moduli were attributed to the formation of a localized layer of compressive stressed or crushing and densification of the material beneath the indentor counteracting with the material. In contrast, extending the simulated masticatory regime to 10^5 cycles at 80 N reduced the Weibull moduli of the specimens as a result of the combined influence of the accumulation of microfracture damage from crushing and densification of the material.[124]

The objective of the study done by Fischer was to evaluate the strength of zirconia single anterior crowns related to coping design modification. A model of upper canine was circumferentially prepared with a shoulder margin, the CoCr alloy was made followed by fabrication of cast die that was scanned. Eighty-eight Zirconia frameworks were made using Lava system: half of the fabricated copings had a constant wall thickness, the other half presented with anatomical shape of the framework similar to metal-ceramic substructure. Specimens were divided into subgroups with twenty-two frameworks veneered with Lava Ceram or Cerabien ZR, subjected to thermal shock test and examined for fractures. Remaining ten samples were cemented to metal master dies with Ketac Cem and subjected to the static 45º-angled load. According to the author Lava Ceram and Cerabien ZR demonstrated similar fracture toughness, however the strength increased by 30% in specimens with anatomical design of the zirconia coping.[125]

An in vitro study, conducted by Potiket et al. tested the fracture strength of teeth restored with a variety of ceramic systems. Forty intact, human maxillary central incisors were divided into 4 groups (n=10): Group MCC (control), metal-ceramic crown (JRVT High Noble Alloy); Group AC4, crown with 0.4-mm aluminum oxide coping (Procera AllCeram); Group AC6, crown with 0.6-mm aluminum oxide coping (Procera AllCeram); and Group ZC6, crown with 0.6-mm zirconia ceramic coping (Procera AllZirkon). Teeth were prepared for complete-coverage all-ceramic crowns so that a final dimension of 5.5 ± 0.5 mm was achieved incisocervically, mesiodistally, and faciolingually with a 1.0-mm deep shoulder finish line. All restorations were treated with bonding agent (Clearfil SE Bond) and luted with phosphate-monomer–modified adhesive cement (Panavia 21). After storage in 100% relative humidity of a normal saline solution for 7 days all specimens were subjected to fracture strength with a universal testing machine at a crosshead speed of 2 mm per minute with an angle of 30 degrees to the long axis of the tooth. The mode of fracture was examined visually. Statistical analysis revealed that there was no significant difference in fracture toughness of teeth prepared for all-ceramic crowns with 0.4mm and 0.6mm aluminum oxide copings or zirconia ceramic copings and teeth prepared for metal-ceramic restorations. The means of fracture strength were: Group MCC, 405 ± 130 N; Group AC4, 447 ± 123 N; Group AC6, 476 ± 174 N; and Group ZC6, 381 ± 166 N. There was no significant difference in the fracture strength of the teeth restored with all-ceramic crowns with 0.4- and 0.6-mm aluminum oxide copings, 0.6-mm zirconia ceramic copings, and metal ceramic crowns. Fracture mode of the crown was classified according to a classification proposed by Burke; Class I, minimal fracture or crack in crown; Class II, less than half of crown lost; Class III, crown fracture through midline, half of crown displaced or lost; Class IV, more than half of crown lost; Class V, severe fracture of tooth and/or crown. Visual analysis of the fractured specimens showed that all the specimens (100%) in every group exhibited a Class V mode of fracture. No crowns were dislodged from the prepared tooth, and there were no fractures of the all-ceramic or metal-ceramic crowns. All fractures occurred through the natural tooth.[80]

The objective of the study performed by Behrens et al. was to assess the fracture strength of colored zirconia copings with reduced wall thickness to 0.3 mm compared to a control group of 0.5 mm. Chamfer preparations for anterior teeth were made followed by the fabrication of the copings employing the Lava Y-TZP (3M ESPE) system. A total of 48 copings were produced out of colored Y-TZP-ZrO_2. Twenty-four copings were prepared from a tangential and twenty-four from a chamfer preparation and each of these were further divided in subgroups of 12 for the fabrication of wall thicknesses of 0.3 mm and 0.5 mm. Copings were cemented with Ketac-Cem on brass dies and loaded via universal testing machine until fracture occurred and fractography with SEM was performed. No significant difference was discovered ($p<0.05$) for the fracture strength of copings fabricated with tangential or chamfer preparation but with the same wall thickness. However, there was significant strength reduction between wall thickness of 0.5 and 0.3 mm (Kruskal-Wallis test, Dunn's post test, $p<0.001$) (Table 1). Therefore, it has been concluded that the fracture strength of Y-TZP-ZrO_2 copings with a wall thickness of 0.3 mm is three times lower compared to the expected chewing forces in the anterior region.[126]

Table 1. Fracture strength (N)[126]

Wall thickness	Tangential preparation			Chamfer preparation		
	Mean (SD)	Min	Max	Mean (SD)	Min	Max
0,3 mm	908 (115)	714	1185	913 (106)	729	1095
0,5 mm	1476 (192)	1075	1672	1333 (107)	1180	1527

Reich et al. studied zirconia copings manufactured with Lava CAD/CAM system. Different coping thickness between 0.5mm to 0.3mm as well as tangential versus chamfer preparation design was utilized. Four groups of ten copings each made of Y-ZrO_2 were produced with the Lava™ CAD/CAM system on respective standardized dies made of brass. The copings were fixed on the dies with Ketac™ Cem, stored for 24 hours in water of 37°C and fracture strength was tested in a universal testing with a crosshead speed of 0.5 mm/min. The data were statistically compared by One-way Anova and the Student-Newman-Keuls post hoc routine at $p \leq .05$. The result of the fracture toughness test performed by Universal testing presented that tangential preparation had significantly highest fracture forces for 0.5mm (1110 N) and 0.3mm (730 N) thickness of copings. [127]

In the study done by Van Der Zel et al. the effect of shoulder design on the failure load of zirconia crowns was assessed. Two groups of zirconia coping for anterior restorations were studied: overpressed crowns with a zirconia free PTC shoulder (CS) and overpressed crowns with zirconia up to the margin (CC). The zirconia-free shoulder extended 0.8 mm over the finishing line of the coping. Eight zirconia copings per group of first maxillary anteriors were fabricated with CerconBrain CAM system (DeguDent) with standard thickness of 0.6 mm. After milling, the copings were sintered at 1350°C to final density, coping was waxed-up to a standard contour, sprued and invested. The PTC Ceramic was pressed at 940°C over the zirconia coping and after devesting and separation from the sprues the crowns were veneered with two layers CerconCeramS porcelain. Crowns were cemented on a CoCr die with zinc phosphate cement and held under constant load of 5 kg. Failure loads were measured after vertical compression loading at 0.5 mm/min followed by SEM inspection for surface fracture. Failure loads [kN (SD)] were as follows: Group CS: 4228(515) and group CC: 5408(806). The study presented a significant ($p<0.05$) decrease of 22 % in breaking strength with the overhanging shoulder as compared to fully supported PTC crowns. Furthermore, surface fracture analysis revealed crack initiation typically located on the glass-zirconia interface. [128]

Unfortunately, none of the studies implemented the modification of the zirconia-based coping design in posterior area application.

4. References

1. Anusavice K. Phillips' Science of Dental Materials. 11th ed. St. Louis: Saunders; 2003.

2. Ring M. An illustrated history. 1985:180.

3. Yamada H. Dental porcelain: the state of the art. 1 ed. Los Angeles: University of Southern California School of Dentistry.; 1997.

4. Yoshinari M, Derand T. Fracture strength of all-ceramic crowns. Int J Prosthodont 1994;7(4):329-38.

5. Andersson M, Oden A. A new all-ceramic crown. A dense-sintered, high-purity alumina coping with porcelain. Acta Odontol Scand 1993;51(1):59-64.

6. Duret F, Blouin JL, Duret B. CAD-CAM in dentistry. J Am Dent Assoc 1988;117(6):715-20.

7. Duret F. The practical dental CAD/CAM in 1993. J Can Dent Assoc 1993;59(5):445-6, 48-53.

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

9. Abed HM. Studies on The Procera All-Ceramic Porcelain Crown. (Thesis) 2000.

10. Craig RG. Mechanical Properties in Restorative Materials. 11 ed: Mosby; 1997.

11. Raigrodski AJ. All-ceramic full-coverage restorations: concepts and guidelines for material selection. Pract Proced Aesthet Dent 2005;17(4):249-56; quiz 58.

12. Holand W, Rheinberger V, Apel E, van 't Hoen C, Holand M, Dommann A, et al. Clinical applications of glass-ceramics in dentistry. J Mater Sci Mater Med 2006;17(11):1037-42.

13. Albakry M, Guazzato M, Swain MV. Fracture toughness and hardness evaluation of three pressable all-ceramic dental materials. J Dent 2003;31(3):181-8.

14. Quinn JB, Sundar V, Lloyd IK. Influence of microstructure and chemistry on the fracture toughness of dental ceramics. Dent Mater 2003;19(7):603-11.

15. Taskonak B, Mecholsky JJ, Jr., Anusavice KJ. Fracture surface analysis of clinically failed fixed partial dentures. J Dent Res 2006;85(3):277-81.

16. Taskonak B, Sertgoz A. Two-year clinical evaluation of lithia-disilicate-based all-ceramic crowns and fixed partial dentures. Dent Mater 2006;22(11):1008-13.

17. Dong JK, Luthy H, Wohlwend A, Scharer P. Heat-pressed ceramics: technology and strength. Int J Prosthodont 1992;5(1):9-16.

18. Luthy H, Dong JK, Wohlwend A, Scharer P. Effects of veneering and glazing on the strength of heat-pressed ceramics. Schweizer Monatsschrift fur Zahnmedizin 1993;103(10):1257-60.

19. Magne P, Belser U. Esthetic improvements and in vitro testing of In-Ceram Alumina and Spinell ceramic. International Journal of Prosthodontics 1997;10(5):459-66.

20. Hwang JW, Yang JH. Fracture strength of copy-milled and conventional In-Ceram crowns. Journal of Oral Rehabilitation 2001;28(7):678-83.

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

22. Fradeani M, Redemagni M. An 11-year clinical evaluation of leucite-reinforced glass-ceramic crowns: a retrospective study. Quintessence Int 2002;33(7):503-10.

23. Heffernan MJ, Aquilino SA, Diaz-Arnold AM, Haselton DR, Stanford CM, Vargas MA. Relative translucency of six all-ceramic systems. Part II: core and veneer materials. J Prosthet Dent 2002;88(1):10-5.

24. Guazzato M, Proos K, Sara G, Swain MV. Strength, reliability, and mode of fracture of bilayered porcelain/core ceramics. Int J Prosthodont 2004;17(2):142-9.

25. Guazzato M, Albakry M, Swain MV, Ironside J. Mechanical properties of In-Ceram Alumina and In-Ceram Zirconia. Int J Prosthodont 2002;15(4):339-46.

26. Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of all-ceramic materials. Part I. Pressable and alumina glass-infiltrated ceramics. Dent Mater 2004;20(5):441-8.

27. Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of all-ceramic materials. Part II. Zirconia-based dental ceramics. Dent Mater 2004;20(5):449-56.

28. Suarez MJ, Lozano JF, Paz Salido M, Martinez F. Three-year clinical evaluation of In-Ceram Zirconia posterior FPDs. Int J Prosthodont 2004;17(1):35-8.

29. Tinschert J, Natt G, Hassenpflug S, Spiekermann H. Status of current CAD/CAM technology in dental medicine. Int J Comput Dent 2004;7(1):25-45.

30. Duret F. Dental CAD/CAM. J Am Dent Assoc 1992;123(6):11-2, 14.

31. Sun F, Zhang GR, Zhang F, Liu F, Mao H, Huang L, et al. [The use of CAD/CAM system with zirconia in modern prosthodontics]. Shanghai Kou Qiang Yi Xue 2006;15(4):337-44.

32. Witkowski S. CAD/CAM inDental Technology. Quintessence Dent Technol 2005:1-16.

33. Andersson M, Oden A. A new all-ceramic crown. Acta Odontol Scand 1993;51(1):59-64.

34. Andersson M, Razzoog ME, Oden A, Hegenbarth EA, Lang BR. Procera: a new way to achieve an all-ceramic crown. Quintessence Int 1998;29(5):285-96.

35. Zeng K, Oden A, Rowcliffe D. Flexure tests on dental ceramics. Int J Prosthodont 1996;9(5):434-9.

36. Wagner WC, Chu TM. Biaxial flexural strength and indentation fracture toughness of three new dental core ceramics. J Prosthet Dent 1996;76(2):140-4.

37. Christel P, Meunier A, Heller M, Torre JP, Peille CN. Mechanical properties and short-term in-vivo evaluation of yttrium-oxide-partially-stabilized zirconia. J Biomed Mater Res 1989;23(1):45-61.

38. Castellon P, Potiket N, Soltys JL, Johnson J, Zavala J. All-ceramic restorative system for esthetic implant-supported crowns: in vitro evaluations and clinical case report. Compend Contin Educ Dent 2003;24(9):673-6, 78, 80-3; quiz 84.

39. Raigrodski AJ. Contemporary materials and technologies for all-ceramic fixed partial dentures: a review of the literature. J Prosthet Dent 2004;92(6):557-62.

40. Luthardt RG, Holzhuter MS, Rudolph H, Herold V, Walter MH. CAD/CAM-machining effects on Y-TZP zirconia. Dent Mater 2004;20(7):655-62.

41. Razzoog ME, Lang LA, McAndrew KS. AllCeram crowns for single replacement implant abutments. J Prosthet Dent 1997;78(5):486-9.

42. O'Brien WJ. Dental Materials and Their Selection. third ed; 2002.

43. Griffith AA. The phenomenon of rupture and flow in solids.: Phil. Trans. Roy. Soc. Lond, ; 1920.

44. Flinn. Engeneering materials and their application. 2 ed: Houghton Mifflin, Boston, MA; 1981.

45. O'Brien WJ. High expansion feldspathic porcelain and magnesia core material. . Ceramic Engineering and Science Proceedings 1985;6(1):10-18.

46. McLean JW. The bonded alumina crown. The bonding of platinum to aluminous dental porcelain using tin oxide coatings. Austral. Dent J. 1976;21(2):119-27.

47. Jones DW. The strength and strengthening mechanisms of dental ceramics. (in McLean, J.W. ed. Dental Ceramics. Procedeengs of the First International Symposium on Ceramics, Chicago, Quintessence) 1983:83-141.

48. Batchelor. Trans. VIIth Int. Ceram. Congr. London. 1960:31.

49. Binns. The science of ceramics, Academic press, London 1960:315-34.

50. O'Brien WJ. Properties of a new high expansion core material for porcelain crowns. J. Dent. Res.Abstract No.410 1984;36.

51. White SN. Mechanical fatigue of a feldspathic dental procelain. Dent Mater 1993;9(4):260-4.

52. Edwards MR, Jacobsen PH, Williams GJ. The three-point beam test for the evaluation of dental porcelain. J Dent Res 1983;62(10):1086-8.

53. Jones DW, Jones PA, Wilson HJ. The relationship between transverse strength and testing methods for dental ceramics. J Dent 1972;1(2):85-91.

54. Shimizu K, Oka M, Kumar P, Kotoura Y, Yamamuro T, Makinouchi K, et al. Time-dependent changes in the mechanical properties of zirconia ceramic. J Biomed Mater Res 1993;27(6):729-34.

55. Ichikawa Y, Akagawa Y, Nikai H, Tsuru H. Tissue compatibility and stability of a new zirconia ceramic in vivo. J Prosthet Dent 1992;68(2):322-6.

56. Jones DW. The strength and strengthening mechanisms of dental ceramics. (in McLean, J.W. ed Dental Ceramics, Proceedings of the First International Symposium on Ceramics, Chicago, Quintessence 1983:94.

57. Tinschert J, Zwez D, Marx R, Anusavice KJ. Structural reliability of alumina-, feldspar-, leucite-, mica- and zirconia-based ceramics. J Dent 2000;28(7):529-35.

58. Jung YG, Peterson IM, Kim DK, Lawn BR. Lifetime-limiting strength degradation from contact fatigue in dental ceramics. J Dent Res 2000;79(2):722-31.

59. Sced IR, McLean JW, Hotz P. The strengthening of aluminous porcelain with bonded platinum foils. J Dent Res 1977;56(9):1067-9.

60. Liu CC. Strength of magnesia core porcelain crowns [thesis; Ann Arbor: University of Michigan; 1989.

61. Tan SR, May GJ, McLean JR. Strength controlling flaws in beta-alumina. Trans. J. Br. Ceram. Soc. 1980;79(3):12127.

62. Edwards MR, Jacobsen OH. The "C" -test for the evaluation of dental porcelain. J Dent Res 1985;64(9):1176-78.

63. Bortz SA, Lund HH. The mechanical Properties of Engeneering Ceramics. In Kriegel W.W and Palmour H. editors: Interscience Publishers 1961:383.

64. Hondrum SO. The strength of ceramic crowns tested using different methods [thesis; Ann Arbor: University of Michigan]; 1986.

65. Genho BL, Christensen RP, Harput JK, Wagstaff RS. 2-year clinical performance of Procera/All-Ceram single unit crowns. IADR General Session 2002.

66. Neiva G, Yaman P, Dennison JB, Razzoog ME, Lang BR. Resistance to fracture of three all-ceramic systems. J Esthet Dent 1998;10(2):60-6.

67. Webber B, McDonald A, Knowles J. An in vitro study of the compressive load at fracture of Procera AllCeram crowns with varying thickness of veneer porcelain. J Prosthet Dent 2003;89(2):154-60.

68. Sato T, Al Mutawa N, Okada D, Hasegawa S. A clinical study on abutment taper and height of full cast crown preparations. J Med Dent Sci 1998;45(3):205-10.

69. Christensen GJ. Ensuring retention for crowns and fixed prostheses. J Am Dent Assoc 2003;134(7):993-5.

70. Dodge WW, Weed RM, Baez RJ, Buchanan RN. The effect of convergence angle on retention and resistance form. Quintessence Int 1985;16(3):191-4.

71. Kvam K, Hero H, Oilo G. Fracture toughness measurements of some dental core ceramics: a methodologic study. Scand J Dent Res 1991;99(6):527-32.

72. McLaren EA, White SN. Glass-infiltrated zirconia/alumina-based ceramic for crowns and fixed partial dentures. Pract Periodontics Aesthet Dent 1999;11(8):985-94; quiz 96.

73. Smith TB, Kelly JR, Tesk JA. In vitro fracture behavior of ceramic and metal-ceramic restorations. J Prosthodont 1994;3(3):138-44.

74. Aboushelib MN, de Jager N, Kleverlaan CJ, Feilzer AJ. Microtensile bond strength of different components of core veneered all-ceramic restorations. Dent Mater 2005;21(10):984-91.

75. Al-Dohan HM, Yaman P, Dennison JB, Razzoog ME, Lang BR. Shear strength of core-veneer interface in bi-layered ceramics. J Prosthet Dent 2004;91(4):349-55.

76. Kelly JR, Tesk JA, Sorensen JA. Failure of all-ceramic fixed partial dentures in vitro and in vivo: analysis and modeling. J Dent Res 1995;74(6):1253-8.

77. Proos K, Steven G, Swain M, Ironside J. Preliminary studies on the optimum shape of dental bridges. Comput Methods Biomech Biomed Engin 2000;4(1):77-92.

78. White SN, Caputo AA, Vidjak FM, Seghi RR. Moduli of rupture of layered dental ceramics. Dent Mater 1994;10(1):52-8.

79. Guazzato M, Proos K, Quach L, Swain MV. Strength, reliability and mode of fracture of bilayered porcelain/zirconia (Y-TZP) dental ceramics. Biomaterials 2004;25(20):5045-52.

80. Potiket N, Chiche G, Finger IM. In vitro fracture strength of teeth restored with different all-ceramic crown systems. Journal of Prosthetic Dentistry 2004;92(5):491-5.

81. Pallis K, Griggs JA, Woody RD, Guillen GE, Miller AW. Fracture resistance of three all-ceramic restorative systems for posterior applications. J Prosthet Dent 2004;91(6):561-9.

82. Gardner FM. Margins of complete crowns--literature review. J Prosthet Dent 1982;48(4):396-400.

83. Lang NP, Kiel RA, Anderhalden K. Clinical and microbiological effects of subgingival restorations with overhanging or clinically perfect margins. J Clin Periodontol 1983;10(6):563-78.

84. American Dental Assocation: ANSI/ADA Specification No. 8 for zinc phosphate cement. Guide to dental materials and devices. 5th ed. Chicago: American Dental Association 1970:71.

85. Hertlein G. Marginal fit of zirconia restorations with three/four abutment teeth. IARD, 3M ESPE AG, Seefeld, Germany 2005.

86. Yeo IS, Yang JH, Lee JB. In vitro marginal fit of three all-ceramic crown systems. J Prosthet Dent 2003;90(5):459-64.

87. Hilgert E, Neisser MP, Vasquez EMV, Costa V, Giannini V, Bondioli IR. Evaluation of marginal adaptation of ceramic crowns depending on the marginal design and the addition of ceramic. IARD General Session of the International Association for Dental Research 2003.

88. Al-Ghazzawi T, Essig M, Liu P, Lacefield WR, Chen DT. Marginal adaptation of CAD-CAM generated alumina and zirconia copings. IADR General Session 2004.

89. Rosentritt M, Sikora M, Behr M, Handel G. Fracture Strength and Marginal Adaptation of All-Ceramic FPDs ADEA/AADR/CADR Meeting & Exhibition 2006.

90. Komine F, Iwai T, Shiratshuchi H, Matsumura H. Adaptation of zirconia ceramic copings with different finish line designs. IADR General Session & Exhibition 2006:1287-8.

91. Blatz MB, Sadan A, Kern M. Resin-ceramic bonding: a review of the literature. J Prosthet Dent 2003;89(3):268-74.

92. Luthy H, Loeffel O, Hammerle CH. Effect of thermocycling on bond strength of luting cements to zirconia ceramic. Dent Mater 2006;22(2):195-200.

93. Escribano N, de la Macorra JC. Microtensile bond strength of self-adhesive luting cements to ceramic. J Adhes Dent 2006;8(5):337-41.

94. Ernst CP, Cohnen U, Stender E, Willershausen B. In vitro retentive strength of zirconium oxide ceramic crowns using different luting agents. J Prosthet Dent 2005;93(6):551-8.

95. Kim BK, Bae HE, Shim JS, Lee KW. The influence of ceramic surface treatments on the tensile bond strength of composite resin to all-ceramic coping materials. J Prosthet Dent 2005;94(4):357-62.

96. Blatz MB, Sadan A, Martin J, Lang B. In vitro evaluation of shear bond strengths of resin to densely-sintered high-purity zirconium-oxide ceramic after long-term storage and thermal cycling. J Prosthet Dent 2004;91(4):356-62.

97. Martin J, Sada A, Castellon P, Burgess JO, Blatz MB. In vitro comparative shear bond strength to Procera All Zircon. IADR General Session 2003.

98. Diaz-Arnold AM, Aquilino SA. An evaluation of the bond strengths of four organosilane materials in response to thermal stress. J Prosthet Dent 1989;62(3):257-60.

99. O'Kray K, Aj S, JW.. S. Shear strength of porcelaion repair materials. J Dental Res; 1987.

100. Matinlinna JP, Heikkinen T, Ozcan M, Lassila LV, Vallittu PK. Evaluation of resin adhesion to zirconia ceramic using some organosilanes. Dent Mater 2006;22(9):824-31.

101. Derand T, Molin M, Kvam K. Bond strength of composite luting cement to zirconia ceramic surfaces. Dental Materials 2005;21(12):1158-62.

102. McLean JW. New dental ceramics and esthetics. J Esthet Dent 1995;7(4):141-9.

103. Al-Dohan HM. Shear strength of core-veneer interface in bi-layered ceramics [The University of Michigan; 2003.

104. Helmer J. Research on bioceramics. In:Symposium on Use of Ceramics as Surgical Implants. South Carolina, Clemson University 1969.

105. Christel P, Meunier A, Dorlot JM, Crolet JM, Witvoet J, Sedel L, et al. Biomechanical compatibility and design of ceramic implants for orthopedic surgery. Ann N Y Acad Sci 1988;523:234-56.

106. Keith O, Kusy RP, Whitley JQ. Zirconia brackets: an evaluation of morphology and coefficients of friction. Am J Orthod Dentofacial Orthop 1994;106(6):605-14.

107. Edelhoff D, Sorensen JA. Retention of selected core materials to zirconia posts. Oper Dent 2002;27(5):455-61.

108. Heydecke G, Butz F, Hussein A, Strub JR. Fracture strength after dynamic loading of endodontically treated teeth restored with different post-and-core systems. J Prosthet Dent 2002;87(4):438-45.

109. Glauser R, Sailer I, Wohlwend A, Studer S, Schibli M, Scharer P. Experimental zirconia abutments for implant-supported single-tooth restorations in esthetically demanding regions: 4-year results of a prospective clinical study. Int J Prosthodont 2004;17(3):285-90.

110. Kohal RJ, Weng D, Bachle M, Strub JR. Loaded custom-made zirconia and titanium implants show similar osseointegration: an animal experiment. J Periodontol 2004;75(9):1262-8.

111. Porter D, Heuer H. Reply to further discussion of precipitation in partially stabilized zirconia. J. Am.Ceram. Soc. 1977;60:280-81.

112. O'Brien WJ. Strengthening mechanisms of current dental porcelains. Compend Contin Educ Dent 2000;21(8):625-30; quiz 32.

113. Tateishi T. Research and development of advanced biocomposite materials and application to the artificial hip joint. Bull Mech Eng Lab Jap 1987;45:1-9.

114. American E. Advanced Material Informations. http://www.americanelements.com/newpage3.htm 2005.

115. Takagi H, Nishioka K, Kawanami T, al e. The properties of a closely sintered zirconia. Ceram Forum Int 1985;62:195-98.

116. 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(3):231-8.

117. Hauptmann H, Suttor D, Hoescheler S, S F. Material properties of all-ceramic zirconia prosthesis. ESPE Dental AG, 82229 Seefeld, Germany 2000.

118. Ardlin BI. Transformation-toughened zirconia for dental inlays, crowns and bridges: chemical stability and effect of low-temperature aging on flexural strength and surface structure. Dent Mater 2002;18(8):590-5.

119. Hogg KD. Load to fracture of different all- ceramic croown systems. IADR General Session 2004.

120. Snyder MD, Razzoog ME. Comparison of compressive strength: zirconium vs. aluminum oxide copings. IADR General Session 2002.

121. Bates JF, Stafford GD, Harrison A. Masticatory function - a review of the literature. III. Masticatory performance and efficiency. J Oral Rehabil 1976;3(1):57-67.

122. Kelly JR. Clinically relevant approach to failure testing of all-ceramic restorations. Journal of Prosthetic Dentistry 1999;81(6):652-61.

123. Sorensen JA, Kang SK, Torres TJ, Knode H. In-Ceram fixed partial dentures: three-year clinical trial results. J Calif Dent Assoc 1998;26(3):207-14.

124. Curtis AR, Wright AJ, Fleming GJ. The influence of simulated masticatory loading regimes on the bi-axial flexure strength and reliability of a Y-TZP dental ceramic. J Dent 2006;34(5):317-25.

125. Fischer J. Strength of zirconia single crowns related to coping design. IADR General Session 2005.

126. Behrens A. Fracture strength of colored zirconia copings with reduced wall thickness. IADR General Session 2004.

127. Reich S, Reusch B. Fracture force of ZrO2 copings dependent on preparation and thickness. IADR General Session 2003.

128. Van Der Zel JM, Grinwis T, De Kler M, Tsadok Hai T. Effect of shoulder design on failure lodad of PTCercon crowns. IADR General Session 2004.

CHAPTER II

1. Abstract

Statement of problem: The recommended thickness of zirconium coping supporting veneered ceramic restorations is dependent on the manufacturer, ranging from 0.5-0.8 mm and extending to the cavosurface margins of the tooth preparation. This coping design contributes to an opaque and unnatural appearance, particularly at the cervical third of the restoration.

Purpose: The purpose of this study was to investigate the effect of zirconium oxide coping modification on the fracture resistance and fracture location of all-ceramic restorations using an axial load testing methodology. Fracture analysis was conducted on all samples using a new fracture classification.

Materials and methods: A stainless steel block was milled to simulate a preparation on a maxillary premolar: 1.0 mm wide cervical chamfer, round internal line angles, 1.5 mm axial reduction and 2.0 mm rounded occlusal reduction. Thirty-six high-filler content resin dies were fabricated to replicate the model. Thirty-six zirconium oxide copings (Procera AllZircon) were fabricated according to three different coping designs (n=12) using CAD/CAM technology: Design 1) coping extending to cavosurface finish line of the preparation, Design 2) coping cut back to the axial cervical line angle, Design 3) coping with a lingual full shoulder and cut-back to the middle of the facial axial wall. All cores were air abraded (50µm Al_2O_3), steam cleaned and covered with veneering porcelain to predetermined dimensions of 1.5 mm axial and 2.0 mm occlusal thickness.

After firing, the samples were air-abraded (50µm Al_2O_3) at 80 psi, cleaned and dried. All crowns were treated with silane coupling agent and cemented with Panavia 21 (J. Morita Inc., Tustin, CA, USA) resin cement. Specimens were placed under a 5-kg uniform static load for 5 minutes and stored in 100 % humidity at 37ºC for 24 h. Compressive load was applied to the specimen long axis 2 mm from the external occlusal edge in a universal load-testing machine at a cross head speed of 0.5 mm/min until failure. One-Way ANOVA test was used to determine the significance of the failure loads between groups ($p<0.01$). Failure modes were classified into rating scales for veneering porcelain (P), core (C) and die (D). One specimen of each design was examined microscopically at 7.5, 8.5, 10 and 20X to evaluate the cement space.

Results: The mean maximum loads in Newtons were similar for Design 1 (1773.13 ± 235.05), and Design 2 (1653.08 ± 380.63), but significantly lower for Design 3 (1256.71 ± 190.78) at ANOVA $p<0.01$. Pearson Chi-Square showed significant differences in fracture modes in porcelain between Design 1 and 2 (0.034) and Design 1 and 3 (0.03), while no difference was found between Design 2 and 3 (0.819) at $p<0.05$. Comparison of fracture mode in core material and design revealed significant differences in Design 1 and 2 (0.018) and Design 1 and 3 (0.007), while no difference was found between Design 2 and 3 (0.165) at $p<0.05$.

Conclusion: No difference was found within mean maximum loads between Designs 1 and 2 while Design 3 showed decreased maximum load resistance. Porcelain fracture was more catastrophic in Design 2 and 3 than Design 1.

Clinical relevance: This in-vitro study provides a basis for clinicians to select the appropriate all-ceramic zirconia core design that would offer superior esthetics without compromising the strength of the restoration.

2. Introduction

Patients have become increasingly concerned about the biocompatibility and esthetic properties of the materials used in dentistry. The conventional porcelain-fused to metal restoration is the most popular treatment option for most dentists due to their long term clinical application, well known fabrication technique, high strength with a superior metal accuracy and physical properties. Metal ceramic fixed partial dentures have proven to be successful 80% after 15 years, and 53% after 30 years and recognized as a gold standard in terms of success rate and restoration predictability.[1, 2] Porcelain fused to metal restorations are composed of a metal casting or coping that is covered with a layer of ceramics. The metal coping may be designed to extend all the way to the shoulder margin or cut back and replaced with a porcelain shoulder to hide the metal substructure. The thickness of the metal or the coping design has an important effect on the success or failure as well as on esthetics of the restoration. Maximum coping strength and restoration longevity is accomplished by using a rigid coping that will withstand the occlusal forces. For adequate strength and rigidity, a thickness of 0.3 to 0.5 mm of metal substructure is required.[3] According to Kuwata, the 50 degree critical margin angle allows the thickness of sufficient amount of metal, 0.3 mm, opaque layer of 0.25 mm and porcelain of 0.216 mm.[4] The superior marginal seal of 50 µm or less is achieved with a featheredge margin, while occlusal seat of less than 85 µm occurs when a 90º shoulder is implemented.[5] Metal-ceramic materials possess adequate mechanical properties, but often lack the superior esthetics. Gingival discoloration surrounding the margins or even metal exposure due to gingival recession have been a common unesthetic and undesirable outcome.[6] The modification of the coping has been developed where the framework shoulder is cut back 1 to 3 mm to enhance the esthetics, reduce the facial metal margin exposure and to allow the higher translucency and natural appearance to occur.[7, 8] That modification introduced a controversy regarding the framework's ability to adequately resist masticatory forces and maintain the physical properties. It was concluded that collarless metal-ceramic restorations were able to withstand the forces when a 90 degree shoulder preparation was employed with a metal substructure cut back up to 2 mm.[9, 10] In many situations where the use of alloy or gold supported restorations was previously indicated, they are replaced today by all-ceramic restorations. Ahnlide et al. have demonstrated an association between gold allergy and the presence of dental gold restorations and indicated that "gold is released from dental restorations and taken up into the circulation".[11] It was assumed that well known gingival reactions and eventual tissue biological incompatibility to metal often observed in all-metal or porcelain fused to metal restorations will be completely eliminated with all-ceramic crown replacement.[12]

Any dental material requires sufficient physical properties to achieve good esthetic results, marginal integrity and high strength to withstand an occlusal load. However, elimination of the metal substructure has raised concerns about the resistance to fracture of some of the all ceramic restorations.

The high mechanical properties of the all ceramic restorations has been maintained by developing high strength core ceramics that are classified according to chemical content: glass ceramics, glass-infiltrated ceramics and polycrystalline ceramics.[13] Glass ceramics are partially crystallized, amorphous glasses that are produced by enucleation and growth of crystals in the glass matrix phase.[14] Glass infiltrated core ceramics are glass infiltrated to partially sintered oxides. That group is mainly represented by In-Ceram Alumina, In-Ceram Spinell and In-Ceram Zirconia. In-Ceram Spinell was marketed more recently to improve the esthetic potential. All ceramic restorations offer improved esthetic results, compared to metal-ceramic restorations, due to the natural translucent and reflection effect.[15, 16] Certain all-ceramic restorative materials have been recommended based on the translucency of the teeth to be matched. In vivo evaluation of specific esthetic parameters inherent to different types of cores was made and revealed the relative opacity of alumina while spinell was found to have the ability to blend in with the underlying substrate. Nevertheless both materials demonstrated a general lack of fluorescence.[17]

Polycrystalline ceramic material is composed of densely sintered particles with no glassy components and is solely processed by CAD/CAM technology. Studies have demonstrated that zirconia-based industrially optimized ceramic materials are more structurally reliable for dental application and comparatively superior in terms of mechanical properties.[18-22] Yttrium tetragonal zirconia polycrystals (Y-TZP) is a glass-free ceramic material containing about 3% mol Y_2O_3 with high flexural strength and fracture toughness. Guazzato et al. investigated the strength, fracture toughness and microstructure of ceramic materials, and demonstrated that zirconia-based dental porcelain is a stronger and tougher material compared to a conventional glass-ceramic, having fracture toughness values ranging from 680-840 MPa and 476-630 MPa, respectively.[21] According to Christel et al. yttrium tetragonal zirconia material has a flexural strength of 900-1200 MPa and a fracture toughness of 9-10 MPa x $m^{1/2}$.[23] Other studies demonstrated that zirconia-YTZ material has a fracture strength of 420 ± 82.45 kgF, 913.0 ± 50.2 MPa or 2000 N.[22, 24, 25] High physical strength allows the zirconia core to withstand normal physiologic occlusal values ranging from 2-150 N as well as extreme masticatory forces of 880 N exhibited in bruxism.[26-29]

In an effort to improve strength, various core substrates were developed that are becoming increasingly popular due to improved biocompatibility, physical, mechanical and esthetic properties. Significant research has been done to develop materials that would combine good strength and better translucency and overall esthetics.[15, 16, 30-32]

The coping design, its crystalline structure, and length or thickness influences the strength, adaptation and, more importantly, the esthetics. Potiket reported that there was no significant difference in the fracture strength of teeth restored with all-ceramic crowns with 0.4 mm and 0.6 mm aluminum oxide copings, 0.6 mm zirconia ceramic copings, and metal ceramic crowns.[31] Studies on Procera CAD/CAM using alumina all-ceramic copings of 0.4 mm thickness demonstrated clinically acceptable fit at the margin ranging from 44 to 59 µm in the premolar to central incisor area respectively.[33] Behrens assessed the fracture strength of colored zirconia copings with reduced wall thickness to 0.3 mm and concluded that the copings demonstrated three times higher fracture strength compared to the expected chewing forces in the anterior region.[30] Another study affirmed that fracture forces of the coping with reduced wall thickness of 0.3 mm were significantly lower (730 N) compared to 0.5 mm thick copings (1110 N).[34]

Currently the recommended width of zirconium coping supporting the veneered ceramic restoration is dependent on the manufacturer and ranges from 0.5-0.8 mm, with the coping shoulder covering the margins of the tooth preparation. Unfortunately the application of the recommended coping design contributes to an opaque and unnatural appearance, particularly at the cervical third of the restoration contour. The purpose of this study was to determine the effect of modified zirconium copings on the fracture resistance of Procera® All-Zircon. Additionally, the effect of the coping design modification on the fracture location was investigated.

3. Research Design and Methods

3.1. Specimen Preparation

A stainless steel master die was machined to simulate the dimensions and contours of a crown preparation for all-ceramic crowns with a zirconium core on a maxillary premolar: 1.0 mm deep chamfer, round internal line angles, 1.5 mm axial reduction, 2.0 mm rounded occlusal reduction (Figure 3). Thirty-nine duplicate dies were poured in a high-filler content resin (Viade Products Inc., Camarillo, CA, USA) (Figure 3). The bottom surfaces of dies were trimmed to achieve an occlusal plane perpendicular to the long axis of the tooth. The purpose of using the resin material was to closely mimic a natural tooth's physical properties where the resin's modulus of elasticity (12.9 MPa) resembles that of human dentin (14.7 MPa).[35, 36]

Thirty-six replicated dies were divided into three groups of 12 each within the three coping designs (Table 2). The control group had ceramic copings extended to the finishing line of the preparation as recommended by manufacturer (Figure 4). In the second group the entire zirconia ceramic coping was cut back to the axial cervical line angle (Figure 5). The third group was designed to have a full coping shoulder on the lingual and a cut-back to the middle of the axial wall on the facial (Figure 6).

Table 2. Grouping of Samples for Strength Test and Fracture Location

	Control (D1)	Shoulder- free (D2)	Half axial cut-back (D3)	Total
Total	12	12	12	36

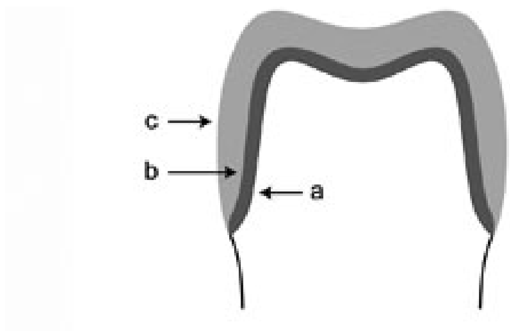


Figure 4. Control group - zirconia coping

a - crown die
b - zirconia coping
c - veneering porcelain

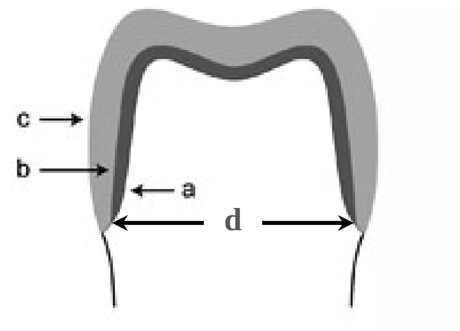


Figure 5. Shoulder - free zirconia coping

a - crown die
b - modified zirconia coping
c - veneering porcelain
d - shoulder cut back

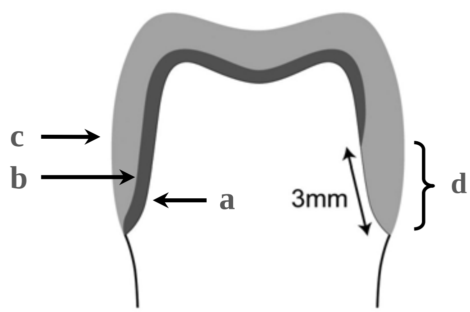


Figure 6. Facial half cut-back coping

a - crown die
b - modified zirconia coping
c - veneering porcelain
d - facial shoulder cut back

The bases of two epoxy dies were modified to facilitate the scanning procedure for variations in coping Designs 2 and 3 (Figure 3). The three final dies were then scanned using a Procera Scanner (Piccolo, Nobel Biocare AB, Goteborg, Sweden) with a 1.25 mm diameter scanning probe. The probe contacted the die at a 45-degree angle and moved from the cervical margin to the occlusal surface as the die rotated 360 degrees. Once the scanning process was completed the electronic order was placed to have zirconia copings fabricated at Nobel Biocare Procera production facility (Nobel Biocare, Mahwah, NJ, USA). Thirty-six Procera ® All-Zircon copings (12 of each design) were pre-sintered, milled and sintered to a thickness of 0.5-0.7 mm following manufacturer's guidelines. The manufactured coping thicknesses were measured at specific locations using a dial caliper (Kori Dial Caliper, Pfingst &Company, Inc., Tokyo, Japan) (Figure 7).

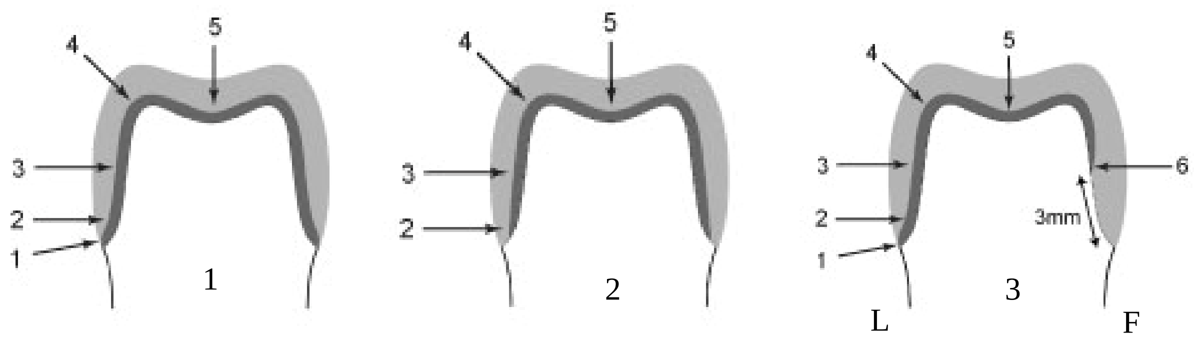


Figure 7. Coping measuring points

The surface of the zirconia cores were air abraded for 10 seconds with 50 µm Al_2O_3 at 80 psi (Pure Blast, Macro Cab, Danville Engineering Inc., USA) at a distance of approximately 50 mm and steam cleaned (Steaman II, Bar Instruments, CA, USA). Low-fusing veneering body porcelain (color - A2B Cerabien ZR, Noritake Dental Supply Co., Nagoya, Japan) was mixed with wetting agent (Sta-Wet Liquid, Ceramco, Burlington, NJ, USA) and applied to the cores using a traditional condensation technique to cover the zirconia coping. The sculpting device developed by Philip and Brukl was used to produce constant specified thickness of the veneering porcelain and flat occlusal surface (Figure 8).[37] The dimensions of the finished specimens were an axial thickness of 1.5 mm and an occlusal thickness of 2.0 mm. All specimens had two layers of body porcelain (A2B) and one glazing layer applied. Additionally, specimens in Design 2 had 2 layers of body porcelain applied on the shoulder while Design 3 specimens had 4 layers of body porcelain applied on the exposed shoulder. The ceramic specimens within Design 2 and 3 that required repair had additional veneering porcelain or glazing applied after Custom-Peg Putty (Hankins Laboratories, Cupertino, CA, USA) was placed on the internal surface of the specimens to maintain internal and margin integrity and to prevent distortion. Each porcelain and glazing layer was individually fired (VITA Vacumat 40®T Vident, CA, USA) according to the manufacturer's firing schedule (Table 3).

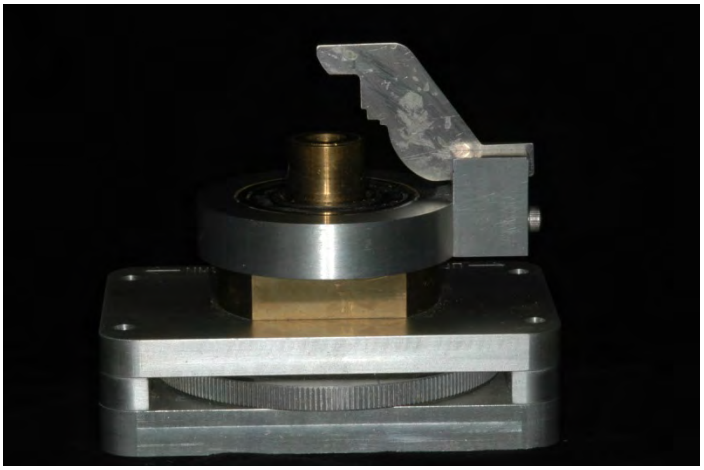


Figure 8. Sculpting device

Table 3. Firing Schedules

All-ceramic system	Time (min.)		Temp (ºC)		Vacuum (ºC)		Heat Rate ºC/min
	Dry	Hold	Low	High	Start	Stop	
Cerabien CZR Body	7	1	600	930	600	930	45
Cerabien CZR Glazing powder	5	-	600	920	-	-	50

All measurements were performed using a dial caliper and any excessive thickness was adjusted with a porcelain bur (#H79DF, Brasseler, GA, USA). All specimens had a glazing powder applied (Cerabien ZR, Noritake Dental Supply Co., Nagoya, Japan) and were fired (Table 3). After the glazing procedure, the specimens were remeasured to confirm the required thickness. The internal surface was air abraded for 10 seconds with 50 µm aluminum oxide at 80 psi (Pure Blast, Macro Cab, Danville Engineering Inc., USA) at a distance of approximately 10 mm, ultrasonically cleaned in distilled water for 10 minutes and air dried. Mixed primer and bond activator (ED Primer liquid A and ED Primer liquid B, J Morita Inc., Tustin, CA, USA) were mixed for 3 seconds and applied to the surface of the duplicate dies. After 60 seconds, dies were gently air dried until the glossy surface appeared. The internal ceramic surface was etched with 37% phosphoric acid (Panavia Etching Agent V, J. Morita Inc., Tustin, CA, USA), rinsed with distilled water and air dried followed by the application of silane coupling agent (Calibra Dentsply Caulk, York, PA, USA) that was allowed to air dry. Universal paste and catalyst (Panavia 21 OP, J. Morita Inc., Tustin, CA, USA) were mixed for 20 seconds creating a uniform paste and applied to the internal ceramic surface of the individual specimens. All restorations were luted with firm finger pressure to the tooth replica. Immediately after cementation the specimens were placed under a 5-kg uniform static load for 5 minutes and cement excess was removed. To accomplish complete polymerization of the resin cement, Oxyguard II (J. Morita Inc., Tustin, CA, USA) was placed around the restoration margins. Following removal of residual cement, the specimens were stored in 100% humidity at a temperature of 37˚C for 24 hours.

2.2 Strength and Fracture Location Test Measurement

A stainless steel rod 1.0 mm in diameter was placed 2 mm from the external edge of the restoration over the occlusal surface of each specimen. Each bonded specimen was held with a metal device and axially loaded using a screw-driven Instron Universal Testing Machine (Model 5560, Instron Corp., Canton, MA, USA) at a cross head speed of 0.5 mm/min until fracture occurred. Fracture initiation was determined by the appearance of a dip on the computerized chart and the sound of the fracture. The maximum and fracture loads were quantified for each group of 12 specimens to obtain a range, mean value and standard deviation. All samples were visually evaluated and the mode of fracture was classified according to a classification proposed by Fotek (Table 4). A rating greater than 0 in any of the 3 parameters tested was recognized as clinically unacceptable. To examine the space and cement thickness one sample was selected from each of the designs (Design 1- #2, Design 2- #10, and Design 3- #12) in order to represent an average of fracture mode frequencies and maximum load values. Each of those specimens was reassembled with cyanoacrylate cement (Super Glue, Super Glue Corp., Hollis, NY, USA) and embedded in self-curing methacrylate based resin (Koldmount, Vernon-Benshoff Co., Albany, NY, USA). All three samples were sectioned parallel to the facial-lingual long axis of the die and through the center using a low-speed diamond saw (SBT model 650, South Bay Technology Inc., San Clemente, CA, USA) under distilled water cooling (Figure 9). Each of the sectioned samples was polished under water cooling with 400 grit silicon carbide paper (Forcimat, Micro Star 2000, Inc., Concord, Canada) (Figure 10). Photographs were taken with a 35 mm Nikon camera (Nikon SMZ1500, Nikon Inc., Japan) and the Spot Advanced Software (Diagnostic Instruments, Sterling Heights, MI, USA) at 7.5, 8.5, 10 and 20X magnification. To evaluate the fracture resistance of the epoxy die, twelve samples were tested using universal testing machine where the stainless steel rod 1.0 mm in diameter was placed in the center of the occlusal surface of each die. Each specimen was held with a metal device and axially loaded at a cross head speed of 0.5 mm/min until fracture occurred. The maximum and fracture loads were quantified for each of the twelve specimens to obtain a range, mean value and standard deviation.

Table 4. Fracture Classification

P (porcelain)	**C (coping)**	**D (die)**
0 - no evidence of fracture	**0** - no evidence of fracture	**0** - no evidence of fracture
1 - fracture line	**1** - crack line (closed)	**1** - horizontal fracture
2 - chip not extending more than ½ of the crown circumference	**2** - fracture line (open)	**2** - vertical fracture
3 - chip extending more than ½ of the crown circumference	**3** - chip	**3** - combination fracture
4 - chip infringing the shoulder		

Figure 9. Diamond saw

Figure 10. Polisher

4. Statistical Analysis of Data

Mean fracture strength values were analyzed using independent student t-test. A one-way analysis of variance (ANOVA) was carried out to determine the presence of statistical difference in maximum load and fracture load among the three coping designs. Tukey multiple comparison test was used to determine statistical significance among the three coping designs. Pearson Chi-square test was used to determine whether or not paired observations on both variables (substrate and design) were independent of each other. All statistical analysis was carried out using statistical software (SYSTAT Software Inc., Evanston, IL, USA) at α=0.05.

5. Results

Table 5 shows the coping thickness values (Figure 7) ranging on the cervical (locations 1 and 2) from 0.4-0.6 mm, axial (location 3) from 0.6-0.7 mm, occluso-axial line angle (location 4) from 0.8-0.9 mm, and mid-occlusal (location 5) from 1.1-1.3 mm. Means and standard deviations for maximum load (N) for the three experimental groups are as follows: Design 1, 1773.13 (± 235.05); Design 2, 1653.08 (± 380.63); Design 3, 1256.71 (± 190.78) (Tables 6, 7). Means and standard deviations for fracture load (N) for the three experimental groups are as follows: Design 1, 1612.87 (± 469.92); Design 2, 1621.86 (± 419.62); Design 3, 1222.62 (± 191.94) (Table 6, 7). The one-way ANOVA test showed that there was a statistically significant difference for the fracture strengths of maximum (Max. Load) and fracture loads (Fx. Load) within the three designs at $\alpha=0.05$. There was a highly significant difference for maximum load ($p<0.001$) and fracture load ($p=0.02$) (Table 7). Tukey test for maximum load comparison revealed: no statistically significant difference between Design 1 and Design 2 ($p=0.553$), a highly significant difference between Design 1 and Design 3 ($p<0.001$) and a significant difference between Design 2 and Design 3 ($p=0.004$) (Table 7). Tukey test for fracture load comparison revealed: no significant difference between Design 1 and Design 2 ($p=0.998$), a significant difference between Design 1 and Design 3 ($p=0.044$) and a significant difference between Design 2 and Design 3 ($p=0.038$) (Table 7). The mean value of maximum load for Design 3 was 30% lower than mean value of maximum load for Design 1 and 24% lower when compared to Design 2.

Table 5. Coping Thickness Values for Design 1, 2, 3 (Figure 7, page 99)

Coping Thickness							
Design	**Sample**	**1**	**2**	**3**	**4**	**5**	**6**
1	1	0.4	0.6	0.7	0.9	1.2	-
1	2	0.5	0.5	0.7	0.9	1.2	-
1	3	0.5	0.6	0.7	0.9	1.2	-
1	4	0.5	0.6	0.7	0.9	1.2	-
1	5	0.4	0.6	0.7	0.9	1.1	-
1	6	0.4	0.5	0.7	0.9	1.2	-
1	7	0.4	0.6	0.7	0.9	1.1	-
1	8	0.4	0.6	0.7	0.9	1.2	-
1	9	0.5	0.6	0.7	0.9	1.2	-
1	10	0.5	0.6	0.6	0.9	1.2	-
1	11	0.5	0.6	0.7	0.9	1.2	-
1	12	0.4	0.5	0.7	0.9	1.2	-
2	1	-	0.4	0.7	0.8	1.2	-
2	2	-	0.4	0.7	0.8	1.3	-
2	3	-	0.4	0.7	0.8	1.2	-
2	4	-	0.4	0.7	0.8	1.1	-
2	5	-	0.4	0.7	0.8	1.2	-
2	6	-	0.5	0.7	0.8	1.2	-
2	7	-	0.4	0.7	0.8	1.2	-
2	8	-	0.4	0.7	0.8	1.2	-
2	9	-	0.4	0.7	0.8	1.2	-
2	10	-	0.4	0.7	0.8	1.2	-
2	11	-	0.4	0.6	0.8	1.2	-
2	12	-	0.4	0.7	0.8	1.2	-
3	1	0.4	0.5	0.7	0.9	1.1	0.4
3	2	0.4	0.5	0.6	0.9	1.1	0.4
3	3	0.5	0.6	0.7	0.9	1.1	0.4
3	4	0.5	0.6	0.7	0.9	1.1	0.4
3	5	0.4	0.5	0.7	0.9	1.2	0.5
3	6	0.4	0.5	0.7	0.8	1.2	0.5
3	7	0.5	0.6	0.7	0.9	1.1	0.4
3	8	0.5	0.6	0.7	0.9	1.1	0.4
3	9	0.5	0.5	0.7	0.9	1.2	0.4
3	10	0.4	0.5	0.7	0.9	1.2	0.4
3	11	0.4	0.5	0.7	0.9	1.1	0.4
3	12	0.4	0.5	0.7	0.9	1.1	0.4

Table 6. Maximum Load and Fracture Load Values for Design 1, 2, 3

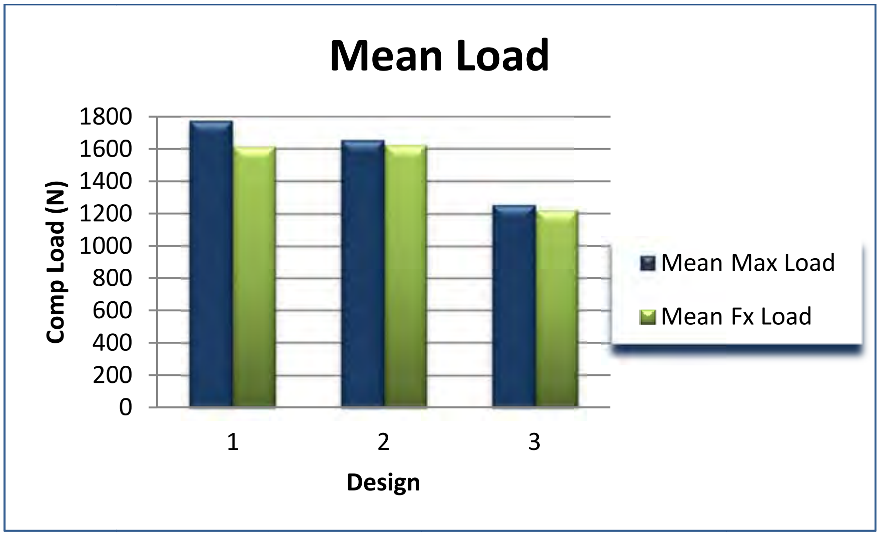
	Max Load (N)			Fx Load (N)		
Design	**1**	**2**	**3**	**1**	**2**	**3**
# of samples	12	12	12	12	12	12
Minimum	1464.98	1042.74	1011.14	667.47	884.90	991.09
Maximum	2129.32	2195.19	1706.22	2096.16	2195.19	1663.95
Mean	1773.13	1653.08	1256.71	1612.87	1621.86	1222.62
Variance	55248.10	144875.96	36394.58	220821.33	176083.28	36840.64
Std Dev	235.05	380.63	190.78	469.92	419.62	191.94

Table 7. Mean Values and Standard Deviations for Maximum Load and Fracture Load (Design 1, 2, 3)

Design	Mean Max Load (N)	Mean Fx Load (N)
1	1773.13 ± 235.05 [a]	1612.87 ± 469.92 [a]
2	1653.08 ± 380.63 [a]	1621.86 ± 419.62 [a]
3	1256.71 ± 190.78	1222.62 ± 191.94

Values with the same letter ([a]) are not significantly different at $\alpha<0.05$

Figure 11 is a graphic illustration of the mean values for maximum and fracture load in Designs 1, 2, and 3. The distribution of the maximum load and fracture load (Figure 12-17) for the individual specimens within Designs 1, 2, and 3 can be seen in Figures 12-14 and Figures 15-17 respectively.



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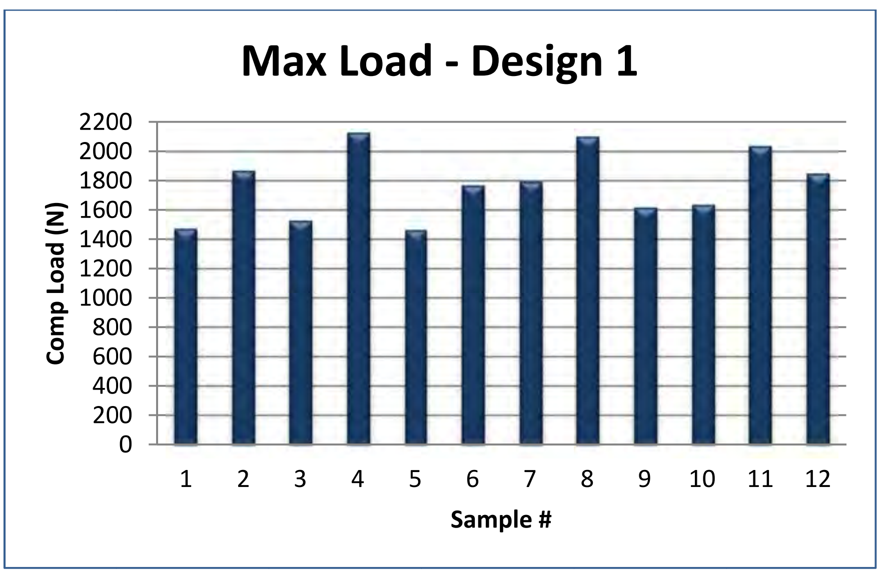
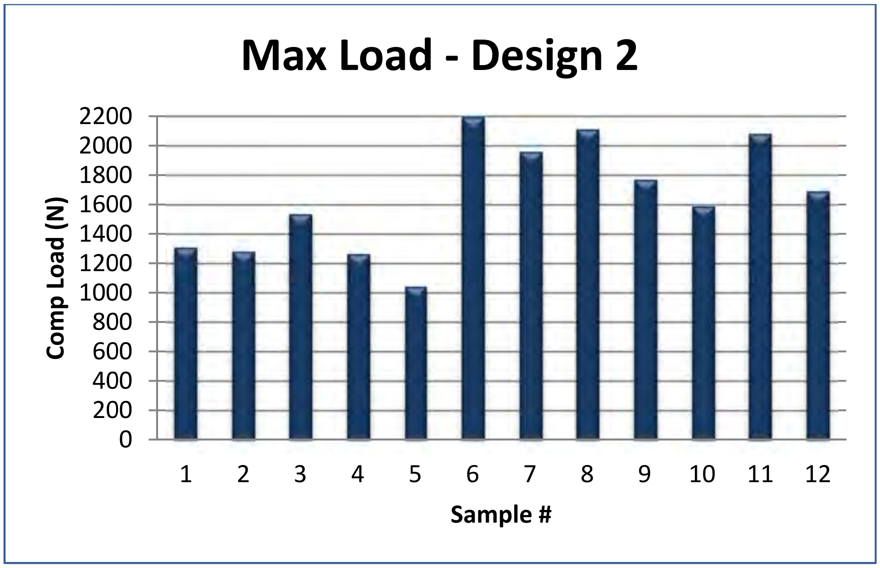
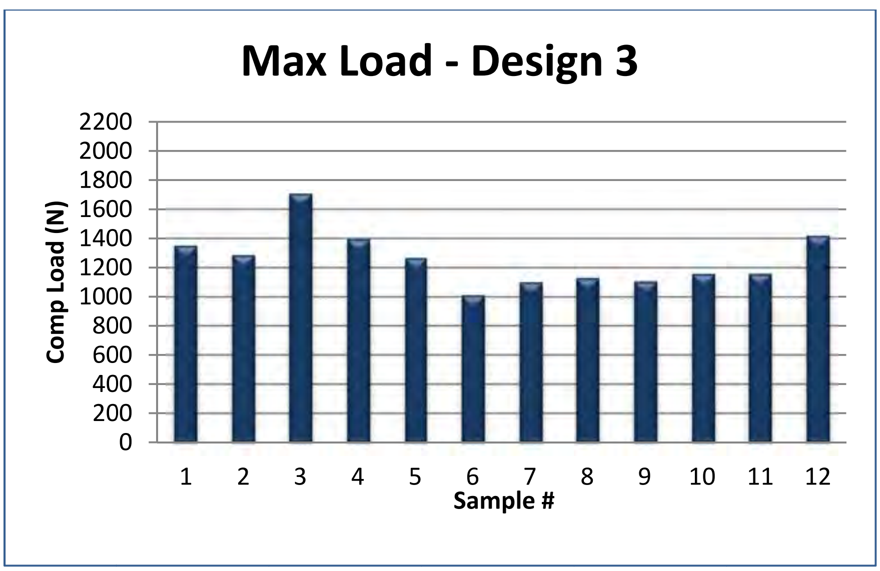


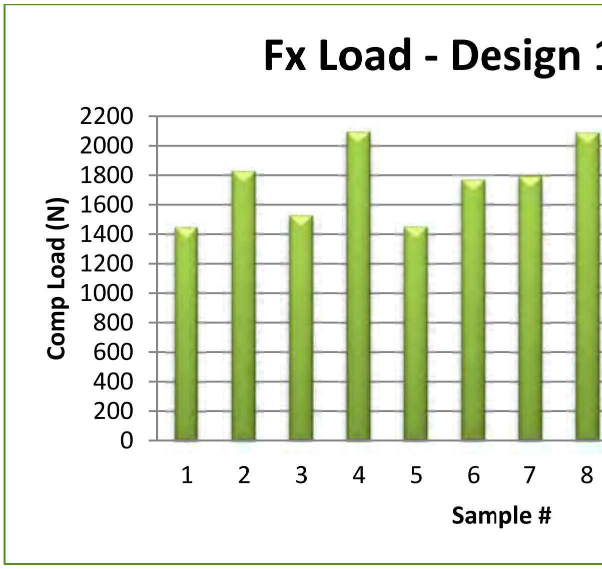
gn 1


Max Load - Design 2
Comp Load (N)
0
200
400
600
800
1000
1200
1400
1600
1800
2000
2200
1 2 3 4 5 6 7 8 9 10 11 12
Sample #

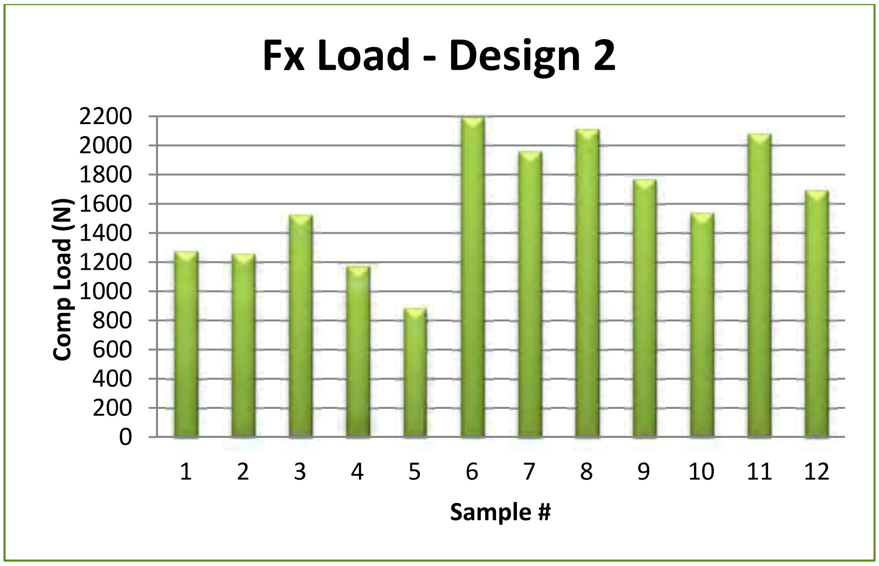
.gn 2


Max Load - Design 3
Comp Load (N)
0
200
400
600
800
1000
1200
1400
1600
1800
2000
2200
1 2 3 4 5 6 7 8 9 10 11 12
Sample #

ign 3


Fx Load - Design
Comp Load (N)
2200
2000
1800
1600
1400
1200
1000
800
600
400
200
0
1
2
3
4
5
6
7
8
Sample #

n 1


Fx Load - Design 2
Comp Load (N)
2200
2000
1800
1600
1400
1200
1000
800
600
400
200
0
1
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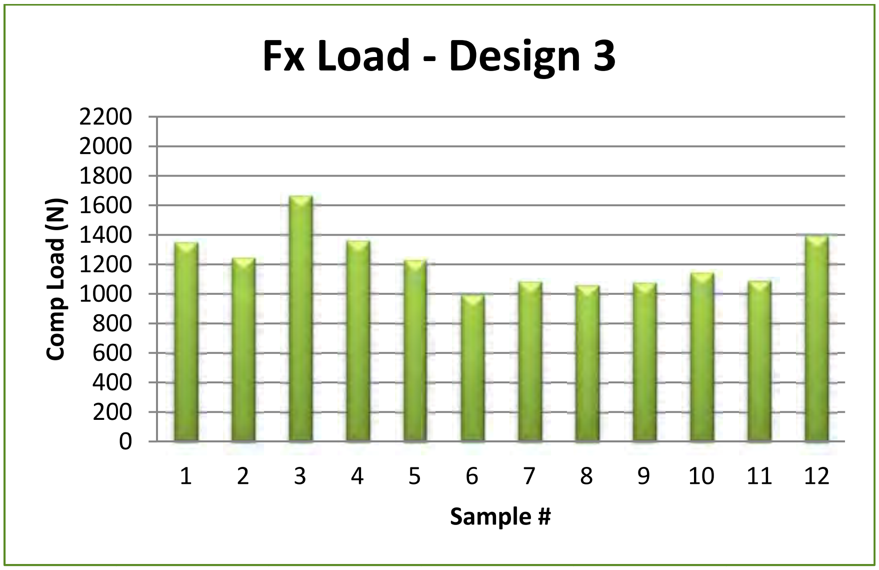
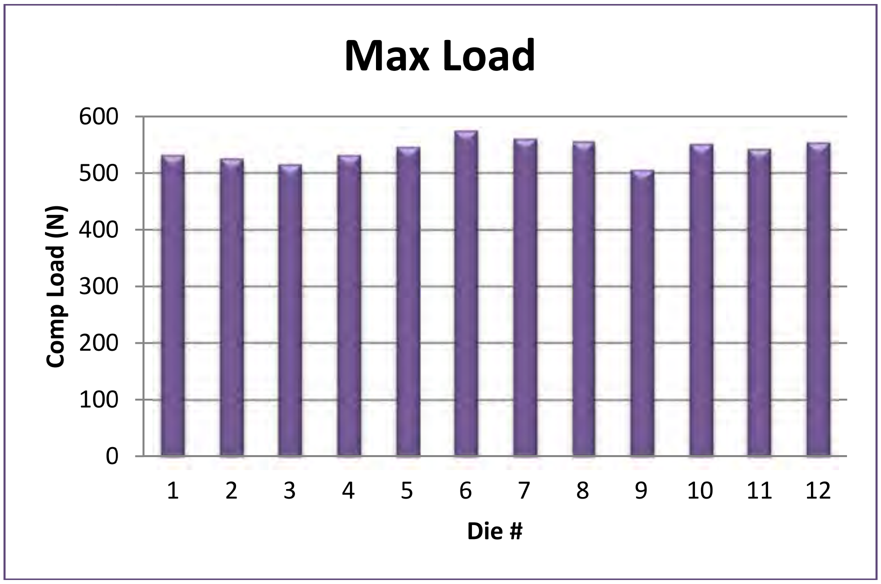
6
7
8
9
10
11
12
Sample #

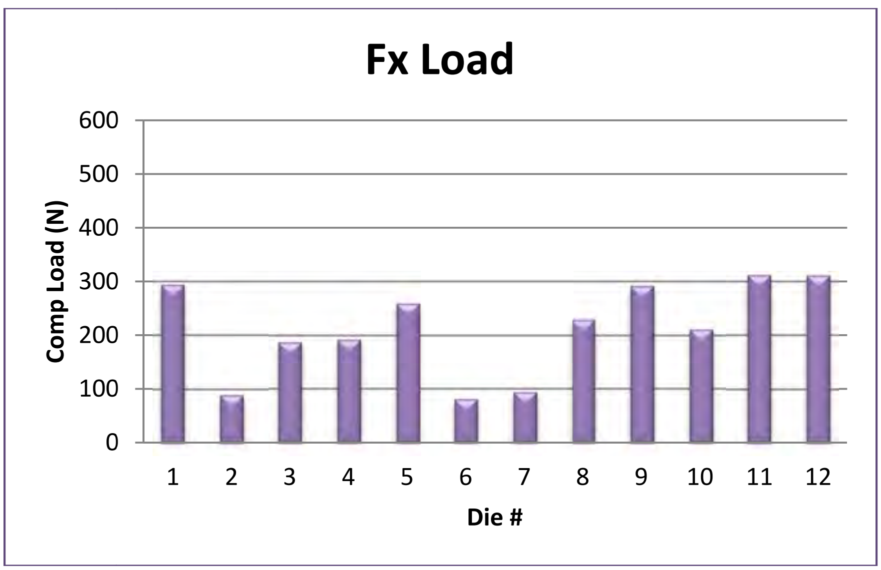
ı 2



n 3

Maximum loads of the dies are illustrated graphically in Figure 18, while fracture loads of the tested dies are demonstrated in Figure 19. The mean and standard deviation values of the maximum load and fracture load in Newtons were 541.66 ± 20.10 and 212.71 ± 86.66 respectively.


Max Load
600
500
400
300
200
100
0
Comp Load (N)
1
2
3
4
5
6
7
8
9
10
11
12
Die #


Fx Load
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500
400
300
200
100
0
Comp Load (N)
1
2
3
4
5
6
7
8
9
10
11
12
Die #

Fracture mode distribution within the porcelain, coping and die is presented in Table 8, 9, and 10 respectively. None of the samples fractured at porcelain mode (P) 0 and 1 (according to the classification in Table 4, page 104); therefore for statistical analysis fracture mode rating of porcelain 2, 3 and 4 was used.

Table 8. Fracture Mode Rating in Porcelain within Design 1, 2, 3

PORCELAIN				
Rating	**2 chip<50%**	**3 chip>50%**	**4 chip to shoulder**	**Total**
Design 1	5	6	1	12
Design 2	2	3	7	12
Design 3	3	2	7	12
Total	10	11	15	36

Due to the porcelain fragility, the fracture line in some of the samples transformed to a chip. Therefore, for statistical analysis fracture mode rating of 2 and 3 within the coping (C) were combined and called 2.

Table 9. Fracture Mode Rating in Coping within Design 1, 2, 3

COPING				
Rating	**0 no fracture**	**1 crack line**	**2 fracture line / chip**	**Total**
Design 1	12	0	0	12
Design 2	6	3	3	12
Design 3	5	6	1	12
Total	23	9	4	36

For statistical analysis the fracture mode rating of 2 within the die (D) was disregarded since none of the dies fractured vertically, therefore 0, 1 and 3 were used.

Table 10. Fracture Mode Rating in Die within Design 1, 2, 3

DIE				
Rating	**0 no fracture**	**1 horizontal**	**3 horizontal/vertical**	**Total**
Design 1	8	4	0	12
Design 2	7	3	2	12
Design 3	12	0	0	12
Total	27	7	2	36

Pearson Chi-Square tests were used to assess statistical significance between each substrate and different coping designs. A significant difference was found for porcelain fracture mode between the Designs 1 and 2 ($p=0.034$), as well as Designs 1 and 3 ($p=0.03$), whereas no significance was found between Designs 2 and 3 (Table 11). Statistical analysis of coping fracture mode within designs using Pearson Chi-Square test revealed a significant difference between Designs 1 and 2 ($p=0.018$), as well as between Designs 1 and 3 ($p=0.007$). No significance was found between Designs 2 and 3 (Table 12). The same test was used to investigate the significance of the fracture mode in the dies between designs (Table 13). Designs 1 and 2 were significantly different from Design 3 but were similar to each other. All samples in Design 1, 2, and 3 had fracture propagating from the point where the load was applied.

Table 11. Porcelain Pearson Chi-Square Test

Pearson Chi-Square - Design - Porcelain		
Comparisons	**p value**	**Significance**
Design 1: Design 2	0.034	significant
Design 1: Design 3	0.03	significant
Design 2: Design 3	0.819	nonsignificant

Table 12. Coping Pearson Chi-Square Test

Pearson Chi-Square - Design - Coping		
Comparisons	**p value**	**Significance**
Design 1: Design 2	0.018	significant
Design 1: Design 3	0.007	significant
Design 2: Design 3	0.165	nonsignificant

Table 13. Die Pearson Chi-Square Test

Pearson Chi-Square - Design - Die		
Comparisons	**p value**	**Significance**
Design 1: Design 2	0.331	nonsignificant
Design 1: Design 3	0.028	significant
Design 2: Design 3	0.043	significant

Figures 20-29 are representative of the three selected samples from each of the three designs examined under an optical microscope. Visually the largest gap between the coping and the die was seen in Design 1 and Design 3, where the thickest cement film was noticed on occlusal surfaces, while the axial wall cement film was thinner. Design 3 had the largest cement space within the occlusal surface and facial axial wall. All of the designs had a thick cement space at the cervical margin. Figures 23-26 show the marginal gap in each selected sample at 20X magnification. Figures 27-29 demonstrate the occluso-axial surfaces and coping adaptation to the cemented dies at magnification of 20X. It is clearly observed that the axio-cervical line angle had the poorest coping adaptation to the die, whereas the axial surface of the coping is well adapted and further cement space appears to be significantly thinner. The cement space on the facial axial surface of the Design 3 is more evident than on the lingual axial surface (Figures 25, 26).

Figure 20. Sectioned crown (Design 1) at 10X magnification

Figure 21. Sectioned crown (Design 2) at 10X magnification

Figure 22 Sectioned crown (Design 3) at 10X magnification

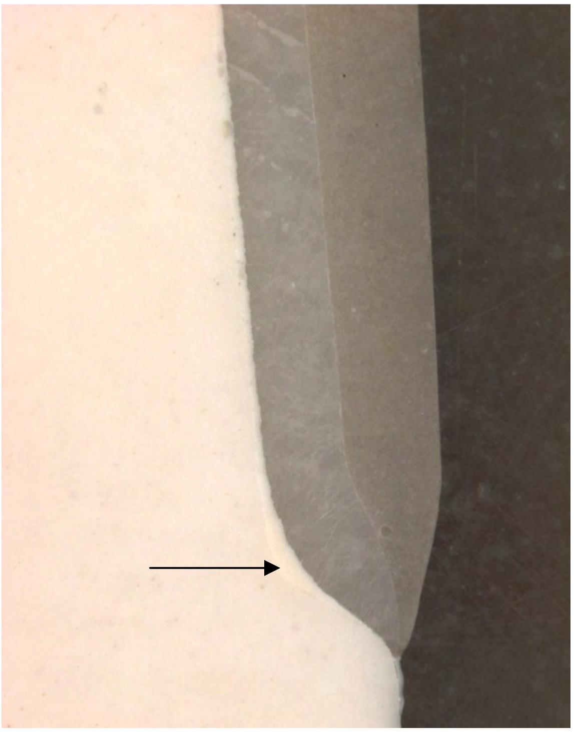


Figure 23. Sectioned crown (Design 1) at 20X magnification

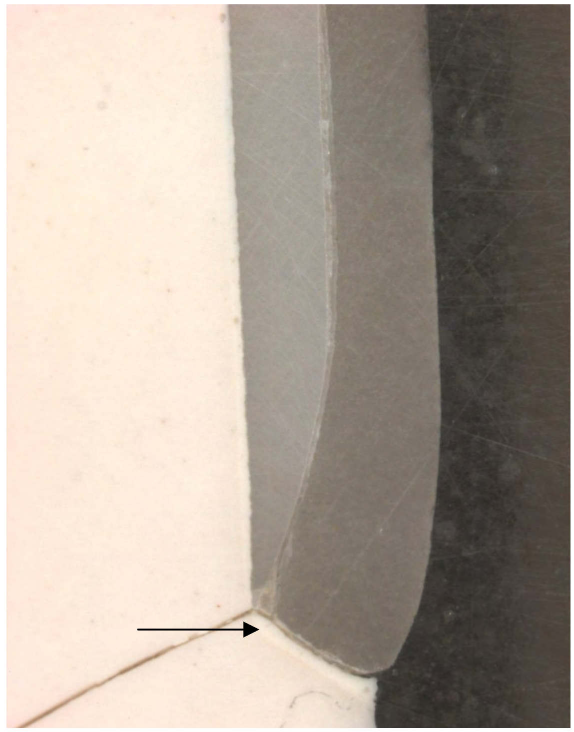


Figure 24. Sectioned crown (Design 2) at 20X magnification

Figure 25. Sectioned crown (Design 3 - facial surface) at 20X magnification

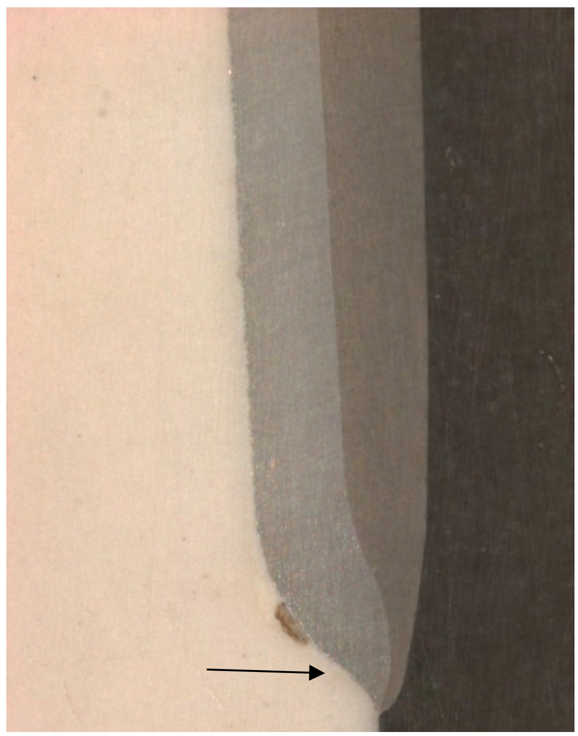


Figure 26. Sectioned crown (Design 3 - lingual surface) at 20X magnification

Figure 27. Sectioned crown (Design 1 - axio-occlusal surface) at 20X magnification

Figure 28. Sectioned crown (Design 2 - axio-occlusal surface) at 20X magnification

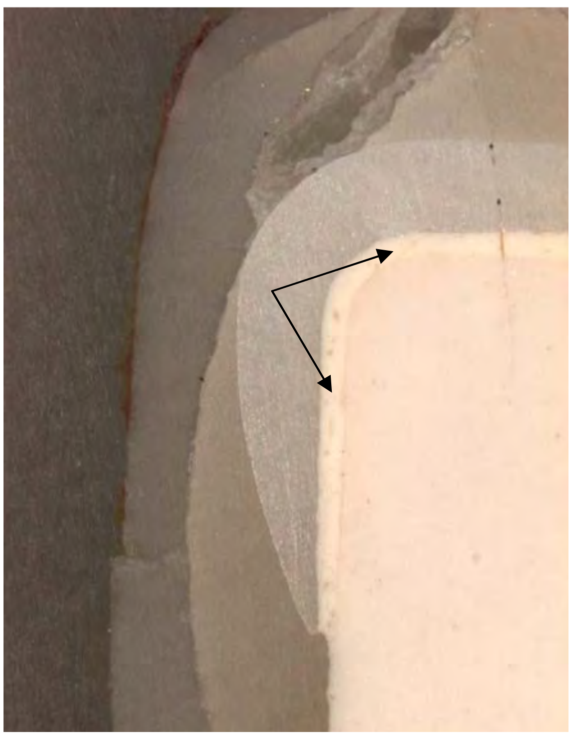


Figure 29. Sectioned crown (Design 3 - axio-occlusal surface) at 20X magnification

6. Discussion

The high flexural and fracture strengths, biocompatibility and esthetic appearance of all-ceramic zirconia core restorations potentiates their ability to replace porcelain fused to metal restorations. Nevertheless, the zirconia coping material is characterized by an opaque line at the cervical margin. In order to improve the diminished esthetic appearance and mask this inherent flaw, clinicians attempt to modify the coping by reducing its margin thickness. The effect of the coping's modification on the success or failure of the restoration has not been investigated.

The aim of the present study was to determine and compare the fracture strength and fracture mode of three different zirconia coping designs. The three investigated Procera All-Zircon coping designs were as follows: 1) coping extended to cavosurface margin; 2) coping cut back to axial wall; 3) coping cut back 3mm from the facial cervical margin. Fracture strength values were obtained by axially loading the crowns to fracture using an Instron Testing Machine.

Fracture strength test is defined as a test where samples are loaded axially, in this case at 90 degree angle, using a stainless steel rod of a set diameter and specific load rate until fracture of the specimens occurs. This type of test is relatively easy to perform and produces rapid and predictable results. Nevertheless, load to fracture testing does not mimic the intraoral environment where saliva and intermittent masticatory occlusal forces are present. Although the specimens were kept in humidity using distilled water for 24 hours, this did not reliably mimic salivary ionic content that might possess detrimental effect on the long-term success of the restoration. Aging, the mechanical property of zirconia degradation, is referred to as progressive spontaneous material transformation from the tetragonal phase to a monoclinic phase at temperatures above 200 ºC and in the presence of water or vapor. This process reduces material toughness and density through the formation of micro and macro porosities that further propagate crack formation.[38] Zirconia performance and its stability were reported in numerous studies that gave evidence of multifactorial parameters acting on the zirconia properties, such as yttria grain size, residual stress, environment moisture content, distribution of flaws as well as oxides and their concentration.[38, 39]

The normal physiologic masticatory forces within the anterior dentition are maintained at about 108 Newtons while within the posterior dentition, a greater force was observed with a range between 2-150 N.[26-29] It has also been shown that masticatory forces are highly gender dependant where females expressed a smaller occlusal load than their male counterparts.[29] When bruxism occurs, the biting forces can significantly increase up to 500-880 N.[40, 41]

From a clinical point of view, the primary role in determining long-term restoration success is ability to withstand dynamic material fatigue. In the case where material fatigue occurs, crack initiation and slow progression will lead to eventual material failure. In vitro restorative material evaluation is definitely a valid method for preliminary testing of material fracture resistance, but undoubtedly does not reliably provide a true clinical model.

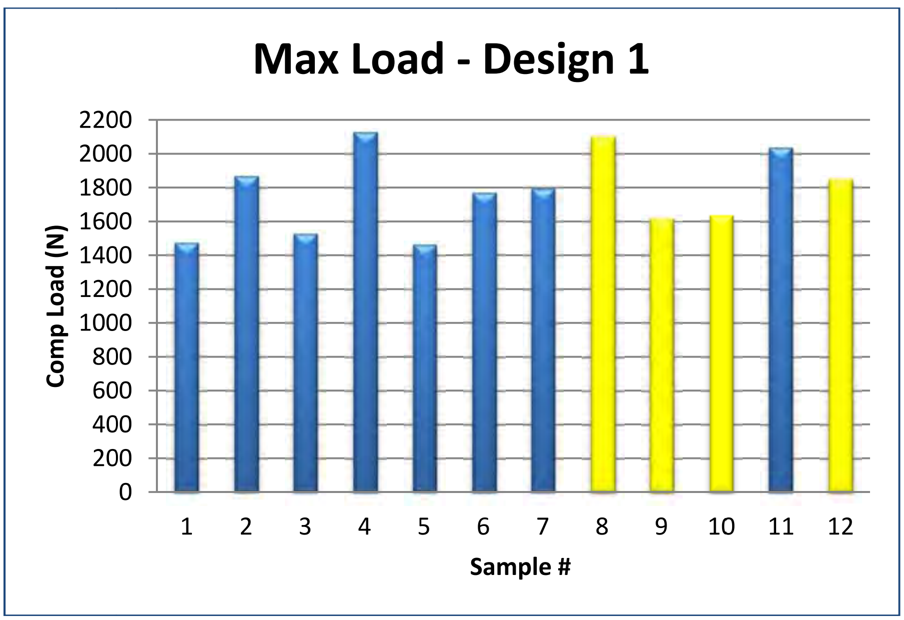
The present study does not replicate the physiological tooth mobility which has been demonstrated to possess a decisive role in the evaluation of fracture resistance. The chewing simulation tests implemented for alumina fixed restorations with and without the artificial periodontium showed the fracture resistance values of 676 N and 256 N respectively.[42]

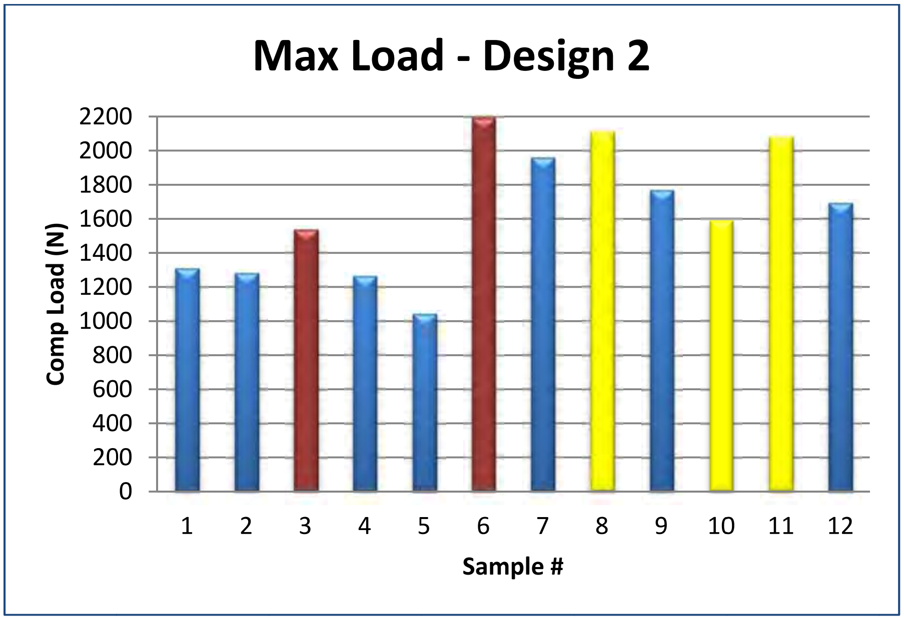
Therefore, if nearly a threefold force increase was necessary for failure to occur within a periodontally supported dentition as shown by Rosentritt et al., we could make an assumption that the fracture resistance values of 1256-1773 N observed within this study should translate to intraoral forces of over 3000 N.

Literature evidence provides abundant support to state that zirconia-based ceramics are stronger and tougher than conventional glass ceramics.[21] Since all of the tested specimens exceeded the normal physiologic as well as habitual biting force values, it can be presumed that all tested zirconia coping modifications have the potential to withstand both physiologic and pathologic occlusal forces and remain intact during load.

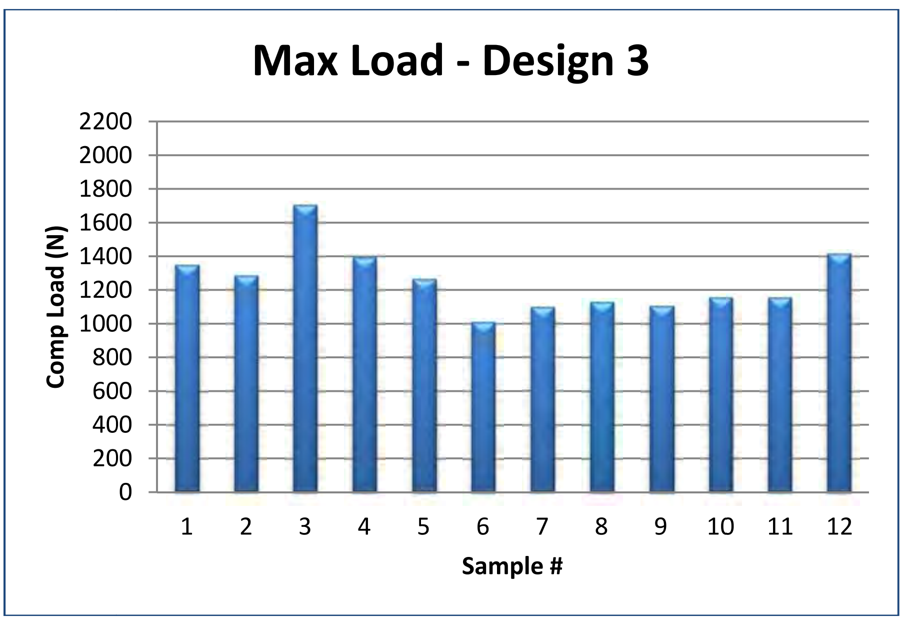
The cementing medium and surface treatment of the restorations has an influence on material strength. The methodology included etching and sandblasting with 50 µm aluminum oxide. According to Zhang et al., aluminum oxide abrasion may propagate crack formation and decrease fracture toughness.[43] Due to the methodology in this study, the inherent crack flaws may have had a detrimental and hidden impact on the test results. However, silanization and phosphate containing cement (Panavia 21/F) have been shown to significantly increase the bond strength of the restoration.[44-46] This particular luting agent has a low film thickness of 19 µm that allows it to flow into microscopic porosities, including microcracks, improving the micromechanical retention and the fracture resistance of the restoration.[47] The intimate bonding provides a “die-crown” unit effect, which reduces the chance of the stress creating failure of the restoration by allowing partial stress factors to be transferred to the die. All the dies, tested for the fracture resistance, exhibited similar strength values (541.66 ± 20.10 N) which show consistency within the epoxy specimens. Lower fracture values of the epoxy dies compared to values of the ceramic samples (D1-1773.13 ± 235.05, D2-1653.08 ± 380.63, and D3-1256.71 ± 190.78) provides evidence that it is the ceramics that is carrying most of the stress and is supporting as well as preventing the die from fracture (Figure 30-32).

The investigation of the relationship between the die fracture resistance and overall fracture values of restorations within Design 1, 2 and 3 did not demonstrate any trend (Figure 30-32). Bars presented in yellow color represent dies that fractured in Class I (horizontally). Bars that are colored in red represent dies that fractured in Class III (combination of vertical/horizontal). The remaining blue colored bars had no evidence of fracture in dies. The porcelain fractures did not seem to be related to the die fractures and loss of support as the test strength increased.




Max Load - Design 2
Comp Load (N)
2200
2000
1800
1600
1400
1200
1000
800
600
400
200
0
1 2 3 4 5 6 7 8 9 10 11 12
Sample #

d


Max Load - Design 3
Comp Load (N)
2200
2000
1800
1600
1400
1200
1000
800
600
400
200
0
1 2 3 4 5 6 7 8 9 10 11 12
Sample #

A critical aspect in fracture prevention is the precision of fit between the copings and their respective dies. The clinical success of the CAD/CAM restorations depends on their mechanical properties as well as restoration design and accuracy of the CAD/CAM process. Marginal fit of cemented prostheses in the 25 to 40 μm range has been suggested as a clinical goal.[48] Within the present study, visual inspection of selected samples under 20X magnification revealed a lack of cement space uniformity. A noticeably larger cement space was observed on the occlusal surfaces and cervical margins of all specimens. The cervical internal line angle showed decreased adaptation of the coping that could be explained by an inadequate intimate contact of the scanning probe which was larger in diameter than the axial-cervical line angle of the shoulder preparation. This results in a larger cement space and possibly in a weak area, thereby contributing to coping fracture. A second assumption could be made from the manner the coping is fabricated whereby it is designed to be 20% larger than the final coping to accommodate the shrinkage during the sintering process. Also, coping without shoulder support may seat deeper on the preparation and adapt more accurately to occlusal and axial walls of the preparation. Lastly, the potentially compromised fit of the coping could also occur due to die variations between the scanned dies and those being used for cementation. The two types of dies used should be identical; nevertheless, microscopic variations may occur and influence the coping to die adaptation. The selected specimen within Design 2 had a thicker cement space at the marginal gap which could be explained by the condensation technique of the veneering porcelain, and/or additional shrinkage of the unsupported porcelain during the firing process even though the peg-putty was used with the intent to prevent such marginal discrepancy. A larger cement space was also found on the facial wall of the coping which was cut back 3 mm from the facial margin (Design 3). This again could be an inherent flaw of the fabrication and firing process compared to fabrication and firing of the other 2 coping designs. The increased cement space could be detrimental to the specimen’s fracture resistance due to the presence of a larger gap between the restorations and dies. Additionally, it is possible that the substantial area of overlaid porcelain that remained unsupported in Design 3 could have a detrimental effect on the fracture values observed. This was previously described by Van Der Zel et al., where unsupported crowns with shoulder free zirconia copings had 22% lower breaking strength than those with supported margins.[49] An interesting speculation was made by Tuntiprawon et al. who stated that the greater the cement space the less force is needed to fracture the ceramic. [50] Therefore, if we could control the processing, scanning and cementation errors, we would in consequence improve fracture resistance of all 3 coping designs.

The zirconia material has an inherent strength but in the fabrication process of the desired restoration its strength may increase or decrease resulting from improper handling techniques. The process begins when the coping is designed to produce proper fit, then it may be altered by sintering, aluminum oxide abrasion, the number and temperature of firings, finishing techniques, bonding agents used and lastly by the constant onslaught of saliva and mastication. Other factors that could affect fracture resistance could be a mismatching of coefficient of thermal expansion between the coping and veneering porcelain. In this study both materials had similar coefficients of thermal expansion (coping had slightly higher expansion than veneering porcelain) therefore coefficient of thermal expansion discrepancy did not play a crucial role. Comparison of the individual specimens revealed significant variation in fracture strengths (N); 1464-2129 for Design 1, 1042-2195 for Design 2, and 1011-1706 for Design 3. The wide range in failure values within each design could be a result of different number of firings required to accommodate the margin veneering porcelain shrinkage and to build up the shoulder porcelain not supported by the coping as well as operator inexperience. These two factors may have inadvertently introduced flaws into the veneering ceramic that would remain hidden and undetected.

The results of the fracture tests on three coping designs clearly present evidence that the coping modification within the shoulder of the preparation (Design 2) had approximately 7 % lower strength than the original coping model (Design 1). However, the stress resistance of the restoration with the coping cut back 3 mm on the facial surface (Design 3) was significantly reduced by 30 % compared to the original design (Design 1).

It is controversial whether the variable of anatomical morphology or constant coping thickness should be considered. However, Fisher's study demonstrated that fracture strength increased by 30% in specimens with an anatomical design of the framework.[32] The presented research used a concave occlusal surface on both coping and veneering porcelain. Therefore, it can be speculated that the fracture strengths could have been lower with constant coping thicknesses. Wakabayashi, on the other hand, related the core/veneer thickness ratio of the specimens to crack formation in bilayered ceramics and determined that crack initiation starts more readily within a thin veneer. The study also showed that as the veneer thickness increases, concomitantly increasing the veneer to the core ratio, the crack initiation changes its origin from the veneer to core.[51] The standard coping thickness recommended for the Procera All-Zircon system is 0.6 mm. The present study was designed to standardize the absolute restoration thickness while maximizing final fracture results by maintaining coping thickness uniformity. The measurements of each individual coping at different surfaces were consistent and ranged from 1.1-1.3 mm on occlusal surfaces, 0.6-0.7 mm on axial surfaces and 0.4-0.5 mm at the margin. With the veneering porcelain in place, a total restoration thickness of 1.5 mm was maintained for the axial wall and 2.0 mm for the occlusal surface, thereby the core/porcelain ratio was approximately 1:2.3 for axial and 1:1.7 for occlusal surfaces. Contradicting Wakabayashi's data, fracture load values did not seem to be affected by the coping margin thickness having a low core/veneer ratio. This study also did not provide any evidence of the difference in the crown strength that would result from a minimal change of coping thickness that was again suggested by Wakabayashi. Although the research data shows that there was not any difference between the frequency of material fracture at the axial (1:1.7) versus occlusal (1:2.3) position, a general trend was observed whereby the fractures most frequently occurred at the junction of the two. Possibly, there may be a detrimental effect of a sharp core/veneer ratio shift or the manner stress is distributed at the veneering edge. Additionally, a frequent porcelain chip extending to the shoulder was observed in Design 2 and Design 3 where the coping margin thickness was thinned out, increasing the core/porcelain ratio.

Ceramics have been a material of choice advocated to fulfill the principles of esthetics and where a natural look is desired. However, clinicians have been struggling with white pigment and an opaque appearance at the margin area of the zirconia supported restorations. As the light enters the zirconia supported ceramic restoration, part of it is reflected, while the remaining light penetrating enamel is scattered and possibly gives an effect of shadowing the tooth structure adjacent to the restoration. The light reflection is commonly affected by ceramic thickness and the reduction of the coping at the cervical margin improves esthetics but leaves the veneering porcelain with a surface area that is not supported by the coping.[15, 52] As a result, the veneering porcelain has a lower load bearing capacity and greater probability of fracture. Visual inspection of the tested specimens revealed that in cases where the coping was modified, the veneering porcelain usually fractured to the cervical margin in addition to the occluso-axial load bearing point.

In addition to discussing fracture resistance values, it is important to analyze the fracture modes in each experimental group. Fractures observed at the interface between the ceramic core material and the veneering porcelain, have been commonly related to the stress that is introduced by differences in elastic modulus between the core and veneering porcelain, prosthesis geometry and size and location of flaws in the material. [53] The present study, where the load was applied 2 mm from the external edge of the restoration, closely reproduced clinical function where the occlusal load is applied at certain shearing location onto the tooth structure. The effect of the forces during the fracture test exceeded the mechanical capacity of the veneering porcelain, resulting in catastrophic failure. For the purpose of standardization and clear visual explanation of the fracture location a new classification was developed. In all groups, fractures appeared first on the restoration surface prior to the die fracture, which was affirmed by Burke.[54] Microscopic examination performed in one of the studies found the origin of failure at the interface between ceramic coping and luting agent.[53] Another common finding in the literature is delamination. Nevertheless, in this study, only one specimen from Design 3 had a core delaminated from the die and cement and that could be attributed to a technique flaw or improper cement flow. Another factor, previously mentioned in the discussion that influences the failure location is the core/porcelain ratio. Cracks initiated within the porcelain surface approached the core but were arrested at the interface between the veneer and the core. These results were in agreement with the study done by Wakabayashi. [51] All of the specimens demonstrated chipped porcelain in the area where load was applied and in cases where the veneering porcelain was not supported by coping, the fracture occurred additionally at the cervical area. Design 2 (p=0.034) and Design 3 (p=0.03) had significantly higher rating values of the porcelain fractures than Design 1. The porcelain delamination has been commonly attributed to the elastic modulus and fracture toughness mismatch between core and porcelain.[55] From the clinical perspective, the fracture of the coping, either crack, chip or fracture line, catastrophically affects the success of the restoration, yet the suggested fracture mode categories help determine the severity of the fracture within the different coping designs. Specimens in Design 2 (p=0.018) and 3 (p=0.007) had significantly higher rating values of coping fracture locations than samples in Design 1.

It is up to the practitioner to become familiar with the correct protocol during all procedures for the fabrication of ceramic crowns, beginning from the treatment plan, preparation and restoration design, and finishing with the cementation process. Failure to do so can significantly diminish the success of the restoration.

The study was designed to closely reproduce the clinical and laboratory situation, knowing that in-vitro experimental conditions do not accurately replicate those found in the oral cavity (in-vivo). The results of this study, where the coping modification improves the esthetics by diminishing the margin opaque line without compromising the strength, introduce a promising outcome. Nevertheless, clinical trials need to be implemented to determine the effects of cyclic loading, saliva and function on the proposed coping design modifications.

7. Conclusions

Within the limitations of this study, the following conclusions can be drawn:

1. Cutting back the coping shoulder to the axial wall did not significantly decrease the fracture resistance of zirconia ceramic crowns.
2. Cutting back the coping shoulder 3mm from the cervical margin significantly decreased the fracture resistance of zirconia ceramic crowns.
3. The fracture mode of veneering porcelain was more catastrophic in the modified zirconia coping restorations.
4. The fracture mode of the zirconia coping was more catastrophic in the modified zirconia coping restorations.
5. All three tested zirconia coping modifications have the potential to withstand physiological occlusal forces.

8. Clinical Significance

The finding of this in vitro study allows clinicians to select with confidence the appropriate all-ceramic zirconia restoration design that would offer superior esthetics without compromising the strength.

9. References

1. Walton TR. An up to 15-year longitudinal study of 515 metal-ceramic FPDs: Part 2. Modes of failure and influence of various clinical characteristics. Int J Prosthodont 2003;16(2):177-82.

2. Holm C, Tidehag P, Tillberg A, Molin M. Longevity and quality of FPDs: a retrospective study of restorations 30, 20, and 10 years after insertion. Int J Prosthodont 2003;16(3):283-9.

3. Mumford G. The Porcelain Fused to Metal Restoration. Dent Clin North Am 1965;23:241-9.

4. Kuwata M. Gingival margin design of abutments for ceramo-metal restorations. Quintesence of Dental Technology 1979.

5. Gavelis JR, Morency JD, Riley ED, Sozio RB. The effect of various finish line preparations on the marginal seal and occlusal seat of full crown preparations. J Prosthet Dent 1981;45(2):138-45.

6. Christensen GJ. Ceramic vs. porcelain-fused-to-metal crowns: give your patients a choice. J Am Dent Assoc 1994;125(3):311-2, 14.

7. Hobo S, Shillingburg HT, Jr. Porcelain fused to metal: tooth preparation and coping design. J Prosthet Dent 1973;30(1):28-36.

8. Goodacre CJ, Van Roekel NB, Dykema RW, Ullmann RB. The collarless metal-ceramic crown. J Prosthet Dent 1977;38(6):615-22.

9. O'Boyle KH, Norling BK, Cagna DR, Phoenix RD. An investigation of new metal framework design for metal ceramic restorations. J Prosthet Dent 1997;78(3):295-301.

10. Lehner CR, Mannchen R, Scharer P. Variable reduced metal support for collarless metal ceramic crowns: a new model for strength evaluation. Int J Prosthodont 1995;8(4):337-45.

11. Ahnlide I, Ahlgren C, Bjorkner B, Bruze M, Lundh T, Moller H, et al. Gold concentration in blood in relation to the number of gold restorations and contact allergy to gold. Acta Odontol Scand 2002;60(5):301-5.

12. Christensen GJ. Choosing an all-ceramic restorative material: Porcelain-fused-to-metal or zirconia-based? J Am Dent Assoc 2007;138(5):662-5.

13. Raigrodski AJ. All-ceramic full-coverage restorations: concepts and guidelines for material selection. Pract Proced Aesthet Dent 2005;17(4):249-56; quiz 58.

14. Holand W, Rheinberger V, Apel E, van 't Hoen C, Holand M, Dommann A, et al. Clinical applications of glass-ceramics in dentistry. J Mater Sci Mater Med 2006;17(11):1037-42.

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

16. Heffernan MJ, Aquilino SA, Diaz-Arnold AM, Haselton DR, Stanford CM, Vargas MA. Relative translucency of six all-ceramic systems. Part II: core and veneer materials. J Prosthet Dent 2002;88(1):10-5.

17. Magne P, Belser U. Esthetic improvements and in vitro testing of In-Ceram Alumina and Spinell ceramic. International Journal of Prosthodontics 1997;10(5):459-66.

18. Hauptmann H, Suttor D, Hoescheler S, S F. Material properties of all-ceramic zirconia prosthesis. ESPE Dental AG, 82229 Seefeld, Germany 2000.

19. Takagi H, Nishioka K, Kawanami T, al e. The properties of a closely sintered zirconia. Ceram Forum Int 1985;62:195-98.

20. Quinn JB, Sundar V, Lloyd IK. Influence of microstructure and chemistry on the fracture toughness of dental ceramics. Dent Mater 2003;19(7):603-11.

21. Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of all-ceramic materials. Part II. Zirconia-based dental ceramics. Dent Mater 2004;20(5):449-56.

22. Hogg KD. Load to fracture of different all- ceramic croown systems. IADR General Session 2004.

23. Christel P, Meunier A, Heller M, Torre JP, Peille CN. Mechanical properties and short-term in-vivo evaluation of yttrium-oxide-partially-stabilized zirconia. J Biomed Mater Res 1989;23(1):45-61.

24. 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(3):231-8.

25. Tinschert J, Zwez D, Marx R, Anusavice KJ. Structural reliability of alumina-, feldspar-, leucite-, mica- and zirconia-based ceramics. J Dent 2000;28(7):529-35.

26. Bates JF, Stafford GD, Harrison A. Masticatory function - a review of the literature. III. Masticatory performance and efficiency. J Oral Rehabil 1976;3(1):57-67.

27. Gibbs CH, Mahan PE, Mauderli A, Lundeen HC, Walsh EK. Limits of human bite strength. J Prosthet Dent 1986;56(2):226-9.

28. Kiliaridis S, Kjellberg H, Wenneberg B, Engstrom C. The relationship between maximal bite force, bite force endurance, and facial morphology during growth. A cross-sectional study. Acta Odontol Scand 1993;51(5):323-31.

29. Helkimo E, Carlsson GE, Helkimo M. Bite force and state of dentition. Acta Odontol Scand 1977;35(6):297-303.

30. Behrens A. Fracture strength of colored zirconia copings with reduced wall thickness. IADR General Session 2004.

31. Potiket N, Chiche G, Finger IM. In vitro fracture strength of teeth restored with different all-ceramic crown systems. Journal of Prosthetic Dentistry 2004;92(5):491-5.

32. Fischer J. Strength of zirconia single crowns related to coping design. IADR General Session 2005.

33. Al-Reyahi M. The precision of fit of the procera all-ceramic coping of 0.4 mm thickness. thesis University of Michigan, 2003.

34. Reich S, Reusch B. Fracture force of ZrO2 copings dependent on preparation and thickness. IADR General Session 2003.

35. Sano H, Ciucchi B, Matthews WG, Pashley DH. Tensile properties of mineralized and demineralized human and bovine dentin. J Dent Res 1994;73(6):1205-11.

36. Neiva G, Yaman P, Dennison JB, Razzoog ME, Lang BR. Resistance to fracture of three all-ceramic systems. J Esthet Dent 1998;10(2):60-6.

37. Philip GK, Brukl CE. Compressive strengths of conventional, twin foil, and all-ceramic crowns. J Prosthet Dent 1984;52(2):215-20.

38. Shimizu K, Oka M, Kumar P, Kotoura Y, Yamamuro T, Makinouchi K, et al. Time-dependent changes in the mechanical properties of zirconia ceramic. J Biomed Mater Res 1993;27(6):729-34.

39. Burger W, Richter HG, Piconi C, Vatteroni R, Cittadini A, Boccalari M. New Y-TZP powders for medical grade zirconia. J Mater Sci Mater Med 1997;8(2):113-8.

40. Kelly JR. Ceramics in Restorative and Prosthetic Dentistry. Annu Rev Mater Sci 1997;27:443-68.

41. Kelly JR. Clinically relevant approach to failure testing of all-ceramic restorations. Journal of Prosthetic Dentistry 1999;81(6):652-61.

42. Rosentritt M, Plein T, Kolbeck C, Behr M, Handel G. In vitro fracture force and marginal adaptation of ceramic crowns fixed on natural and artificial teeth. Int J Prosthodont 2000;13(5):387-91.

43. Zhang ZS, Lawn B, Malament K, Thompson V, Rekow D. Damage accumulation and fatigue life of particle - abraded ceramics. Int J Prosthodont 2006;19(4):442-48.

44. Martin J, Sada A, Castellon P, Burgess JO, Blatz MB. In vitro comparative shear bond strength to Procera All Zircon. IADR General Session 2003.

45. Luthy H, Loeffel O, Hammerle CH. Effect of thermocycling on bond strength of luting cements to zirconia ceramic. Dent Mater 2006;22(2):195-200.

46. Escribano N, de la Macorra JC. Microtensile bond strength of self-adhesive luting cements to ceramic. J Adhes Dent 2006;8(5):337-41.

47. Juntavee N, Millstein PL. Effect of surface roughness and cement space on crown retention. J Prosthet Dent 1992;68(3):482-6.

48. American Dental Assocation: ANSI/ADA Specification No. 8 for zinc phosphate cement. Guide to dental materials and devices. 5th ed. Chicago: American Dental Association 1970:71.

49. Van Der Zel JM, Grinwis T, De Kler M, Tsadok Hai T. Effect of shoulder design on failure lodad of PTCercon crowns. IADR General Session 2004.

50. Tuntiprawon M, Wilson PR. The effect of cement thickness on the fracture strength of all-ceramic crowns. Aust Dent J 1995;40(1):17-21.

51. Wakabayashi N, Anusavice KJ. Crack initiation modes in bilayered alumina/porcelain disks as a function of core/veneer thickness ratio and supporting substrate stiffness. J Dent Res 2000;79(6):1398-404.

52. McLean JW. New dental ceramics and esthetics. J Esthet Dent 1995;7(4):141-9.

53. Pallis K, Griggs JA, Woody RD, Guillen GE, Miller AW. Fracture resistance of three all-ceramic restorative systems for posterior applications. J Prosthet Dent 2004;91(6):561-9.

54. Burke FJ. The effect of variations in bonding procedure on fracture resistance of dentin-bonded all-ceramic crowns. Quintessence Int 1995;26(4):293-300.

55. Guazzato M, Proos K, Quach L, Swain MV. Strength, reliability and mode of fracture of bilayered porcelain/zirconia (Y-TZP) dental ceramics. Biomaterials 2004;25(20):5045-52.

10. Appendix

Experimental Flow Chart

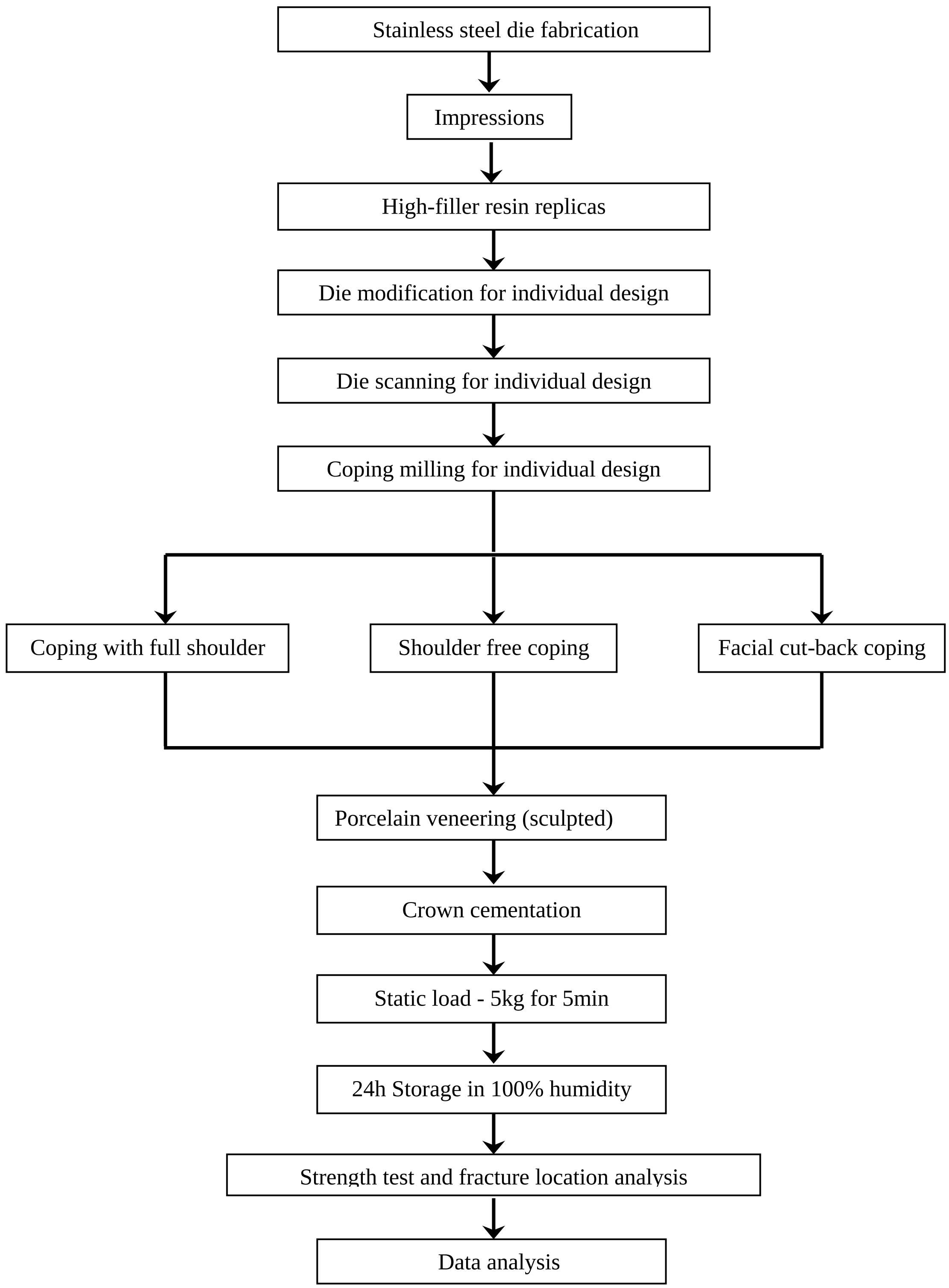


Table 14. Fracture Mode (FM) of Porcelain (P), Coping (C) and Die (D) for Design 1, 2, 3 (Table 4, page 104)

Fracture Mode				
Design	**Sample**	**P**	**C**	**D**
1	1	3	0	0
1	2	3	0	0
1	3	2	0	0
1	4	3	0	0
1	5	2	0	0
1	6	2	0	0
1	7	3	0	0
1	8	3	0	1
1	9	2	0	1
1	10	2	0	1
1	11	3	0	0
1	12	4	0	1
2	1	2	0	0
2	2	2	0	0
2	3	4	2	3
2	4	3	1	0
2	5	4	2	0
2	6	4	2	3
2	7	4	1	0
2	8	4	0	1
2	9	3	1	0
2	10	4	0	1
2	11	4	0	1
2	12	3	0	0
3	1	2	0	0
3	2	2	0	0
3	3	4	1	0
3	4	4	0	0
3	5	4	3	0
3	6	2	1	0
3	7	4	0	0
3	8	3	1	0
3	9	3	1	0
3	10	4	0	0
3	11	4	1	0
3	12	4	1	0