EFFECT OF COPING DESIGN ON THE FRACTURE RESISTANCE OF PRESSABLE ZIRCONIA CORE CERAMICS

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DEDICATION

This thesis and my accomplishments over the past three years of graduate training are dedicated to my family:

My adorable children Anna and Alex, your sweet faces and unconditional love inspire me to be a better person every day.

My husband Shree (my Love), you always believe in me. Your constant love and support have enabled me to successfully achieve all my goals.

My dear mother, you instilled in me the drive to keep learning, praying and rising above every challenge.

My brothers and sister-in-laws, I am truly blessed to have such a wonderful family. You are my support structure and you continue to guide my success.

My father-in-law and mother-in-law, I deeply appreciate your generosity, thoughtful words and sound advice.

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CHAPTER I

BACKGROUND AND SIGNIFICANCE

The current patient population has more access to information regarding dental products via the internet, news and entertainment magazines and television. This is further impacted by exposure to intense and uncensored advertising among dental material competitors. The result has been an increase in the demand for superior restorations and often unattainable esthetic dental expectations. Patients are aware that ceramic materials are best able to mimic the appearance of natural teeth and over the past several decades this has driven vast improvements in ceramic dentistry. However, improvements in the esthetics of dental restorations cannot compromise their strength to withstand maximum bite forces which are around 1031 N for the partially dentate and 1243 N in patients with full a dentition.[1]

Metal ceramic restorations have been used in dentistry for half a century. Improvements in alloys, substrates and veneering porcelain during this period have contributed to its widespread acceptance. The conventional porcelain fused to metal (PFM) restoration undoubtedly remains a popular treatment option as dentists are more confident with the clinical preparation, fabrication technique, high strength and accuracy. However, metal-ceramic materials are appropriate for less visible areas of the oral cavity because they lack excellent esthetics. The metal substructure of PFM restorations contributes to its opacity and when directly visible, the metal is dark and unsightly. Furthermore, for adequate strength and rigidity, a thickness of 0.3-0.5 mm of metal of metal substructure is required.[2] The esthetic deficit of traditional PFM restorations was addressed by utilization of a shoulder preparation. This design accommodates full porcelain coverage of the metal substructure with no metal collar at the margin. In order to achieve full porcelain coverage, the metal coping design can vary from extension to the shoulder margin or retraction just short of the margin. In the later circumstance, the metal is replaced with a porcelain shoulder margin. Despite these improvements, the unnatural opacity of metal-ceramic restorations remains unaltered.

Improvements in translucency and color of indirect restorations were maximized by the development of all-ceramic systems. This esthetic improvement is attributable to the natural translucent and reflection effects of ceramic materials.[3-4] Wide varieties of ceramic systems have been developed to match the growing application of ceramic restorations. This has lead to an extensive and overlapping classification of porcelain according to its composition, application, manufacture, and processing as well as substructure material.

The choice of all-ceramic restoration is not only determined by esthetics but also the shape of the crown preparation, occlusal and axial clearance, margin thickness and occlusal load. Initial concerns regarding the strength of all-ceramic restorations were alleviated with the advent of high strength core ceramics. Researchers and manufacturers have developed advanced formulas to prevent crack propagation mainly by using yttrium-tetragonal zirconia polycrystals (Y-TZP), commonly known as zirconia.[5-6] The introduction of zirconia-based ceramics as a restorative dental material has generated considerable interest in the dental community. This has been matched with extensive industrial, clinical and research activity.[7] However core materials have been found to generally lack fluorescence.[8] Ceramics with high strength tend to be more opaque and pose a challenge when trying to match natural tooth color, but they can mask discoloration when present. For this reason, knowledge of the optical properties of available ceramic systems enable the clinician to make appropriate choices when faced with various esthetic challenges.[9]

Strength, adaptation and esthetics of ceramic restorations are determined by the design, crystalline structure and dimensions of the coping. No significant difference in fracture strength has been found in 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.[10] However, zirconia-based ceramics have displayed more structural reliability and superior mechanical properties.[11]

The manufacturer's recommended thickness of zirconium copings which support veneering porcelain ranges from 0.5-0.8 mm with the coping shoulder covering the margins of the tooth preparation. However, this design contributes to an opaque and unnatural appearance at the cervical third of the restoration. Unlike metal-ceramic restorations, alterations in the core design of all-ceramic restorations for esthetic improvement have not been widely investigated. Clinicians should be able to select the appropriate all-ceramic zirconia core design which offers superior esthetics without compromising the strength of the restoration.

Developments in ceramic technology have vastly broadened its application in dentistry. The rapid diversification in equipment and materials available for fabrication of computer-aided design/computer-aided manufacturing (CAD-CAM) prostheses along with an increase in the availability of dental laboratory processed CAD-CAM restorations is driving the use of polycrystalline zirconia copings and framework materials. The relatively high stiffness and good mechanical reliability of partially stabilized zirconia allows for thinner core layers, longer bridge spans, and the use of all-ceramic fixed partial dentures (FPDs) in posterior locations. Studies to determine the extent to which cores and frameworks can be modified are ongoing.

More clinically relevant specimen geometry, surface finish, and mechanical loading are being applied to in vitro studies. Therefore, in vitro studies are becoming more reliable indicators of the clinical performance of ceramic prostheses. Regardless of these improvements, clinicians should exercise caution when extrapolating from the laboratory data to clinical cases.[12] Clinical failures can be simulated by blunt contact loading, cyclic fatigue loading, and loading in an aqueous environment.[13] Regardless of these efforts, physical testing does not guarantee a clinically relevant mode of failure. Specimens should be prepared and loaded using a clinically applicable method that reproduces clinical modes of failure. Experiments with novel loading geometry have been shown to reproduce clinically similar failure modes in the laboratory.[14] The relevance of physical tests which do not fracture in this manner is uncertain. Clinicians should also be able to discern whether data obtained from in vitro tests has any clinical relevance.

PURPOSE AND HYPOTHESES

PURPOSE

The aim of this in vitro study was to determine the effect of a modified coping margin design on the fracture resistance and location of heat pressed Procera® All-Zircon crowns. It was also to determine whether physical tests on regular geometrically designed restorations can reliably predict the behavior of anatomical restorations with irregular occlusal surfaces.

Two basic core designs on two different die forms were evaluated. The first die was intended to mimic the topography of an anatomical premolar preparation whereas the second die was a tapered cylinder with a flat occlusal surface. The main features of the four groups evaluated by this research are listed:

Anatomical crown with cut-back core design (Group 1)

Zirconia coping extended to the complete shoulder margin on the lingual and 3 mm short of the crown margin on the facial fabricated on an anatomical die with an irregular occlusal surface.

Flat crown with cut-back core design (Group 2)

Zirconia coping extended to the complete shoulder margin on the lingual and 3 mm short of the crown margin on the facial fabricated on a die with a flat occlusal surface.

Anatomical crown with normal core design (Group 3)

Zirconia coping designed to extend to the entire shoulder margin of the preparation fabricated on an anatomical die with an irregular occlusal surface.

Flat crown with normal core design (Group 4)

Zirconia coping designed to extend to the entire shoulder margin of the preparation fabricated on a die with a flat occlusal surface.

HYPOTHESES

Primary Hypothesis

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

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

Secondary Hypothesis

Ho: The coping design modification is not associated with fracture location of heat pressed Zirconia core ceramic crowns.

Ha: The coping design modification is associated with fracture location of heat pressed Zirconia core ceramic crowns.

Tertiary Hypothesis:

Ho: Physical tests on anatomical die preparations do not yield comparable results with regular geometric die preparations with flat occlusal surfaces.

Ha: Physical tests on anatomical die preparations yield comparable results with regular geometric die preparations with flat occlusal surfaces.

SPECIFIC AIMS

To evaluate the effect of a modified coping design on the fracture strength of heat pressed ceramic restorations.

To evaluate the effect of a modified coping design on the fracture location of heat pressed ceramic restorations.

To evaluate whether tests on regular geometrically shaped crowns with flat occlusal surfaces yield comparable results with tests on anatomical crowns.

REVIEW OF THE LITERATURE

HISTORY OF CERAMICS

The word 'ceramic' is derived from the Greek word keramos, meaning made of clay. In human evolution, the controlled use of fire was a breakthrough adaptation which provided heat and light and later allowed the physical properties of materials to be manipulated for the production of ceramics and metals.[15] Consequently, ceramics were among the first materials to be artificially made by humans after primitive man became aware of the plastic properties of mud and clay.[16] Examples of early fabrication of ceramic articles have been found and dated as far back as 23,000 B.C..[17] Man's skill and wider use of ceramic materials steadily increased from the Stone Age through the twentieth-century space age.[16] Translucent porcelain was first manufactured by the Chinese during the T'ang Dynasty. Porcelain was so highly regarded that the Chinese would neither divulge the ingredients, nor correct the proportions of those ingredients.[18] The secret of Chinese porcelain had to be obtained by an example of early industrial espionage. A Jesuit Father named d'Entercolles was able to gain the confidence of Chinese potters and learn the secret in 1717. In 1728 Pierre Fauchard published the, 'The Art of Dentistry' in which he wrote about the use of porcelain in dentistry and used porcelain to enamel metal bases of dentures. In 1774 Alexis Duchateau and Nicholas Dubois de Chemant then fabricated the first successful all porcelain dentures but it was not until 1830 that Stockton and Wildman developed the first porcelain teeth in the United States.[16] Following the development of dental aluminous porcelain by McLean and Hughes in 1965, porcelain inlays were used as dental restorations. It was during the mid to late 1900's that metal-ceramic systems were implemented and still remain the most popular indirect restoration used in dentistry to date. The first international standard published for dental ceramic powders was developed in 1984 and since then the frequency and application of all-ceramic restorations has rapidly increased. Presently we are in the age of zirconia in dentistry. Good chemical and dimensional stability, mechanical strength and toughness, coupled with a Young's modulus in the same order of magnitude as stainless steel alloys have stimulated interest in using zirconia as a ceramic biomaterial.[6] Initial works by Garvie and Nicholson focused on yttrium-oxide partially stabilized zirconia polycrystals (Y-TZP) and has lead to a host of modern zirconia biomaterials and products. These products in dentistry alone range from extracoronal attachments, full and partial coverage crowns, fixed prostheses, veneers, post and cores, orthodontic brackets and implant abutments.[5]

BASIC STRUCTURE AND COMPOSITION OF CERAMICS

The term *ceramic* refers to any product made essentially from a non-metallic inorganic material usually processed by firing at high temperatures to achieve desirable properties. Therefore, concrete, glass, fine crystal and gypsum are all ceramics. Porcelain refers more specifically to a compositional range of ceramic materials.[19] The structure of ceramics may be crystalline or noncrystalline. Crystalline ceramics (eg, quartz) have a regular arrangement of atoms in a lattice pattern. However, noncrystalline ceramics (eg, granite) are typically amorphous in structure. All ceramics have high melting points ranging from 1100 °C to 1700 °C and low thermal and electrical conductivity.[20] The oxide nature of synthetic ceramics also make them nonreactive with excellent biocompatibility.[21] Traditional feldspathic dental porcelain is a vitreous ceramic composed of silica (quartz- SiO_2) and kaolin (clay- $Al_2O_3 \cdot 2SiO_2 \cdot 2H_2O$), metallic pigments, potash feldspar ($K_2O \cdot Al_2O_3 \cdot 6SiO_2$) and soda feldspar ($NaO_2 \cdot Al_2O_3 \cdot 6SiO_2$).[22] Dental ceramics are formed by the union of metallic and non-metallic elements.[21] The metallic pigments are called color frits which are ground together with glass and feldspar, fused and ground again. The color frits include titanium oxide: yellow/brown, manganese oxide: lavender, iron or nickel oxide: brown, cobalt oxide: blue, copper or chromium oxide: green, tin oxide: increases opacity and uranium oxide: increases florescence.[23] Two phases are distinguished during traditional porcelain manufacture which include the vitreous or glass phase and the crystalline or mineral phase.[23] However some structural ceramics may either be partially fused with glass or contain no glass phase at all. The later belongs to a newer group of polycrystalline ceramics which are composed of either Alumina or Zirconia.[24] A schematic diagram of the structure of glass containing ceramic versus polycrystalline ceramic is shown in figure 1.

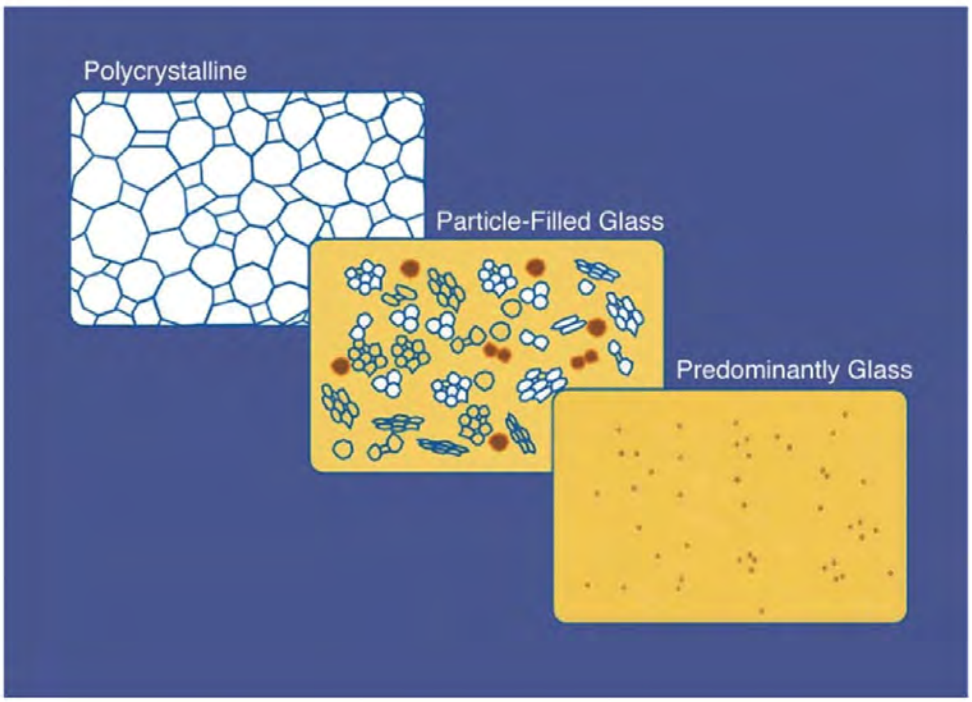

Figure 1. Schematic representation of the structure of three basic compositions of dental ceramics.[25]

CLASSIFICATION OF CERAMICS

Historically, three basic types of ceramic material were developed. These included earthenware, stoneware and porcelain. With advances in the physical properties of ceramics and ceramic technology, the categories for classifying ceramics have broadened considerably. Multiple ceramic systems are now available for use and there is no evidence which supports the universal application of a single ceramic material and system for all clinical situations.[26] Categories are overlapping and include fusing temperature, composition, manufacture, physical properties, core material, type of restoration and clinical application. The classification of dental ceramics according to fusing temperature dates back to the early 1940s and is based on ceramics mainly composed of quartz, feldspar and clay. High-fusing ceramics have a fusing range from 1315 - 1370 °C, medium-fusing from 1090 – 1260 °C and low-fusing from 870-1065 °C.

All-ceramic dental restorations can be fabricated either by machining, slip-casting, and heat-pressing or sintering. This classification is shown in table 1 with examples of product names and manufacturers.

Table 1. Classification of dental ceramic materials according to fabrication technique with examples of commercially available products.[19, 25]

FABRICATION	CRYSTALLINE PHASE	PRODUCTS	MANUFACTURERS
Machined	Zirconia (ZrO_2)	Cercon	Dentsply
		Lava	3M ESPE
		Procera	Nobel Biocare
		IPS e.max ZirCAD	Ivoclar
	Alumina (Al_2O_3)	Procera	Nobel Biocare
	Feldspar ($KAlSi_3O_8$)	Vita Mark II	Vident
	Mica ($KMg_{2.5}Si_4O_{10}F_2$)	Dicor MGC	Dentsply
	Leucite ($KAlSi_2O_6$)	Procad	Ivoclar
	Lithium disilicate ($Li_2Si_2O_5$)	IPS e.max CAD	Ivoclar
Slip-cast	Alumina (Al_2O_3)	In-Ceram alumina	Vident
	Spinel ($MgAl_2O_4$)	In-Ceram spinell	Vident
	Zirconia (ZrO_2)	In-Ceram zirconia	Vident
Heat-pressed	Leucite ($KAlSi_2O_6$)	IPS Empress	Ivoclar
		OPC	Pentron
		Finesse all-ceramic	Denstply
	Lithium disilicate ($Li_2Si_2O_5$)	IPS Empress Esthetic	Ivoclar
		IPS e.max Press	Ivoclar
		OPC 3G	Pentron
	Lithium phosphate (Li_3PO_4)	IPS Empress Cosmo	Ivoclar
	Fluorapatite ($Ca_5(PO_4)F$)	IPS e.max ZirPress	Ivoclar
	Feldspar ($KAlSi_3O_8$)	GC Initial IQ Pressable	GC America
Sintered	Leucite ($KAlSi_2O_6$)	IPS Empress layering ceramic	Ivoclar
	Alumina (Al_2O_3)	Procera Allceram	Nobel Biocare
	Fluorapatite ($Ca_5(PO_4)F$)	IPS Empress layering ceramic	Ivoclar
		IPS e.max Ceram	Ivoclar

The classification of ceramics based on its composition or absence of glass comprises of three basic categories. These include predominantly and particle filled glass ceramics and polycrystalline ceramics.

Predominantly glass

Glass ceramics are partially crystallized amorphous glasses produced by enucleation and growth of crystals in a matrix phase. High melting and leucite reinforced glass ceramics belong to the predominantly glass group. Ceramics with a high glass content are most capable of mimicking the optical properties of natural teeth. These contain small amounts of filler particles which control optical effects such as opalescence, color and opacity.[27] Manufacturers pair most veneering materials with a corresponding substructure ceramic with a compatible flexural strength.

Particle-filled glass

Either crystalline or high-melting glass particles are added to the base glass composition to improve mechanical properties such as strength, thermal expansion and contraction behavior. These high-melting glasses that are stable at the firing temperatures of the ceramic and can be etched for bonding.[28] The term glass ceramics refers to ceramic produced by precipitation of solid particles within the glass using nucleation and growth heating treatments. Particles can also be added mechanically during manufacturing as powder. Examples of glass ceramics include high content lithium disilicate glass-ceramics (IPS e.max Press and IPS e.max CAD, Ivoclar Vivadent, Amherst, N.Y.) and three-dimensional (3-D) interpenetrating-phase composite, in which the filler particles and glass are both continuous in space (In-Ceram, Vita Zahnfabrik, Bad Säckingen, Germany). The filler is alumina, magnesium aluminate spinel or a mixture of 70 percent alumina and 30 percent zirconia.[25]

Polycrystalline

Polycrystalline ceramics do not contain a glass base therefore particles are arranged in a regular crystalline structure that is densely sintered and stronger than glass-based ceramics. CAD/CAM technology is used to produce dental restorations made of polycrystalline ceramic materials. Examples of computer-aided systems include Procera (Nobel Biocare, Göteborg, Sweden), ZirCAD (Ivoclar Vivadent), Cercon Zirconia (Dentsply Prosthetics, York, Pa.), Lava Zirconia (3M ESPE, St. Paul, Minn.) and Vita In-Ceram YZ (Vita Zahnfabrik, Germany).[25] The classification of ceramics based on its composition or absence of glass as it relates to esthetic and structural functions are shown in figures 2, 3, 4 and 5.

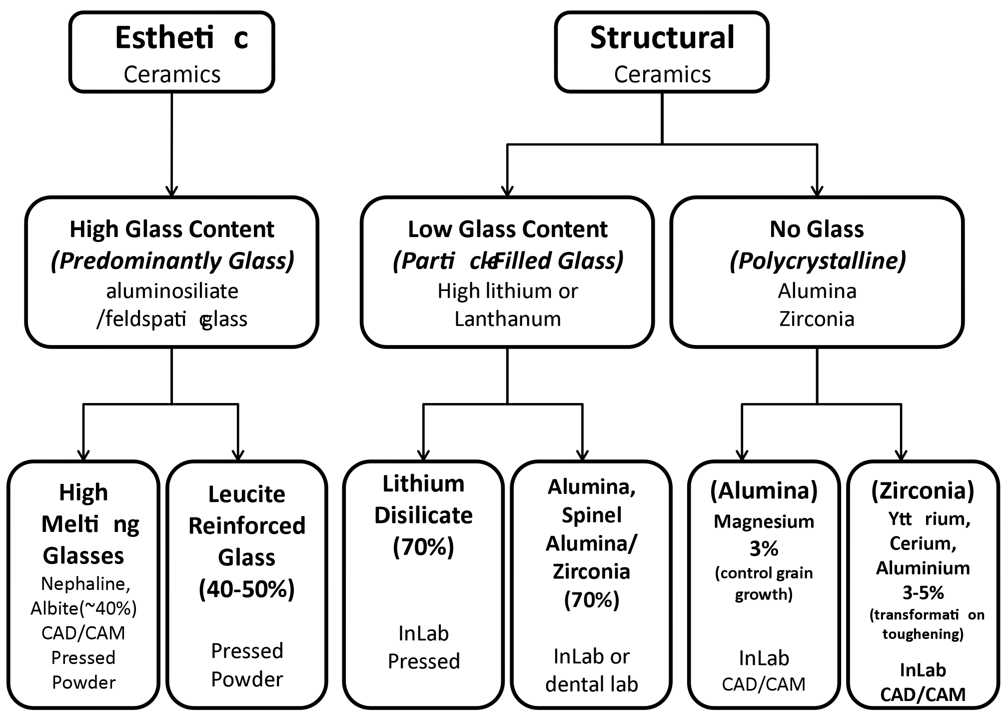

Figure 2. Flow diagram of the classification of dental ceramics by esthetic versus structural restorative requirements.[25]

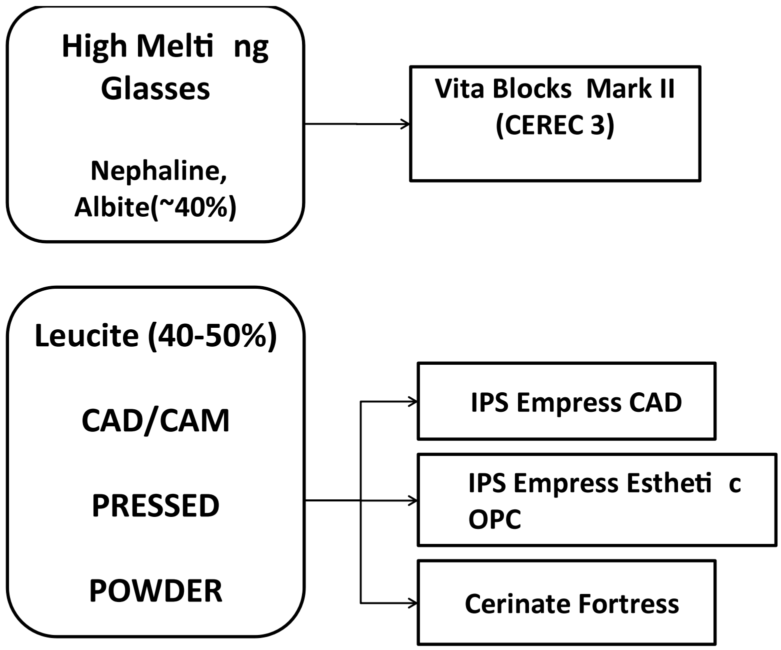

Figure 3. Flow diagram of the sub-classification of esthetic ceramic systems and commercial examples.[25]

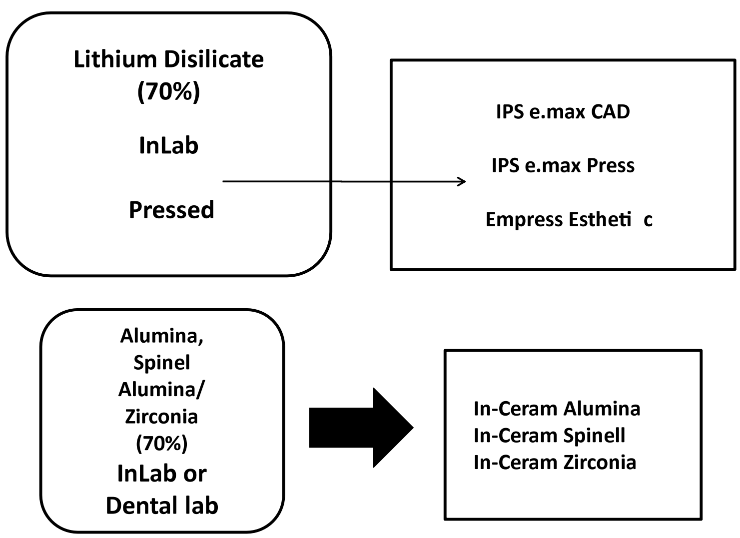

Figure 4. Flow diagram of the sub-classification of structural (particle-filled glass) ceramic systems.[25]

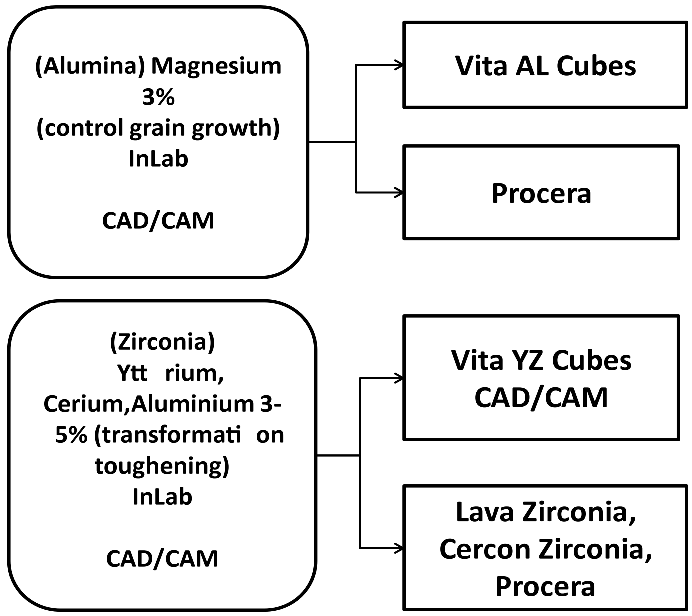

Figure 5. Flow diagram of the sub-classification of structural (polycrystalline) ceramic systems.[25]

The following table lists the classification by glass content as it relates to specific fillers, manufacturers products and specific clinical applications of dental restorations.[27]

Table 2. Classification by glass content as it relates to filler content, commercial examples and specific clinical applications of dental restorations.[27]

	BASE	FILLERS	USES	COMMERCIAL EXAMPLES
HIGHLY FILLED GLASSY CERAMICS	Feldspathic glass	Leucite (~40-55 mass%)	Inlays, Onlays, Veneers, Single unit crowns	Empress (Ivoclar) OPC (Pentron) Finesse All-ceramic (Dentsply)
	Feldspathic glass	Aluminium oxide (~55 mass %)	Single-unit crown	Vitadur-N (Vita)
	Lanthanum	Aluminium oxide (~70 vol %)	Single-unit crowns, anterior three-unit bridges	In-Ceram Alumina (Vita)
	Aluminoborosilicate	Aluminium oxide (~ 50 vol %) Zirconium oxide (~ 20 vol %)	Single-unit crowns, three-unit bridges	In-Ceram Zirconia (Vita)
MODIFIED	Feldspathic glass	Lithium disilicate (~ 70 vol %)	Single-unit crowns, anterior three-unit bridges	Empress estheti c (Ivoclar) 3G (Pentron)
POLYCRYSTALLINE	Aluminium oxide	< 0.5 mass %	Single-unit crowns	Procera (Nobel Biocare)
	Zirconium oxide	Ytt rium oxide (3-5 mass %)	Single-unit crowns	Procera (Nobel Biocare)
	Zirconium oxide	Ytt rium oxide (3-5 mass%)	Single-unit crowns, Three-unit bridges, Four-unit bridges	Cercon (Dentsply) Lava (3M-ESPE) Y- Z (Vita)

ZIRCONIA

Introduction:

The metal Zirconium comes from the Arabic word ‘zargon,’ meaning golden in color. This is further derived from two Persian words: ‘zar,’ meaning gold and ‘gun,’ meaning color. The metal dioxide zirconia was discovered by the German chemist Martin Heinrich Klaproth in 1789 and was isolated by the Swedish chemist Jöns Jakob Berzelius, in 1824.[5-6] The interest in using zirconia as a ceramic biomaterial originated from its good chemical and dimensional stability, mechanical strength and toughness and a Young’s modulus in the same order of magnitude as stainless steel alloys.[6]

Compared to conventional ceramic systems, yttrium-stabilized tetragonal zirconia (Y-TZP) ceramics have superior mechanical properties which ensure a broad application in dentistry.[29] Constant improvements in the mechanical properties of dental ceramics are responsible for the increased utilization of metal-free restorations.[30-31] Zirconia systems that are currently used in dentistry contain either greater than 90% zirconia dioxide which is the Y-TZP ceramic and glass infiltrated ceramics with only 35% partially stabilized zirconia. Their clinical applications range from single implant abutments and single crowns to fixed partial dentures with several units.[32-33]

Zirconia is a well known polymorph that occurs in three forms: monoclinic (M), cubic (C) and tetragonal (T). Pure zirconia is monoclinic at room temperature and stable up to 1170 °C. Above this temperature it transforms first into its tetragonal phase and subsequently into its cubic phase at 2370 °C. During cooling, a T-M transformation takes place which is associated with a 3-4% volume expansion. Stresses generated by expansion originate cracks in pure zirconia which causes it to break into pieces at room temperature. In 1929 Ruff et al demonstrated stabilization of the C-phase by the addition of stabilizing oxides such as CaO, MgO, CeO_2 and Y_2O_3.[34] According to theory, tensile stress in crack tips leads to a phase transition of the lattice structure from the metastable tetragonal phase to the monoclinic phase. This phase transition is correlated with a volume increase of 4-5% in which crack propagation is inhibited by creating a compressive stress.[35]

Types of zirconia used in dentistry:

Only three types of zirconia-containing ceramic systems are used in dentistry to date. These are yttrium cation-doped tetragonal zirconia polycrystals (3Y-TZP), magnesium cation-doped partially stabilized zirconia (Mg-PSZ) and zirconia-toughened alumina (ZTA).[36]

Yttrium cation-doped tetragonal zirconia polycrystals (3Y-TZP)

Sintering conditions influence the grain size of 3Y-TZP. This has been proven by the fact that higher sintering temperatures and longer sintering times result in larger grain sizes. Above 1 µm grain size, 3Y-TZP is less stable and more susceptible to spontaneous transformation from phases T to M.[37-38] On the other hand, grain sizes below 1 µm have a lower transformation rate.[39] This solidifies the argument that sintering conditions are critical due to their strong impact on both stability and mechanical properties of the final product and correlation with grain size.[36-38] Furthermore, below the grain size of 0.2 µm, transformation is not possible and this leads to a reduction in fracture toughness.[40] Firing veneering porcelain during fabrication of dental restorations can promote reverse transformation resulting in surface relaxation of compressive stresses and a decrease in strength. Additionally, transformation reversibility does not provide a mechanism for fixing previously introduced flaws.

Depending on sintering temperature, the microstructure of 3Y-TZP ceramics for dental applications consists of small equiaxed grains shown in the micrograph below (0.2–0.5 µm in diameter).[11]

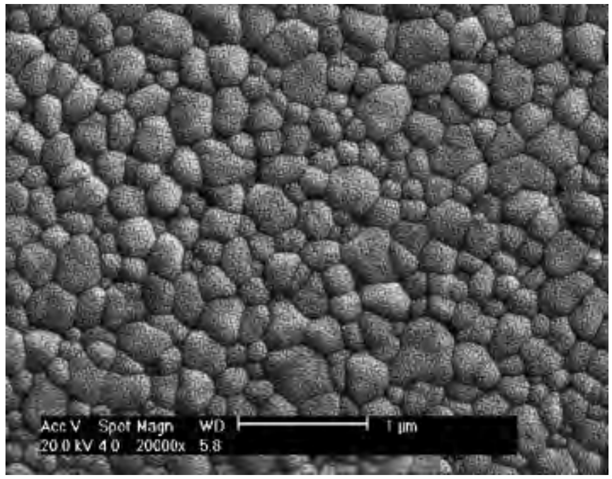

Figure 6. Scanning electron micrograph of 3Y-TZP for dental applications sintered according to manufacturer's recommendations (Cercon®, Dentsply Ceramco).[36]

Controlling the final surface state of 3Y-TZP for biomedical applications was more recently justified by Zhang et al. who studied the effect of sharp indentation damage on the long-term performance of 3Y-TZP. It was shown that both sandblasting and sharp indentations even at very low loads are detrimental to the long-term performance of 3Y-TZP when tested in cyclic loading.[41-43] For that reason, irrespective of strength, long-term performance and reliability should also be considered for dental applications.[36] The mechanical properties exceed those of all other available dental ceramics, with a flexural strength of 800 to 1000 MPa and fracture toughness of 6 to 8 MPa.

The Weilbull modulus strongly depends on the type of surface finish and the processing conditions.[44] Only one short crack is emanating from one of the corners of the indentation shown in the figure below. The absence of cracking from the other corners is indicative of the occurrence of the transformation toughening mechanism.

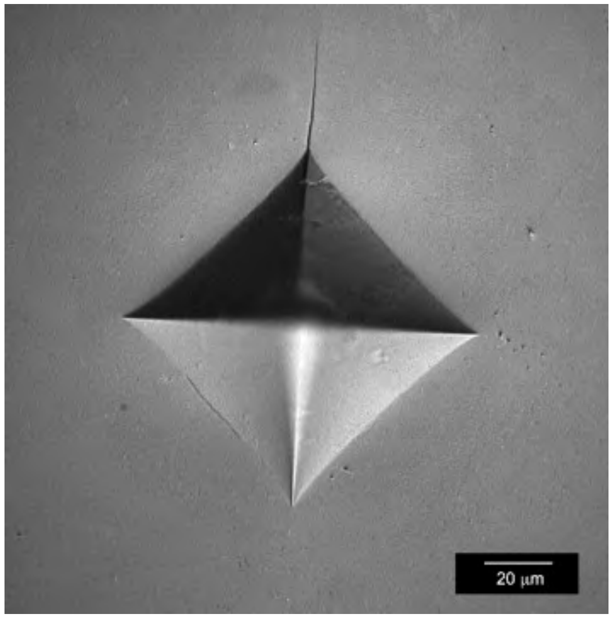

Figure 7. Optical micrograph of a Vickers indentation in a 3Y-TZP for Dental Applications (98.1 N Load).[36]

Glass-infiltrated zirconia-toughened alumina (ZTA)

The stress-induced transformation capability of zirconia can also be utilized by combination with an alumina matrix, leading to a zirconia-toughened alumina (ZTA).[45] In-Ceram® Zirconia® (Vident™, Brea, CA), was developed by adding 33 vol.% of 12 mol% ceria-stabilized zirconia (12Ce-TZP) to In-Ceram® Alumina®[46] as shown in figure 7.

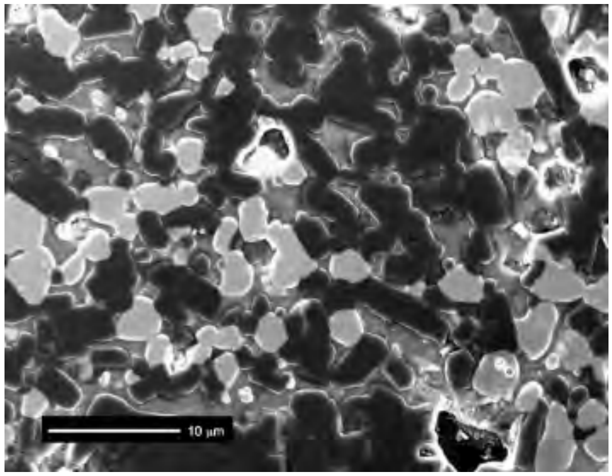

Figure 8. Scanning electron micrograph of In-Ceram® Zirconia® (Vident™, Brea, CA) showing zirconia grains appearing brighter in contrast to darker alumina grains.[36]

In-Ceram® Zirconia® can be processed by either slip-casting or soft machining. One of the advantages of the slip-cast technique is that there is very limited shrinkage but porosity is greater than that of sintered 3Y-TZP and comprises between 8 and 11%. This partially explains the generally lower mechanical properties of In-Ceram® Zirconia® when compared to 3Y-TZP dental ceramics. However, Ce-TZP ceramics usually exhibit better thermal stability and resistance to low temperature degradation than Y-TZP under similar thermo-cycling or aging conditions.[47]

Partially stabilized zirconia (Mg-PSZ)

Although a considerable amount of research has been dedicated to magnesia partially stabilized zirconia (Mg-PSZ) for possible biomedical applications, this material has not been successful due mainly to the presence of porosity, associated with a large grain size (30–60 µm) that can induce wear. The microstructure consists of tetragonal precipitates within a cubic stabilized zirconia matrix.[36]

CERAMIC FABRICATION TECHNIQUES

Among restorative materials, ceramics are most capable of reproducing the appearance of natural teeth. As a result, the proportion of restorative treatments using all-ceramic prostheses is rapidly growing. Furthermore, ceramics having similar composition may be fabricated by different laboratory techniques, and each method of processing can result in a different distribution of flaws, opportunity for depth of translucency, and accuracy of fit. These differences should be important to the clinician because they persist beyond the walls of the dental laboratory and affect clinical performance.[12] Fabrication methods include powder condensation, slip casting, hot pressing and CAD CAM techniques.

Powder Condensation

The ceramic prosthesis is formed by the application of moist porcelain powder in a traditional artistic brush-on technique. Excess moisture is then removed to compact the powder particles. Further compaction of the porcelain under vacuum pressure takes place by viscous flow of the glassy component during firing. Large amounts of residual porosity can result during this process. On a microscopic scale, glassy regions form which separate the crystalline particles that strengthen the material. The overall result can be a relatively low or wide variation in strength due to the porosity and discontinuous nature of the crystalline phase. The advantage of this technique is the production of a ceramic with greater translucency than can be achieved using other methods.[48] Power condensation is used for materials as the esthetic veneer layer on stronger cores and frameworks.

Slip Casting of Dental Ceramics

The process of slip casting begins with solid particles suspended in a fluid. This mixture is referred to as a slip and is of low viscosity. The fluid is usually water and the solid component consists of fine ceramic powder. The slip is poured into a gypsum mold or negative replica of the desired geometric shape. As the slip traverses the walls of the mold, water is extracted via capillary action. As a result, some of the powder particles in the slip become compacted against the walls of the mold. The thin layer of green ceramic that forms becomes the framework of the desired geometric shape. This framework can be removed from the mold after partial sintering until it is strong enough to support its own weight and the residual slip is discarded. Slip casting results in a very porous ceramic and must be either infiltrated with molten glass or fully sintered before veneering porcelain can be applied. Crystalline particles form a strong continuous network throughout the framework which has a greater fracture resistance than ceramics produced by powder condensation. This method is limited to use by products for glass infiltration such as In-Ceram, Vita Zahnfabrik. Application of slip casting in dentistry is limited by a complicated series of steps required for fabrication which can challenge accuracy.[49-51] Consequently, internal defects may arise from incomplete glass infiltration and weaken the final restoration.[12]

Heat Pressing of Dental Ceramics

The first step involved in the fabrication of heat pressed dental restorations is a wax-up of the desired geometric shape of the restoration. This is invested in gypsum and a lost wax method is used to fabricate molds for pressable dental ceramics. Manufacturers supply prefabricated pressable ceramic ingots made of crystalline particles distributed throughout a glassy material with composition similar to that of powder porcelains. In contrast to powder porcelains, pressable ceramics are much less porous with a higher crystalline content. This lack of porosity is attributable to a well controlled manufacturing process which utilizes non-porous glass and heat treatment that transforms some of the glass into crystals. This process produces a homogeneous material in a less technique sensitive manner. In the dental laboratory, the pressable ingots are heated to a temperature at which they become a highly viscous liquid, and they are slowly pressed into the lost wax mold. However, the higher crystalline content and lack of porosity do not increase fracture resistance when compared with traditional fabrication techniques.[52] Pressable ceramics usually have application only as core and framework materials. Pressable veneering materials, such as IPS e.max, ZirPress (Ivoclar-Vivadent) are available, but the depth of layering for esthetics may be limited when using pressable ceramics for veneering purposes.

Although the pressing process is less technique sensitive than both the powder condensation and slip casting techniques, proper preparation of an appropriate investment material, burnout and pressure setting for compressed air are required for a successful press. The thickness of the substructure material must be adequate to prevent cracking under pressure in the press oven and its composition also needs to be compatible with the veneering layer. Furthermore, the overlay cannot exceed the thickness of its core and porosities can occur if the pressing temperature it too high.

CAD-CAM of Dental Ceramics

CAD/CAM ceramics are also available as prefabricated ingots. Unlike pressable ceramics, the ingots used for CAD/CAM restorations are either presintered or densely sintered ceramic blocks.

- *Soft machining of pre-sintered blanks*

Direct ceramic machining of pre-sintered 3Y-TZP has become increasingly popular in dentistry since its development in 2001. It is now offered by a growing number of manufacturers.[53] Briefly, the die or a wax pattern is scanned, an enlarged restoration is designed by computer software (CAD) and a pre-sintered ceramic blank is milled by computer aided machining. The restoration is then sintered at high temperature.

Several variations of this process exist depending on how the scanning is performed and how the large sintering shrinkage of 3Y-TZP (25%) is compensated for. For example, both contact scanners and non-contact scanners are available. Overall, non-contact scanners are characterized by a higher density of data points and a greater digitizing speed compared to contact scanners. Typically the 3Y-TZP powder used in the fabrication of the blanks contains a binder that makes it suitable for pressing. The binder is later eliminated during the pre-sintering step. The blanks are manufactured by cold isostatic pressing. The mean pore size of the compacted powder is very small, in the order of 20 to 30 nm with a very narrow pore size distribution.[54]

Restorations can be colored after machining by immersion in solutions of various metal salts such as cerium, bismuth or iron before final sintering. At the final sintering, temperature influences the color obtained. Alternatively, colored zirconia can be obtained by small additions of various metal oxides to the starting powder. For example, small additions of alumina have been shown to act as a sintering aid, allowing the use of lower sintering temperatures and times.

The ingots are milled or cut by computer-controlled tools which are programmed to fabricate a virtual design of the restoration.[12] Following construction of cores, restorations are finally veneered with porcelains of matching coefficient of thermal expansion. The nature of the interface between 3Y-TZP and the veneering porcelain has not been thoroughly studied. The veneering porcelain is baked at 900 °C, with a hold time of 1 min. Representative systems utilizing soft machining of 3Y-TZP for dental restorations are Cercon® (Dentsply International), Lava™ (3M™ ESPE™), Procera® zirconia (Nobel Biocare™), YZ cubes for Cerec InLab® (Vident™) and IPS e.max® ZirCAD (Ivoclar Vivadent).

In the case of presintered ceramics, the ingots are porous to facilitate fast milling without bulk fracture of the ceramic. In order to achieve good accuracy of fit, computer software must compensate for the shrinkage that occurs during sintering.

- *Hard machining of 3Y-TZP and Mg-PSZ*

At least two systems, Denzir® (Cadesthetics AB) and DC-Zirkon® (DCS Dental AG) are available for hard machining of zirconia dental restorations. Y-TZP blocks are prepared by pre-sintering at temperatures below 1500 °C to reach a density of at least 95% of the theoretical density. The blocks are then processed by hot isostatic pressing at temperatures between 1400 and 1500 °C under high pressure in an inert gas atmosphere leading to a very high density in excess of 99% of the theoretical density. The blocks can then be machined using a specially designed milling system. Due to the high hardness and low machinability of fully sintered Y-TZP, the milling system has to be particularly robust. A study by Blue et al. demonstrated that Y-TZP was significantly harder to machine than fully sintered alumina with lower material removal rates.[54]

In contrast, densely sintered ceramics are available in non-porous ingots, which are already completely sintered and more difficult to mill. Glass infiltrated CAD-CAM ingots have similar composition to slip cast ceramics and eliminates the technique sensitivity of slip casting. After milling, the porosity is eliminated by molten glass infiltration. Dental CAD-CAM systems have been available for 20 years. In recent years, the increasing use of polycrystalline alumina and zirconia as framework materials and the increasing popularity and variety of CAD-CAM systems seem to be mutually accelerating trends[12].

CAD/CAM techniques are gaining importance in fabricating crowns and fixed partial dentures (FPDs) made of Y-TZP. Luthardt et al tested the hypothesis that surface flaws and microcracks are induced by grinding crowns and analyzed grinding-induced surface layer crack size caused by machining Y- TZP. Ceramic disks and sectioned cylinders with polished separation planes were used to analyze the grinding procedure. The inner surface grinding of crowns was also simulated by variation of feed and cutting depth. SEM was used for a quantitative assessment of the machined surface. It was determined that crack length is not significantly influenced by the grinding parameters tested and the type of material removed varied with the cutting depth and feed. Grinding induced surface flaws and microcracks were produced internally at the top surface of the crowns. Half-cylinders machined under conditions simulating inner surface grinding of crowns showed crack lengths between 2 and 15 μm. It was concluded that sectioned specimens with polished section planes are suitable for the analysis of the grinding process using the face and peripheral grinding procedure. The inner surface grinding of fixed restorations is the most challenging step in CAM of crowns and FPDs. The number and shape of the active diamond grains was found to be most significant.[55]

IN VITRO CERAMIC TEST METHODS

Clinical failures can be simulated by blunt contact loading, cyclic fatigue loading, and loading in an aqueous environment.[13] Regardless of these efforts, physical testing does not guarantee a clinically relevant mode of failure. It must be noted that clinically failed crowns display bulk fracture which originates in the core ceramic[56] and surface chipping from contact surface wear.[57] Specimens should be loaded using a clinically applicable method that reproduces clinical modes of failure. The relevance of physical tests which do not fracture in this manner is uncertain. However, experiments with novel loading geometry have reproduced that failure mode in the laboratory.[14]

Furthermore, it is important to test multi-layered specimens with actual or simulated dentin, luting agent, core ceramic, and veneering ceramic layers since the components and treatment of each layer determines the strength of the entire restoration. The strength of the core ceramic can be altered by luting as well as the surface treatments used to prepare the core ceramic for the veneering[58], and the presence of the luting agent also has an effect on ceramic strength.[59] There will always be a need for some simple standard geometry specimens to study the micromechanisms associated with crack growth. However, clinicians should not assume that simple specimens are predictive of clinical performance. In other words, ceramic specimens that have been finely polished, tested dry, or loaded quickly can be expected to have much higher strength than prostheses fabricated from the same materials, and the relative ranking of commercial products may change depending on test method.[12]

The reporting of Weibull statistics to describe ceramic strength data has not improved significantly. The failure load and strength of a ceramic prosthesis or test specimen is controlled by the size of the largest flaw in the highly stressed location and not the average flaw. The Weibull modulus is a measure of variation in strength such that a higher Weibull modulus corresponds to less variation. The Weibull modulus can be more important than the median strength for predicting clinical performance since it predicts the effect of prosthesis size on strength and because it controls the stress level corresponding to low probabilities of failure. The median strength corresponds to a 50% chance of failure, but clinicians are not interested in such a high failure rate. Grigg concluded from his systematic review that clinicians should be cautious and note the sample size when interpreting in vitro studies because a study may conclude no difference in Weibull modulus between groups when there is not enough statistical power to detect a difference.[12]

Kelly et al investigated common tests of single-unit restorations involving loading specimens through spherical indenters and loading curved incisal edges against flat compression platens. The clinical validity of such failure testing was questioned in order to develop more relevant test methods. The characteristics of the traditional load-to-failure test was reviewed and contrasted with characteristics of clinical failure for all-ceramic restorations. The goal was to explain any existing discrepancies. Literature regarding intraoral conditions was reviewed to develop an understanding of how laboratory testing could be revised. Important variables used in simulating clinical conditions were described, along with their laboratory evaluation. It was concluded that traditional fracture tests of single unit all-ceramic prostheses are inappropriate, because they do not create failure mechanisms seen in retrieved clinical specimens. The author proposed the need for validating tests to elucidate the role of cements, bonding, occlusion and substructure materials in the success of fixed prostheses in order to make meaningful comparisons. Research over the past 6 years has shown that crack systems mimicking clinical failure can be produced in all-ceramic restorations under appropriate conditions.[13]

ESTHETICS AND OPTICAL PROPERTIES OF CERAMIC RESTORATIONS

All-ceramic restorations have been advocated for superior esthetics. Various materials have been used to improve ceramic core strength, but it is unclear how they affect the opacity of all-ceramic systems.

Heffernan et al compared the translucency of 6 all-ceramic system core materials at clinically appropriate thicknesses. Disc specimens 13 mm in diameter and 0.49 +/- 0.01 mm in thickness were fabricated (n = 5 per group). Materials used were IPS Empress dentin, IPS Empress 2 dentin, In-Ceram Alumina core, In-Ceram Spinell core, In-Ceram Zirconia core, and Procera AllCeram core. Empress and Empress 2 dentin specimens also were fabricated and tested at a thickness of 0.77 +/- 0.02 mm according to the manufacturer's recommended core thickness of 0.8 mm. A high-noble metal-ceramic alloy served as the control, and Vitadur Alpha opaque dentin was used as a standard. Sample reflectance (ratio of the intensity of reflected light to that of the incident light) 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. Contrast ratios were calculated from the luminous reflectance (Y) of the specimens with a black (Yb) and a white (Yw) backing to give Yb/Yw with CIE illuminant D65 and a 2-degree observer function (0.0 = transparent, 1.0 = opaque). Contrast ratios in order of most translucent to most opaque were evaluated. A range was found of ceramic core translucency at clinically relevant core thicknesses. In order of decreasing translucency, the ranges were Vitadur Alpha dentin (standard) > In-Ceram Spinell > Empress, Procera, Empress 2 > In-Ceram Alumina > In-Ceram Zirconia, 52 SF alloy.[3]

Heffernan in a follow-up study addressed all-ceramic core materials with various strengthening compositions with a range of translucencies. It is unknown whether translucency differs when all-ceramic materials are fabricated similarly to the clinical restoration with a veneered core material. For this reason, this study compared the translucency of 6 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 as described in Part I of this study and veneered with their corresponding enamel 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 (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. Measurements were repeated after a glazing cycle and contrast ratios were then calculated. Statistics used to analyze results found significant differences in contrast ratios were found 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. This study concluded that there was a range of translucency in the veneered all-ceramic systems. Such variability may affect their ability to match natural teeth. The glazing cycle resulted in decreased opacity for all test materials except the completely opaque In-Ceram Zirconia and metal-ceramic specimens.[4]

Raptis et al addressed the difficulty in restoring anterior teeth and identified many different ceramic systems that can be used to achieve highly esthetic results. The purpose of this paper was to briefly discuss metal-ceramics with porcelain margins, Dicor, In-Ceram, Cerestore, Hi-Ceram, IPS-Empress, Cerapearl, Optec, and CAD/CAM ceramics in relation to a material that transmits and refracts light like a natural tooth. Evaluation was performed on the optical behavior of: (1) metal-ceramic crowns with castings 2 mm short of the shoulder preparation and 360-degree porcelain margins; (2) In-Ceram Spinell restorations; and (3) IPS Empress restorations, and this was compared with metal-ceramic crowns with copings to the shoulder preparation and 180-degree porcelain margins. Light transmission characteristics and color matching were subjectively evaluated by five experienced prosthodontists who did not participate in this clinical study. All evaluators agreed that PFM restorations did not allow light transmission for both margin preparations, In-Ceram Spinell allowed better transmission than PFM restorations and IPS empress allowed superior light transmission.[60]

Spear et al recognized the need for ceramic materials in dentistry to evolve as patients' demand for esthetic restorations increases. The goal of this article was to offer guidance to the practitioner in selecting the appropriate all-ceramic systems for crowns when faced with different esthetic demands. It was concluded that clinicians should reserve dental ceramics with high translucency for clinical applications in which high-level esthetics are required and the restoration can be bonded to tooth structure. Ceramics with high strength tend to be more opaque and pose a challenge when trying to match natural tooth color, but they can mask discoloration when present. It was suggested that knowledge of the optical properties of available ceramic systems enable the clinician to make appropriate choices when faced with various esthetic challenges.[9]

Komine et al recognized fluorescence, opalescence, and translucency as critical for restorative materials to mimic the optical properties and appearance of natural teeth. This case report was presented which described the restoration of multiple anterior teeth with CAD/CAM-fabricated glass-infiltrated aluminum oxide ceramic (In-Ceram Alumina, Vita Zahnfabrik) crowns and illustrated the technical steps to achieve an adequate amount of fluorescence in the ceramic veneering material. CAD/CAM aluminum oxide ceramic copings and frameworks were suggested as predictable and successful when replacing missing tooth structure and imitating optical properties of natural teeth. It was concluded that a modified layering technique can enhance fluorescence within the veneering ceramic and provide a better esthetic appearance of glass-infiltrated aluminum oxide ceramic restorations.[61]

TOOTH PREPARATION DESIGN

Similar to the design of metal-ceramic restorations, zirconia-based restorations utilize a high strength ceramic framework to withstand the forces of cyclic loading. The strength of this structure depends on both the fracture resistance as well as a suitable preparation design.[62] For Y-TZP- based single crowns; the shoulder preparation is recommended from both a mechanical and periodontal point of view however, a slight chamfer can be used as a less invasive option. The deep or pronounced chamfer preparation is not recommended.[15] The assumption that increased material thickness automatically produces greater strength was disproven by Fenske et al in an in vitro study investigating fracture resistance as a function of shoulder width. Crowns made from pressable leucite-reinforced ceramic (IPS Empress, Ivoclar Vivadent, FL-Schaan) displayed an increase in Weilbull modulus for more delicate preparations.[63] Glass ceramic crowns investigated by Friedlander and Doyle exhibited highest fracture resistance for a preparation design with a total convergence angle of 10°, 1.2 mm shoulder finish line and sharp axiogingival line angle. All-ceramic single tooth restorations with a minimum thickness of 1.6 mm and a load bearing capacity of 2000 N were found to be a suitable restoration. Whereas any decrease in minimal thickness was associated with the choice of ceramic capable of withstanding higher loads.[64]

Schmidt performed a prospective clinical trial to evaluate the 3-year clinical results of anterior teeth restored with 0.3-mm-thick zirconia copings and feather-edged marginal preparations. Ten patients received 19 single-tooth restorations in the anterior maxilla to restore severely decayed teeth. After a mean observation period of 39.2 months, no material fracture occurred and all crowns had acceptable surfaces. A survival and success rate of 100% was recorded, which suggests that the clinical method may be a reliable treatment modality for restoring severely compromised anterior teeth. However, there is greater need for scientific evidence on the effect of preparation designs on the strength of zirconia-based restorations.

BONDING VENEERING PORCELAIN TO CERAMIC CORES

Sufficient bond strength between the veneering ceramic and the substructure is a concern for the long-term clinical success of zirconia restorations. Bond strength is determined by the strength of chemical bonds, mechanical interlocking, type and concentration of defects at the interface, wetting properties and the degree of compressive stress in the veneering ceramic.[65-67]

Liu et al investigated the fracture and interfacial delamination origins of bilayer ceramic composites for dental restorations. Alumina and zirconia (Y-TZP) based bilayer ceramic dental composites with either 1:1 or 2:1 core to veneer thickness ratios were fabricated using a multiple step firing procedure. Flexural strengths were determined by three-point bend tests and finite element analysis was used to perform a direct fractography investigation to elucidate the origins of fracture and interfacial delamination. The physical properties mismatch between core ceramic and veneering porcelain was also evaluated. Numerical simulations by finite element analysis indicated that interfacial delamination inY-TZP based bilayer composites was clearly linked to flexural strength mismatches between the veneer porcelain and core ceramics. It was determined that this may be prevented by increasing the flexural strength of the veneer porcelain to above 300 MPa. It was also postulated that the formation of microcracks in alumina core immediately one grain-thick under the veneer-core interface were possibly introduced during the veneering operation.[68]

Anunmana et al investigated the interfacial toughness of bilayer dental ceramics based on a short-bar, chevron-notch test. Eight veneered bar specimens were prepared for each of three groups analyzed. The three groups comprised of glass veneer bonded to itself, lithia-disilicate glass-ceramic core and yttria-stabilized polycrystalline zirconia core ceramic. T-shaped short-bars of the core ceramic were prepared according to the manufacturer's recommendations. V-shaped notches were prepared by using 25 μm thick palladium foil, leaving the chevron-notch area exposed, and the bars were veneered with a thermally compatible glass veneer. Specimens were kept in distilled water for 30 days before performing a fracture toughness test using the indentation-strength technique. The mean interfacial toughness of the lithia-disilicate glass-ceramic core did not significantly differ from that of the control glass veneer bonded to itself. However, the difference between the mean interfacial toughness of the lithia-disilicate glass-ceramic core and yttria-stabilized polycrystalline zirconia core ceramic groups were statistically significant. For bilayer all-ceramic restorations with high-strength core materials, the veneering ceramics are the weakest link in the design of the structure. Since all-ceramic restorations often fail from chipping of veneer layers or crack initiation at the interface, the effects of thermal mismatch stresses when designing oral prosthesis should be noted.[69]

Chaiyabutr et al addressed the problem that clinicians are frequently faced with a challenge in selecting materials for adjacent restorations, particularly when one tooth requires a zirconia-based restoration and the next requires a veneer. It was also addressed that little information is available about the use of veneering ceramics over a zirconia-based material. The purpose of this study was threefold: (1) to study the influence of hydrofluoric acid-etched treatment on the surface topography of the zirconia veneering ceramic, (2) to test the bond strength of zirconia veneering ceramic to enamel, and (3) to evaluate the flexural strength and the elemental composition of ceramic veneers. Three zirconia veneering ceramics (Cerabien CZR (CZ), Lava Ceram (L), and Zirox (Z)) and 4 conventional veneering ceramics (Creation (C), IPS d.Sign (D), Noritake EX-3 (E), and Reflex (R)) were evaluated. Twenty ceramic bars of each material were fabricated and surface treated with hydrofluoric acid according to the manufacturer's recommendations. Ten specimens from each group of materials were examined with a profilometer, and a sample of this group was selected for quantitative evaluation using a scanning electron microscope (SEM). Another 10 acid-etched specimens from each group of materials were treated with silane prior to cementing with resin cement (Variolink II) on enamel surfaces. These luted specimens were loaded to failure in a universal testing machine in the shear mode with a crosshead speed of 0.05 mm/min. The data were analyzed with a 1-way ANOVA, followed by Tukey's HSD test (alpha=.05). An additional 10 ceramic bars from each material group were fabricated to evaluate flexural strength and elemental composition. The flexural strength (MPa) of each specimen was determined by using a 4-point-1/4-point flexure test. A Weibull statistic tested the reliability of the strength data and pairwise differences among the 7 groups were evaluated at confidence intervals of 95%. The chemical composition of each bar was determined by energy dispersive spectroscopy (EDS). The results showed a significant difference in the surface roughness in all testing groups. Conventional veneering ceramics (groups C and R) had a mean surface roughness higher than the groups of zirconia veneering ceramics (P<.001). Group D showed no difference in surface roughness compared with the groups of zirconia veneering ceramics. The SEM micrographs revealed differences in the acid-etched surfaces of ceramics. Zirconia veneering ceramics were smooth, with some groove formations, while conventional veneering ceramics had an amorphous, spongy-like structure with numerous microporosites. The mean bond strength (SD) of zirconia veneering ceramics to enamel revealed a significant difference. Group R-Reflex (25.16 (3.40) MPa) followed by group C-Creation (22.51 (2.82) MPa) had significantly higher mean bond strength than the groups of zirconia veneering ceramics (P<.001, P=.009 respectively). Groups D-d.sign (16.54 (2.73) MPa) and E-Noritake EX-3 (17.92 (3.39) MPa) showed no differences. Only group L-Lava Ceram (9.45 (1.62) MPa) exhibited significantly lower mean bond strength when compared with conventional veneering ceramics (P<.001). For flexural strength, only 1 group, group CZ-Cerabien CZR, had a significantly lower flexural strength than all other groups (P<.001). It was concluded that effective ceramic interface management, such as acid etching and enamel bonding, is essential for successful ceramic laminate veneer restorations. Not all zirconia veneering ceramics display the same quality of surface roughness after hydrofluoric acid etching and the same bond strength to enamel when used as laminate veneer materials.[70]

Ozkurt et al investigated the shear bond strength (SBS) of veneering ceramics to zirconia by testing four types of zirconia ceramics (Zirkonzahn, Cercon, Lava, DC-Zirkon). For each zirconia system, 30 disk specimens were layered with IPS e.max Ceram, Vita VM9, and a manufacturer-recommended veneering ceramic. SBS test was conducted, and fracture surface analysis was also performed to determine the failure modes. One-way ANOVA, two-way

ANOVA, and Tukey's HSD tests were used to analyze the data. On shear bond strength between zirconia and their recommended veneering ceramics, statistically significant differences were observed among the different zirconia systems ($p<0.001$). DC-Zirkon exhibited the highest SBS value (40.49+/-8.43 MPa), followed by Lava (27.11+/-2.72 MPa), Zirkonzahn (24.46+/-3.72 MPa), and Cercon (20.19+/-5.12 MPa). On shear bond strength to IPS e.max Ceram and Vita VM9, significantly lower ($p<0.001$) were observed for these veneering ceramics than their recommended veneering ceramics for DC-Zirkon and Lava. For Zirkonzahn and Cercon, similar SBS values were observed for all kinds of veneering ceramics ($p>0.05$). It was concluded that the bonding of manufacturer-recommended veneering ceramic to the zirconia substructure varied for different types of zirconia.[71]

Al-Dohan et al addressed the reported delamination of veneering porcelain from the underlying ceramic substrates of all-ceramic restorations. The goal was to determine if the cause was a weak interface between the veneering and the core porcelains or fracture through the veneering porcelain itself. The study investigated the strength of the substructure and veneering porcelain interface by testing all-ceramic systems with their respective veneering porcelains. These systems included IPS-Empress2 with Eris (IE), Procera AllCeram with AllCeram (PA), Procera AllZircon with CZR (PZ), and DC-Zircon with Vita D (DC). The veneering porcelain recommended by the manufacturer for each material was fired to the ceramic core. A metal ceramic (MC) combination was tested as a control group. Sixty specimens, 12 for each system and control, were made from 1 master die. A cylinder of veneering porcelain 2.4 mm in diameter was applied using a specially designed aluminum split mold. After firing, the specimens were placed in a mounting jig and subjected to shear force in a universal testing machine. Load was applied at a crosshead speed of 0.50 mm/min until failure. Average shear strengths (MPa) were analyzed with a 1-way analysis of variance and the Tukey test (alpha=.05). The failed specimens were examined microscopically at original magnification x20 to classify the mode of failure as cohesive in the core, cohesive in the veneer, or adhesive at the interface. The mean shear strengths (+/-SD) in MPa were metal ceramic control 30.16 +/- 5.88; IPS-Empress2 bonded to Eris 30.86 +/- 6.47; Procera AllZircon bonded to CZR 28.03 +/- 5.03; DC Zircon bonded to Vita D 27.90 +/- 4.79; and Procera AllCeram bonded to AllCeram 22.40 +/- 2.40. All systems except for Procera AllCeram with AllCeram were not significantly different from the metal ceramic control. Microscopic examination showed that adhesive failure, or complete delamination, did not occur between the compatible ceramic core and veneering materials. Failure primarily occurred near the interface with residual veneering porcelain remaining on the core. IPS-Empress2 bonded to Eris exhibited cohesive failure in both the core and the veneer. It was concluded that the bond strengths of 3 of the tested all-ceramic materials, IPS-Empress2 with Eris, Procera AllZircon with CZR, and DC-Zircon with Vita D were not significantly different from the control metal ceramic group.[72]

Guess et al performed a study to evaluate the shear bond strength between various commercial zirconia core and veneering ceramics, and to investigate the effect of thermocycling. The Schmitz-Schulmeyer test method was used to evaluate the core-veneer shear bond strength (SBS) of three zirconia core ceramics (Cercon Base, Vita In-Ceram YZ Cubes, DC-Zirkon) and their manufacturer recommended veneering ceramics (Cercon Ceram S, Vita VM9, IPS e.max Ceram). A metal ceramic system (Degudent U94, Vita VM13) was used as a control group for the three all-ceramic test groups (n = 30 specimens/group). Half of each group (n = 15) was thermocycled (5-55 degrees C, 20,000 cycles). Subsequently, all specimens were subjected to shear force in a universal testing machine. Fractured specimens were evaluated microscopically to determine the failure mode. The initial mean SBS values in MPa+/-S.D. were found to be 12.5+/-3.2 for Vita In-Ceram YZ Cubes/Vita VM9, 11.5+/-3.4 for DC-Zirkon/IPS e.max Ceram, and 9.4+/-3.2 for Cercon Base/Cercon Ceram S. After thermocycling mean SBS values of 11.5+/-1.7 MPa for DC-Zirkon/IPS e.max Ceram, 9.7+/-4.2 MPa for Vita In-Ceram YZ Cubes/Vita VM9, and 9.6+/-4.2 MPa for Cercon Base/Cercon Ceram S were observed. Neither the differences between the SBS values of the all-ceramic test groups nor the influence of thermocycling on all groups were statistically significant. Irrespective of thermocycling the metal ceramic control group (27.6+/-12.1 MPa, 26.4+/-13.4 MPa) exhibited significantly higher mean SBS than all three all-ceramic groups tested. The all-ceramic groups showed combined failure modes as cohesive in the veneering ceramic and adhesive at the interface, whereas the metal ceramic group showed predominately cohesive fractures. The results indicated that the SBS between zirconia core and veneering ceramics was not affected by thermocycling. None of the zirconia core and veneering ceramics could attain the high bond strength values of the metal ceramic combination.[73]

ALL-CERAMIC CROWN FRACTURE

Laboratory failure and fracture load tests should ideally provide clinically comparable results. For test data to be relevant, lab tests should cause the same types of damage observed during clinical failure.[13]

Silva et al proposed that the mechanical performance of all-ceramic crown systems relative to that of metal ceramic restorations (MCR) has yet to be determined. This investigation tested the hypothesis that MCR present higher reliability over two Y-TZP all-ceramic crown systems under mouth-motion fatigue conditions. A CAD-based tooth preparation with the average dimensions of a mandibular first molar was used as a master die to fabricate all restorations. One 0.5-mm Pd-Ag and two Y-TZP system cores were veneered with 1.5 mm porcelain. After 60 days in water crowns were cemented onto composite (Z100, 3M/ESPE) reproductions of the die. Mouth-motion fatigue was performed, and use level probability Weibull curves were determined. Failure modes of all systems included chipping or fracture of the porcelain veneer initiating at the indentation site. Fatigue was an acceleration factor for all-ceramic systems, but not for the MCR system. The latter presented significantly higher reliability under mouth-motion cyclic mechanical testing.[74]

Coelho et al evaluated the mouth-motion step-stress fatigue behavior of two porcelain-zirconia all-ceramic crown systems. The average dimensions of a mandibular first molar crown were imported into CAD software; a tooth preparation was modeled by reducing proximal walls by 1.5 mm and occlusal surface by 2.0 mm. The CAD-based tooth preparation was made by rapid prototyping and used as a master die to fabricate all-ceramic crowns with 1.0 mm porcelain veneered on 0.5 mm Y-TZP cores (LAVA veneer and LAVA frame, 3M/ESPE, and Vita veneer and CERCON frame, Dentsply). Crowns were cemented on aged (60 days in water) composite (Z100, 3M/ESPE) reproductions of the die. Three crowns from the LAVA group were subjected to single cycle load to failure for stress profile design; remainder subjected to step-stress mouth-motion fatigue (three step-stress profiles). All mechanical testing was performed by sliding a WC indenter of 6.25 mm diameter 0.7 mm lingually down the mesio-distal cusp. Master Weibull curves and reliability for missions of 50,000 cycles at 200 N load were calculated (Alta Pro 7, Reliasoft). Single load to failure showed fractures through the zirconia core. Reliability for a 200 N x 50K cycle mission was not significantly different between systems. In fatigue, failure occurred by formation of large chips within the veneer originating from the contact area without core exposure. It was concluded that LAVA and CERCON ceramic systems present similar fatigue behavior; fatigue loading of both systems reproduces clinically observed failure modes.[75]

Coelho et al also found that clinically, zirconia-supported all-ceramic restorations were failing by veneer-chipping without exposing the zirconia interface. It was also hypothesized that mouth motion step-stress-accelerated fatigue testing of standardized dental crowns would permit this previously unrecognized failure mode to be investigated. Using CAD software, the average dimensions of a mandibular first molar crown tooth preparation were imported and modeled. The CAD-based tooth preparation was rapid-prototyped as a die for fabrication of zirconia core porcelain-veneered crowns. Crowns were bonded to aged composite reproductions of the preparation and stored for 14 days in water. Crowns were single-cycle-loaded to failure or mouth-motion step-stress- fatigue-tested. Finite element analysis indicated high stress levels below the load and at margins which modeled only single-cycle fracture origins. As hypothesized, the mouth motion sliding contact fatigue resulted in veneer chipping, reproducing clinical findings. It was concluded that this allows for investigations into the underlying causes of such failures.[76]

Bonafante et al compared the reliability of the disto-facial (DF) and mesio-lingual (ML) cusps of an anatomically correct zirconia (Y-TZP) crown system. The research hypotheses tested were: 1) fatigue reliability and failure mode are similar for the ML and DF cusps; 2) failure mode of one cusp does not affect the failure of the other. The average dimensions of a mandibular first molar crown were imported into CAD software; a tooth preparation was modeled by 1.5mm marginal high reduction of proximal walls and occlusal surface by 2.0mm. The CAD-based tooth preparation was milled and used as a die to fabricate crowns (n=14) with porcelain veneer on a 0.5mm Y-TZP core. Crowns were cemented on composite reproductions of the tooth preparation. The crowns were step-stress mouth motion fatigued (for 0.7mm) by sliding a tungsten-carbide indenter of 6.25mm diameter down on the inclines of either the DF or ML cusps. Use level probability Weibull curve with use stress of 200N and the reliability for completion of a mission of 50,000 cycles at 200N load were calculated. Reliability for a 200N at 50,000 cycles mission was not found to be different between tested cusps. SEM imaging showed large cohesive failures within the veneer for the ML and smaller for the DF. Fractures originated from the contact area regardless of the cusp loaded. It was concluded that no significant difference on fatigue reliability was observed between the DF compared to the ML cusp. Fracture of one cusp did not affect the other.[77]

Bonfante et al compared the reliability and fracture patterns of zirconia cores veneered with pressable porcelain submitted to either axial or off-axis sliding contact fatigue. Forty-two Y-TZP plates were veneered with pressable porcelain and adhesively luted to water aged composite resin blocks. This was stored in water for at least 7 days prior to testing. Profiles for step-stress fatigue (ratio 3:2:1) were determined from single load to fracture tests (n=3). Fatigue loading was performed either axially (n=18) or at 30° off-axis (n=18) to simulate posterior tooth cusp inclination creating and 0.7 mm slide. Single load and fatigue tests utilized a 6.25mm diameter WC indenter. Specimens were inspected by means of polarized-light microscope and SEM. Use level probability Weibull curves were plotted with 2-sided 90% confidence bounds (CB) and reliability for missions of 50,000 cycles at 200N (90% CB) were calculated. The calculated Weibull Beta was 3.34 and 2.47 for axial and off-axis groups, respectively, indicating that fatigue accelerated failure in both loading modes. No difference between loading groups were detected. Deep penetrating cone cracks reaching the core-veneer interface were observed in both groups. Partial cones due to the sliding component were observed along with the cone cracking for the off-axis group. No Y-TZP core fractures were observed. This study concluded that reliability was not significantly different between axial and off-axis mouth-motion fatigued veneer pressed over Y-TZP cores.[78]

Van Der Zel et al evaluate the effect of shoulder design on the failure load of Press-to-Cercon (PTC) crowns. Overpressed crowns with a zirconia free PTC shoulder (CS) and overpressed crowns with zirconia up to the margin (CC) were tested. 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). The thickness was 0.6 mm standard. After milling, the copings were sintered at 1350°C to final density. After sintering the coping was waxed-up to a standard contour, sprued and invested in CarraraUniversalDustless Investment (Elephant). The PTC Ceramic was pressed at 940°C over the zirconia coping. After devesting and separation from the sprues the crowns were veneered with two layers CerconCeramS porcelain (DeguDent). The crowns were cemented on a CoCr die with zinc phosphate cement and held under constant load of 5 kg during setting. The crowns were inspected using SEM for surface fracture analysis. Failure loads were measured using vertical compression loading at 0.5 mm/min. The results showed failure loads [kN (SD)]: Group CS (overpressed crowns with a zirconia free pressed-to-cercon shoulder): 4228(515) and group CC(overpressed crowns with zirconia up to the margin): 5408(806). This study significantly ($p<0.05$) concluded that a decrease of 22 percent in breaking strength was apparent with the overhanging shoulder as compared to fully supported PTC Crowns. Surface fracture analysis revealed the crack initiation site was typically located on the inside of the coping at the glass-zirconia interface. [79]

Curtis et al examine the influence of simulated masticatory loading regimes, to which all-ceramic crown or bridge restorations will routinely be subjected during their service-life, on the performance of yttria-stabilised tetragonal zirconia polycrystalline (Y-TZP) dental ceramic. Ten sets of 30 Y-TZP ceramic discs (13 mm diameter, 1.48-1.54 mm thickness) supplied by the manufacturer were randomly selected. Six groups were loaded for 2000 cycles at 500 N (383-420 MPa), 700N (536-588 MPa) and 800 N (613-672 MPa) with three groups maintained dry and the remaining three groups loaded while immersed in water at 37+/-1 degrees C. A further two groups underwent extended simulated masticatory loading regimes at 80 N (61-67 MPa) for 10(4) and 10(5)cycles under dry conditions. The mean bi-axial flexure strengths, standard deviations and associated Weibull moduli (m) were determined. The surface hardness was also determined using the Vickers hardness indentation technique. No significant difference (P>0.05) was identified in the bi-axial flexure strength of the simulated masticatory loading regimes and the control specimens loaded dry or wet. A significant increase in Weibull moduli (m) was identified for the Y-TZP specimens following loading while immersed in water (8.6+/-1.6, 8.5+/-1.6 and 10.3+/-1.9) compared with the control (7.1+/-1.3). However, the extended loading regime to 10(5)cycles resulted in a significant reduction in the Weibull modulus (m) of the Y-TZP specimens (5.3+/-1.0) compared with the control. Localized areas of increased surface hardness were identified to occur directly beneath the spherical indenter. It was concluded that the occurrence of localized areas of increased surface hardness could be the result of either a transformation toughening mechanism or crushing and densification of the material beneath the indenter manifested as the formation of a surface layer of compressive stresses that counteracted the tensile field generated at the tip of a propagating crack which increased the Weibull modulus of the Y-TZP specimens. The reduced reliability of the Y-TZP specimens loaded to 80 N for

10(5)cycles was associated with the accumulation of subcritical damage as a result of the extended nature of loading. [80]

MARGINAL ADAPTATION

Studies on marginal discrepancies of single restorations using various systems and materials have resulted in statistical inferences that are ambiguous because of small sample sizes and limited numbers of measurements per specimen.[51]

Holden et al compared the marginal adaptation of a pressed ceramic material when used with and without a metal substructure, to a traditional feldspathic porcelain-fused-to-metal restoration with a porcelain butt margin. A maxillary central incisor typodont tooth was prepared with a 1.5 mm 360 degrees shoulder with rounded internal line angle, and 30 polyether impressions were made. Dies were poured in type IV dental stone, and 30 restorations were fabricated: 10 metal ceramic restorations (MCR) with porcelain butt joints, 10 pressed to metal restorations (PTM), and 10 all-ceramic restorations (PCR). All restorations were evaluated on their respective dies at 45x magnification using an Olympus SZX-12, measurements of the marginal openings were made, and ANOVA and Scheffé post hoc tests were used to evaluate the data. The mean marginal opening found was 72.2 +/- 5.9 microns for MCR, 49.0 +/- 5.9 microns for PTM, and 55.8 +/- 5.9 microns for PCR. The post hoc tests showed that there was a statistical difference between the marginal adaptation of the PTM and MCR groups ($p < 0.05$). There was no significant difference in marginal adaptation between the PTM and the PCR groups, or the PCR and the MCR groups. It was concluded that the PTM group demonstrated a smaller mean marginal opening than the MCR group. The mean marginal openings of all three groups were within a clinically acceptable range.[81]

Goldin et al focused on the problem that fabricating a feldspathic porcelain margin on a metal-ceramic restoration with a clinically acceptable marginal fit is a technique-sensitive procedure. Pressable ceramics are advocated to solve this problem. The purpose of this in vitro study was to compare the marginal adaptation of a pressable ceramic system when used with both all-ceramic and metal-ceramic crowns, with a traditional metal-ceramic restoration. A 1.5-mm, 360-degree chamfer margin was prepared on a typodont maxillary central incisor. Polyether impressions were made and poured in a Type IV dental stone, and the following crowns were fabricated on individual dies: 15 metal ceramic restorations (MCR) (Ceramco II, Ceramco, and Argelite 60), 15 pressed-to-metal restorations (PTM) (CPC-MK, and Argelite 60), and 15 pressed ceramic restorations (PCR) (CPC-MK). The marginal fit of the crowns was evaluated every 90 degrees around the crown margin circumference under a microscope at original magnification x 45. A 1-way analysis of variance (ANOVA) was used to compare data (alpha=.05). The mean marginal discrepancy found for MCRs was 94 +/- 41 microns, for PTMs, 88 +/- 29 microns, and for PCRs, 81 +/- 25 microns. The 1-way ANOVA showed no significant difference between groups (P =.568). It was concluded that the marginal fit of pressed-to-metal (PTMs) and pressed all-ceramic crowns (PCRs) was similar to that of traditional metal-ceramic crowns.[82]

Yeo et al compared the marginal adaptation of single anterior restorations made using different ceramic systems. 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. Crown measurements were recorded with an optical microscope, with an accuracy of +/-0.1 μm, at 50 points spaced approximately 400 μm along the circumferential margin. The criterion of 120 μm was used as the maximum clinically acceptable marginal gap. Mean gap dimensions and standard deviations were calculated for marginal opening. The data were analyzed with a 1-way analysis of variance (alpha=.05). The mean gap dimensions and standard deviations at the marginal opening for the incisor crowns were 87 +/- 34 microm 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. Significant differences were found among the crown groups (P<.05). 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. Within the limitations of this study, it was concluded that marginal discrepancies were all within the clinically acceptable standard set at 120 μm. However, the IPS Empress 2 system showed the smallest and most homogeneous gap dimension, whereas the conventional In-Ceram system presented the largest and more variable gap dimension compared with the metal ceramic (control) restoration.[51]

Balkaya et al addresses the lack of information about how the fit is affected by fabrication procedures by examining the effect of porcelain and glaze firing cycles on the fit of 3 types of all-ceramic crowns. Ten standardized all-ceramic crowns were fabricated on a metal die from each of 3 systems: conventional In-Ceram, copy-milled In-Ceram, and copy-milled feldspathic crowns. Copings of the conventional and copy-milled In-Ceram crowns and nonglazed copy-milled feldspathic crowns served as the control. A device was used to apply a uniform load on specimens during measurement and to reposition the specimens on the measurement device after each manufacturing process. The specimens were not cemented and were measured on the metal die using a profile projector. Measurements were recorded at 18 points selected along horizontal and vertical planes. The crown systems were compared by use of the Student t test and 1-way analysis of variance (ANOVA). Data of measurements repeated at identical locations were analyzed with multivariate repeated-measures ANOVA. The Bonferroni post hoc test was used for multiple comparisons (alpha=.05). The conventional In-Ceram (57 +/- 24 μm) and copy-milled In-Ceram (57 +/- 32 μm) crowns demonstrated nearly identical marginal discrepancy values, followed by the copy-milled feldspathic crowns with a mean of 17 +/- 12 μm in the vertical plane. The copy-milled In-Ceram crowns had a mean horizontal discrepancy value of -12 +/- 4 μm, followed by the copy-milled feldspathic crowns with a mean of -4 +/- 5 μm and the conventional In-Ceram crowns with a mean of -6 +/- 4 μm. Statistical analyses demonstrated no significant differences in the marginal discrepancy values among the 3 all-ceramic crown systems, except for the horizontal discrepancy values between the conventional and copy-milled In-Ceram crowns after the porcelain firing cycle. Results indicated that the addition of porcelain to the copings caused a significant change ($P < .05$) in the marginal fit of the crowns, except for the fit in the horizontal plane of the conventional In-Ceram crowns. However, no significant changes occurred in the fit of the 3 all-ceramic crowns after the glaze firing cycle. There were significant differences in the marginal discrepancy values among the measurement locations ($P < .05$), and the discrepancy value at each location was independent of the mean of the entire crown. It was concluded that the 3 all-ceramic crown systems demonstrated a comparable and acceptable marginal fit. The porcelain firing cycle affected the marginal fit of the all-ceramic crowns. However, the glaze firing cycle had no significant effect on fit. The conventional and copy-milled In-Ceram crowns demonstrated medial deformations at the labial and palatal surfaces that might result in occlusal displacement of the crown.[83]

Vigolo et al performed an in vitro study to assess the marginal fit of four-unit fixed partial dentures (FPDs) produced using three different computer aided design/computer aided manufacturing (CAD/CAM) all-ceramic systems before and after porcelain firing cycles and after glaze cycles. An acrylic resin model of a maxillary arch was fabricated. Teeth #6 and 9 were prepared; teeth #7 and 8 were absent. Forty-five four-unit zirconium-oxide-based ceramic FPDs were made following conventional impression and master cast techniques: 15 were made with the Everest system, 15 with the Procera system, and 15 with the Lava system. Marginal gaps along vertical planes were measured for each bridge before (Time 0) and after (Time 1) porcelain firing cycles and after glaze cycles (Time 2) using a total of 8 landmarks (4 for tooth #6 and 4 for tooth #9) by means of a microscope at a magnification of x50. MANOVA was performed to determine whether the 8 landmarks, jointly considered, differed between CAD/CAM systems and time phases. Two-way ANOVA was performed to investigate in detail, for each landmark, how gap measurements were related to CAD/CAM systems and time phases. Differences were considered to be significant at $p < 0.05$. The mean values of the Everest system (in μm) were: 63.37 (Time 0), 65.34 (Time 1), and 65.49 (Time 2); the mean values of the Lava system (μm) were: 46.30 (Time 0), 46.79 (Time 1), and 47.28 (Time 2); the mean values of the Procera system (μm) were: 61.08 (Time 0), 62.46 (Time 1), and 63.46 (Time 2). MANOVA revealed quantitative differences of the 8 landmarks, jointly considered, between the three CAD/CAM systems ($p < 0.0001$), but it did not reveal any quantitative differences among the three time phases ($p > 0.4$). Two-way ANOVA revealed that the Lava system produced gap measurements statistically smaller than the Everest and Procera systems ($p < 0.0001$ for each landmark). Within the limitations of this study, it was concluded that the three zirconium-oxide-based ceramic CAD/CAM systems demonstrated a comparable and acceptable marginal fit. However, the Lava system produced gap measurements statistically smaller than the Everest and Procera systems. The porcelain firing cycles and the glaze cycles did not affect the marginal fit of the zirconium-oxide-based ceramic CAD/CAM systems.[84]

CEMENTATION OF CERAMIC RESTORATIONS

The longevity of an indirect restoration is closely related to the integrity of the cement at the margin. When compared to conventional dental ceramics, Y-TZP ceramics have smaller particle size and no glassy phase at the crystallite border[85]. For this reason, without a silica and glassy phase the effectiveness of conventional adhesive luting is impaired. The result has been the development and application of alternative surface treatments and materials with a chemical affinity for zirconia.[86-87]

Yin et al investigated the adequacy of luting cements when bonding zirconia ceramics to dentin. Blocks of sintered zirconia ceramics were randomly divided into 4 groups with 8 slices in each. Samples were immersed in saliva, airborne-particle abraded ceramic and cleaned with 35% phosphoric acid gel. After treatment, blocks were bonded to dentin with one of four luting cements. Specimens were stored in 37 °C distilled water for 24 hours then tested for shear bonding strength. The highest shear bond strength was obtained by Multilink Automix and 3M RelyX™ Unicem Aplicap™ cements. These values were both significantly higher than Tokuso

Ionomer and Shofu Polycarboxylate Cement groups. It was concluded that total etching resin luting cement is an ideal option to the bonding of zirconia ceramics for strong reliable bonding.[88]

Luthy et al evaluated the shear bond strength of different cements to densely sintered zirconia ceramic after aging by thermocycling. Luting cements evaluated for bonding to ZrO2-TZP were Ketac-Cem, Nexus, Rely X Unicem, Superbond C&B, Panavia F, and Panavia 21. Groups of 30 test specimens were prepared by bonding stainless steel cylinders tribochemically silica-coated with the Rocatec-system to sandblasted ZrO2-TZP ceramic disks. Prior to testing all bonded specimens were stored in distilled water at 37 °C for 48 hour and half of them (n=15) were additionally aged by thermocycling (10,000 times). It was found that one of the fractures occurred at the interface of the metallic rods. The assemblies failed either at the interface between the ceramic surface and the cements or within the cements. Thermocycling affected the bond strength of all luting cements studied except for both Panavia materials and Rely X Unicem. It was concluded within the limits of this in vitro study that after thermocycling, bond strengths for Ketac-Cem and Nexus were quite low. Nexus in combination with tribochemical silica-coating of ceramic surface produced superior bond strength. The four adhesive resin cements (Rely X Unicem, Superbond C&B, Panavia F, and Panavia 21) gave superior results. The strongest bond to zirconia was obtained with Panavia 21.[89]

Piwowarczyk et al studied the shear-bond strength of 11 luting cements from different material classes to manufactured pre-treated zirconia ceramics. Lava 97% ZrO_2 stabilized with 3% Y_2O_3 was used and the influence of the curing method on shear-bond strength was also tested. Zinc-phosphate cement (Fleck's zinc cement), two standard glass-ionomer cements (Fuji I, Ketac-Cem), three resin-modified glass-ionomer cements (Fuji Plus, Fuji Cem, RelyX Luting), four standard resin cements (RelyX ARC, Panavia F, Variolink II, Compolute) and one self-adhesive universal resin cement (RelyX Unicem) were all used in this study. The ceramic surface was sand-blasted with 100 μm alumina or tribochemically coated with silica. One group was tested after 30 minutes all other groups were stored in distilled water at 37 °C for 14 days then subsequently thermocycled 1000 times. After sandblasting, the highest shear-bond strength was obtained for the self-adhesive universal resin cement at 9.7 MPa (after 30 minutes) and 12.7 MPa (after 14 days and thermocycling), respectively. The highest values were found for one of the resin cements at 15.0 MPa (after 30 minutes) and for the self-adhesive universal resin cement at 19.9 MPa (after 14 days and thermocycling).[90]

Valandro et al evaluated the effect of two surface conditioning methods on the microtensile bond strength of resin cement to three high-strength core ceramics. The ceramics used were high alumina-based (In-Ceram Alumina, Procera AllCeram) and zirconia-reinforced alumina-based (In-Ceram Zirconia). Ten blocks (5 x 6 x 8 mm) of In-Ceram Alumina (AL), In-Ceram Zirconia (ZR), and Procera (PR) ceramics were fabricated according to their manufacturer's instructions and duplicated in composite. Specimens were either treated with airborne particle abrasion with 110 μm Al_2O_3 particles and silanated or silica coated with 30 μm SiO particles (CoJet, 3M ESPE) then silanated. Duplicates of ceramic blocks were made with composite resin (W3D-Master, Wilcos, Petrópolis, RJ, Brazil) using a silicon impression material mold. Composite resin layers were incrementally condensed to fill the mold and light polymerized for 40 s for each layer. Composite blocks were bonded to surface-conditioned ceramic blocks with resin cement (Panavia F, Kuraray, Okayama, Japan). One composite resin block was fabricated for each ceramic block. The ceramic-composite was stored at 37 °C in distilled water for 7 days prior to bond tests. The blocks were cut under water cooling to produce bar specimens (n = 30) with a bonding area of approximately 0.6 mm^2. The bond strength tests were then performed in a universal testing machine at crosshead speed of 1 mm/min. After statistical analysis silica coating with silanization was found to increase the bond strength significantly for all three high-strength ceramics (18.5 to 31.2 MPa) compared to that of airborne particle abrasion with 110 μm Al_2O_3 (12.7-17.3 MPa). Procera exhibited the lowest bond strengths after both Al_2O_3 and silica coating (12.7 and 18.5 MPa, respectively). It was concluded that conditioning the high-strength ceramic surfaces with silica coating and silanization provided higher bond strengths of the resin cement than with airborne particle abrasion with 110 μm Al_2O_3 and silanization.[91]

Ntala et al aimed to develop and test multi-phase glaze coatings for zirconia restorations, so that the surface could be etched and adhesively bonded. Zirconia disc specimens (n=125, 16 mm x 1 mm) were cut from cylinders of Y-TZP (yttria-stabilized tetragonal zirconia polycrystals) ZS-Blanks (Kavo, Everest) and sintered overnight. Specimens were subjected to the recommended firing cycles, and next sandblasted. The specimens were divided into 5 groups of 25, with Group 1 as the sandblasted control. Groups 2-5 were coated with overglaze materials (P25 and IPS e.max Ceram glazes) containing secondary phases. Group 2 was (wt%): 10% hydroxyapatite (HA)/P25 glaze, Group 3: 20% IPS Empress 2 glass-ceramic/glaze, Group 4: 20% IPS Empress 2 glass/glaze and Group 5: 30% IPS Empress 2 glass/glaze. After sintering and etching, Monobond-S and composite resin cylinders (Variolink II, Ivoclar-Vivadent) were applied and light cured on the test surfaces. Specimens were water stored for 7 days. Groups were tested using the shear bond strength (SBS) test at a crosshead speed of 0.5mm/min. Overglazed and the fractured specimen surfaces were viewed using secondary electron microscopy. Room and high temperature x-ray diffraction analysis (XRD) and dynamic scanning calorimetry (DSC) were carried out to characterize the materials. The mean (SD) SBS (MPa) of the test groups were: Group 1: 7.7 (3.2); Group 2: 5.6 (1.7); Group 3: 11.0 (3.0); Group 4: 8.8 (2.6) and Group 5: 9.1 (2.6). Group 3 was significantly different to the control Group 1 ($p<0.05$). There was no significant difference in the mean SBS values between Group 1 and Groups 2, 4 and 5 ($p>0.05$). Group 2 showed statistically lower SBS than Groups 3-5 ($p<0.05$). Lithium disilicate fibres were present in Groups 3-5 and fine scale fibres were grown in the glaze following a porcelain firing cycle (Groups 4 and 5). X-ray diffraction analysis (XRD) indicated a lithium disilicate/minor lithium orthophosphate phase (Group 3), and a tetragonal zirconia phase for the sintered Y-TZP ZS-Blanks. Dynamic scanning calorimetry (DSC) and high temperature x-ray diffraction analysis (XRD) confirmed the crystallization temperatures and phases for the IPS Empress 2 glass. It was concluded that the application of a novel glass-ceramic/glaze material containing a major lithium disilicate phase might be a step in improving the bond strength of a zirconia substrate to a resin cement.[92]

Toman et al examined the effect of different adhesive luting systems on the shear bond strength of all-ceramic restorations to dentin surfaces. Forty-eight all-ceramic disks (2 x 3 mm; IPS e.max Press) were fabricated. Forty-eight noncarious extracted human molars were divided into 4 groups. in groups 1 to 4, IPS e.max Press disks were luted with Variolink 2/Excite DSC (etch-and-rinse), Clearfil Esthetic Cement/Clearfil Protect Bond (antibacterial and self-etching), Multilink/Multilink Primer (self-etching), or Multilink Sprint (self-adhesive). All specimens were subjected to 5000 thermocycles (5 degrees C to 55 degrees C, 30-s dwell time). Shear bond strengths were tested using a universal testing machine until failure. The analysis of the fractured dentin surfaces was performed using an optical microscope at 10X and 1000X magnification; the images were analyzed with an image analyzer. Data were analyzed with one-way ANOVA and

Tukey's test at a significance level of $p < 0.05$. Mean shear bond strength data of the groups in MPa were: Variolink 2/Excite DSC: 25.89 +/- 3.71; Clearfil Esthetic Cement/Clearfil Protect Bond: 17.21 +/- 2.71; Multilink/Multilink Primer: 11.6 +/- 3.51; Multilink Sprint: 10.4 +/- 3.15. According to the one-way ANOVA, there were significant differences in shear bond strength ($p < 0.001$). According to Tukey's test, statistically significant differences were observed in shear bond strength between groups 1 and 2 ($p < 0.001$), groups 1 and 3 ($p < 0.001$), groups 1 and 4 ($p < 0.001$), groups 2 and 3 ($p = 0.003$), and groups 2 and 4 ($p < 0.001$). It was concluded the etch-and-rinse dentin bonding system produced higher bond strengths of all-ceramics to dentin surfaces than did the self-etching bonding systems and self-adhesive luting system.

Depending on the grain size of the cement and the preparation angle, crowns will be too high after cementing. The increase in height after cementation ranges from 210 µm for a cement with 25 µm particle grain size and 7° preparation to a 30 µm height increase for a 15 µm particle grain size cement with a preparation angle of 30°[64].

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CHAPTER II

1. ABSTRACT

The current manufacturer's recommendation for the thickness of zirconium copings ranges from 0.5-0.8 mm and must not be less than that of the overlying veneering porcelain. Full shoulder coverage of the coping extending to the margin of the tooth preparation is also required. STATEMENT OF PROBLEM: Full length coverage of the ceramic substructure to the margin of all-ceramic restorations contributes to an opaque and unnatural appearance at the cervical third of the restoration. Furthermore, 1.2 mm-wide shoulders with rounded inner edges can result in as little as 0.7 mm of residual dentin thickness in molars and premolars.[1] Other clinical limitations dictate significantly narrower shoulder widths in the mandible and incisors resulting in further variation from manufacturer's recommendations for all-ceramic margins.[1-2] When all these factors come together both esthetic and anatomic considerations must determine the application of all-ceramic margins and the need for further exploration of alterations in margin design becomes apparent. Unlike metal-ceramic restorations, design modifications of all-ceramic cores for esthetic improvement have not been investigated. PURPOSE: The aim of this study was to determine the effect of a modified coping margin design on fracture resistance and location of heat pressed Procera® All-Zircon crowns. MATERIALS AND METHODS: A total of 43 crowns were tested for strength and mode of fracture using one anatomical and one flat form of an all-ceramic upper premolar die for both modified core and control groups. The all-ceramic core design was equally divided into two groups. The first group of cores had full shoulder coverage extending to the margin of the tooth preparation following current guidelines and acted as a control. The second group of cores was modified to extend to the margin on the lingual,

mesial and distal but 3 mm above the margin on the facial of the restoration. Crowns were heat pressed with veneering porcelain to produce a total thickness of 1.5 mm axially and 2 mm occlusally. Final crowns were cemented onto resin dies and axially loaded. Maximal breaking force was recorded and mode of fracture was evaluated by direct 360° visual inspection of samples under a florescent light source held at 90° to the vertical axis of the crown. RESULTS: The mean maximum loads in Newtons for anatomical and flat all-ceramic crowns with full coverage zirconia cores were 1300.1±365.0 and 1572.2±140.8 respectively. The fracture loads were significantly less for both anatomical (1013.1±159.3) and flat crowns (1243.0±242.6) with the modified copings (Independent t-test, $p<0.05$). For both coping designs, the fracture loads were significantly greater with the flat preparation design (Independent t-test, $p<0.05$). Quantitative analysis of fracture modes revealed similar fracture patterns for both anatomical and flat crowns with the majority of fractures occurring in the veneering porcelain for all samples and no open core fractures in any of the crowns with full coverage cores. CONCLUSIONS: Modification of core design by cutting back its facial length decreases the overall maximum load resistance of the all-ceramic crown by 21-22%. Flat samples have on average 18% higher mean maximum loads than anatomically designed all-ceramic crowns. Regardless of core design, porcelain fracture occurs mainly within the veneering layer for all groups tested. CLINICAL RELEVANCE: This in-vitro study illustrates the importance of sample design in the production of clinically applicable data as well as provides clinicians with a basis to limit the extent to which zirconia cores can be modified without compromise to the strength of all-ceramic restorations.

2. INTRODUCTION

The translucency and color of dental restorations have been significantly improved by the development of all-ceramic systems. The natural translucent and reflection effect of ceramics has been the source of esthetic improvement of ceramic dental restorations. [3-4] When compared with metal alloys and dental composites, limitations to the use of ceramics for dental prostheses include brittleness which can compromise mechanical reliability and the greater effort and processing time they require.[5] The mechanical properties of dental ceramics have been improved by recent advances in ceramic processing methods which have simplified the work of the dental technician and allow greater quality control for ceramic materials. Initial concerns regarding the strength of all-ceramic restorations were alleviated with the advent of high strength core ceramics. However, patients with posterior tooth loss have a mean clenching force of 462 N and a maximum bite force of 1031 N and patients with full a dentition have a larger mean of 720 N with a maximum force of 1243 N.[6] Any improvement in the esthetics of dental restorations cannot compromise their strength to withstand these forces. Furthermore, core materials have been found to generally lack fluorescence. [7] Fluorescence, opalescence, and translucency are critical for restorative materials to mimic the optical properties and appearance of natural teeth. A modified layering technique has been documented to enhance fluorescence within the veneering ceramic and provides an esthetic appearance of glass-infiltrated aluminum oxide ceramic restorations.[8] However, modification in zirconia based substructures for esthetic improvement has not been investigated. Furthermore, clinicians should reserve dental ceramics with high translucency for clinical applications which require high-level esthetics and the restoration can be bonded to tooth structure.[9] The question arises when both strength and esthetics must be equally matched. Although ceramics with high strength can mask discoloration when present, they also tend to be more opaque and pose a challenge when trying to match natural tooth color. For this reason, knowledge of the optical properties of available ceramic systems enable the clinician to make appropriate choices when faced with various esthetic challenges.[9]

Strength, adaptation and esthetics are determined by the coping design, its crystalline structure and dimensions. No significant difference in fracture strength has been found in 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. [10] Also, Zirconia based ceramics have been found to have more structural reliability and superior mechanical properties. [11] The marginal fit of pressed-to-metal and pressed all-ceramic crowns has been found to be similar to the technique sensitive and traditional sintered metal-ceramic crowns.[12] However, there is limited data focused on the strength of heat-pressed ceramics onto zirconia cores. Furthermore, there is no literature which explores the effect of changes in the manufacture's currently recommended ceramic core guidelines.

3. PURPOSE

The aim of this in vitro study was to determine the effect of a modified coping margin design on the fracture resistance and location of heat pressed Procera® All-Zircon crowns. It was also to determine whether physical tests on regular geometrically designed restorations can reliably predict the behavior of anatomical restorations with irregular occlusal surfaces.

4. HYPOTHESES

Primary Hypothesis

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

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

Secondary Hypothesis

Ho: The coping design modification is not associated with fracture location of heat pressed Zirconia core ceramic crowns.

Ha: The coping design modification is associated with fracture location of heat pressed Zirconia core ceramic crowns.

Tertiary Hypothesis:

Ho: Physical tests on anatomical die preparations do not yield comparable results with regular geometric die preparations with flat occlusal surfaces.

Ha: Physical tests on anatomical die preparations yield comparable results with regular geometric die preparations with flat occlusal surfaces.

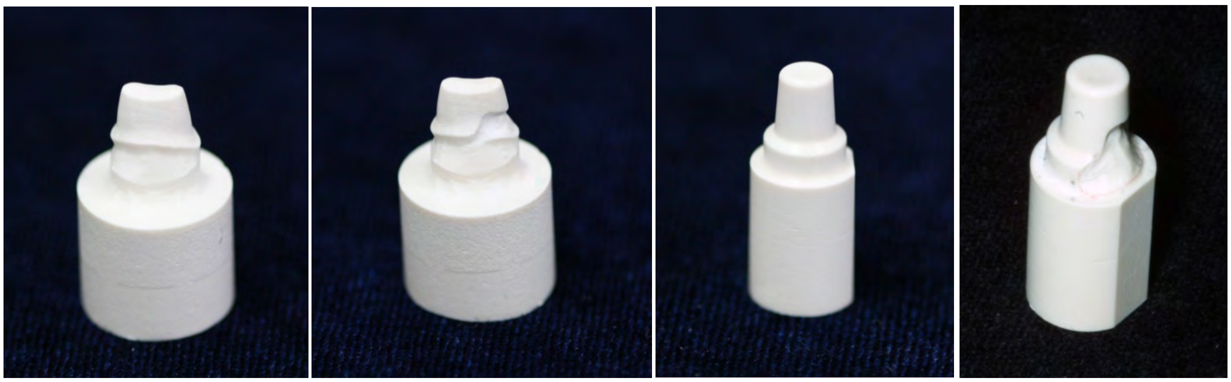
5. MATERIALS AND METHODS

5.1. Specimen Preparation

A high filler content epoxy resin with a modulus of elasticity of 12.9 MPa was used to fabricate anatomical and flat premolar dies (Viade Products Inc., Camarillo, CA USA). This resin simulated the physical properties of dentin which has a comparable modulus of elasticity equal to 14.7 MPa. [13] The dimensions of each die represented the contours of a maxillary premolar all-ceramic crown preparation with 1.0 mm deep chamfer, round internal line angles, 1.5 mm axial reduction, and 2.0 mm rounded occlusal reduction. In order to facilitate axial loading of samples at testing, the cylindrical die was trimmed to create an occlusal plane which is perpendicular to the long axis of the tooth. Similarly, the buccal cusp tip of the occlusal surface of the anatomic die was leveled to compliment the 1.0 mm flat surface of a testing probe.

Single anatomical and flat epoxy resin dies were replicated to produce two identical groups of corresponding anatomical and flat dies. Each group was further subdivided into test and control groups to produce a total of two control and two test groups (each with anatomical and flat forms). In order to scan and design copings for the test groups, the base of the epoxy resin die was modified. The modified dies were used only for the scanning process and not for bonding to final crowns. This modification was performed using a high speed handpiece to trim a horizontal undercut 3 mm above the cervical margin on the facial surface of the test dies. When scanned, this information corresponded to a coping margin on the facial aspect of the die which was 3 mm above the cervical margin of the crown.

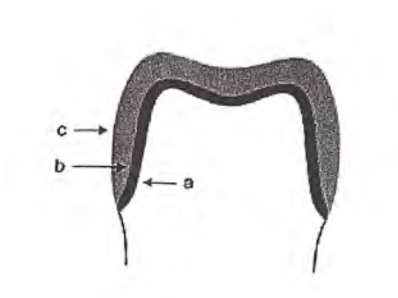
Four dies each corresponding to a single group was scanned using a Procera Scanner (Piccolo, Nobel Biocare AB, Goteborg, Sweden) with a 1.25 diameter scanning probe to produce a total of four groups. A single scan of one modified anatomical die was used to design a coping for the anatomical test group and another single scan of a modified flat die was used to produce its respective test group. Likewise, the original forms of single flat and anatomical dies were used to design copings for their respective control groups. This is shown in Figure 9.



a. Anatomical die b. Anatomical modified die c. Flat die d. Flat modified die.

Figure 9. Photographs of dies scanned and used to design full coverage and cut-back copings.

An electronic order was placed to the Procera production facility (Nobel Biocare, Mahwah, NJ, USA) for 12 identical zirconia copings for each of the four designs. The Zirconia coping of the control groups extended to the finish line (Figures 10 and 11) of the preparation whereas the coping of the test groups extended 3 mm short of the shoulder on the axial wall of the facial (Figures 12 and 13) and to the finish line of the preparation on lingual.



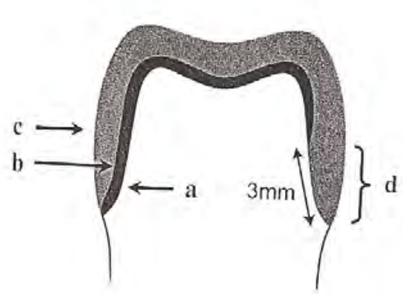
Key:
a – die preparation
b – zirconia coping
c – veneering porcelain

Figure 10. Schematic drawing of control groups of all-ceramic crowns with normal full length copings.

a.

b.

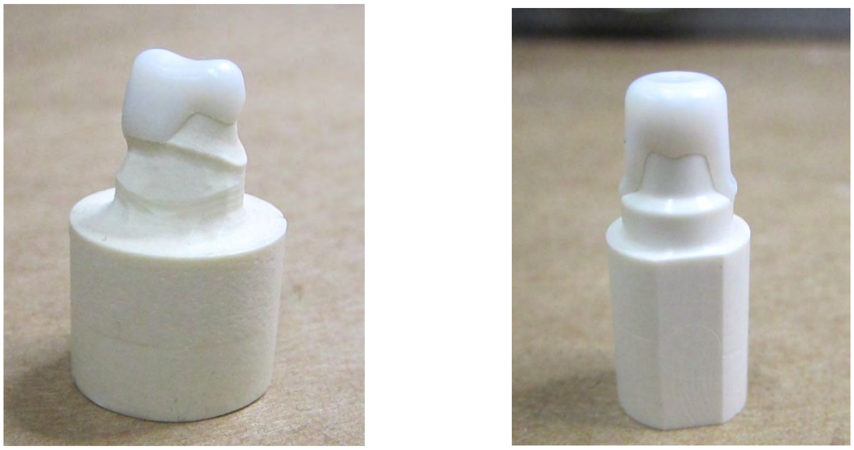
Figure 11. Copings used for control groups- a. Anatomical, b. Flat.



Key:

a – die preparation
b – modified zirconia coping
c – veneering porcelain

Figure 12. Schematic drawing of the test groups of all-ceramic crowns with cut-back copings.



a. b.

Figure 13. Copings used for cut-back groups- a. Anatomical, b. Flat.

A total of forty-eight ™All-Zircon copings for each group were presintered, milled and sintered to a thickness of 0.5-0.7 mm and measured at specific locations using a dial caliper (Kori Dial Caliper Pfingst & Company, Inc., Tokyo, Japan) as shown in the results in Tables 3-6. The copings were placed onto identical replicas of their original corresponding dies with die hardener lightly painted onto the margins. Copings were scanned using a Dental Wings 3D laser scanner and wax crowns were designed using an operating CAD/CAM system (Zahn Dental Laboratory Division, Henry Shein Company, USA – Figure 14 a). The process utilized a virtual wax-up design (Fairlane Dental, Livonia MI, USA) to produce an overall coping-wax thickness of 1 mm at the margin, 1.5 mm axially and 2 mm at the occlusal surface for both crown designs (Figures 14 b and c).

a. b. c.

Figure 14. a. Dental Wings 3D laser scanner,

b. Waxed anatomical crown, c. Waxed flat crown.

Each waxed crown was sealed at the margin with hot ash free wax and sprued with 6 mm length sprue wax (Figure 14 d). This was sprayed with debubbilizer before investing in Galaxy Universal Investment (Talladium, Inc., Valencia CA USA).

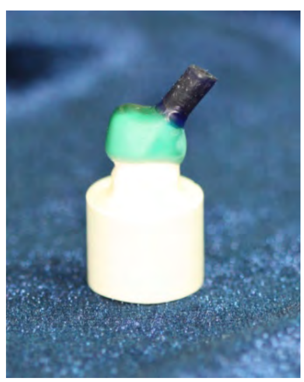


Figure 14. d. Waxed crown was sprued with 6 mm length sprue wax.

Investment material (Galaxy universal investment Batch # G 9097 – 2, Talladium, Inc. Valencia CA, USA) was mixed in a vacuum mixing unit at 600 rpm (Twister Evolution Pro, Renfert IL, USA - Figure 15), allowed to flow around waxed and sprued crowns encased within a cylindrical investment paper (Pentron ceramics universal 100 g investment paper, Somerset NJ, USA - Figure 16) and left to bench set for 15 minutes.

Figure 15. Twister Evolution Pro, vacuum mixing unit, Renfert Il, USA

Figure 16. Pentron Ceramics universal 100 g investment paper containing Galaxy universal investment material

After discarding the investment paper, each 100 g investment cylinder was placed in a burnout oven (Infinity L30 Programmable Multiphase Burnout Furnace, Jelrus International, Melville NY, USA - Figures 17 and 18) at 1562 °C for 45 minutes then directly transferred to the ceramic press oven.

Figure 17. Infinity L30 programmable multiphase burnout furnace, Jelrus International, Melville NY, USA.

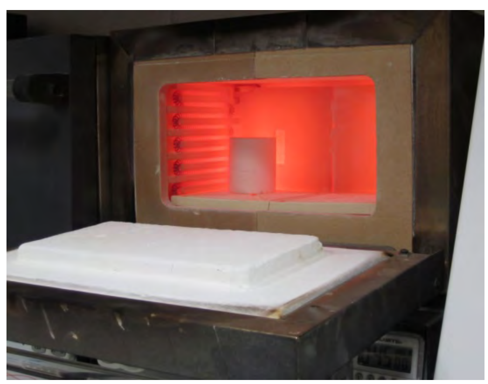


Figure 18. Investment cylinder placed into the burnout furnace.

GC Initial IQ – Press-over-Zircon overpressable feldspathic ceramic system was used to veneer all-ceramic crown control and test groups. These ceramic ingots were heat pressed onto the zirconia cores using a pressable ceramic disposable plunger (Vacalon, OH, USA – Figure 20) in a press oven (Empress High Temperature Injection Molding System, Ivoclar Inc. Amherst, NY, USA – Figure 19) at 4 bars of pressure for 35 minutes and peak temperature of 1562 °F. All-ceramic crowns were produced with a total veneer plus zirconia core axial thickness of 1.5 mm and an occlusal thickness of 2.0 mm. Crowns were devested by sandblasting (Pure Blast, Macro Cab, Danville Engineering Inc., USA) with glass beads at approximately 50 mm and 60 psi and carefully trimmed off their sprues with a diamond blade on a straight handpiece. Specimens were then steam cleaned (Steaman II, Bar Intruments, CA, USA) and air dried before cementation.

Figure 19. Empress high temperature injection molding system, Ivoclar Inc. Amherst, NY, USA

Figure 20. Investment cylinger set-up with a pressable ceramic disposable plunger system, Vacalon, OH, USA.

Measurements were performed using a dial caliper (Figure 21 a) and thickness adjustments were performed using brasseler porcelain polishing wheels (Figure 21 b).

a. b.

Figure 21. a. Dial caliper used to measure copings and final all-ceramic crowns,

b. Brasseler porcelain polishing wheels.

The buccal cusp tips of the anatomical crowns and the occlusal surface of the flat crowns were leveled with porcelain polishing wheels to facilitate loading with a 1 mm diameter probe in a universal testing machine (Figure 22). The total core and veneering porcelain thickness at the load point was 1.5 mm for the anatomical group and 2 mm for the conical group of specimens (Figure 23).

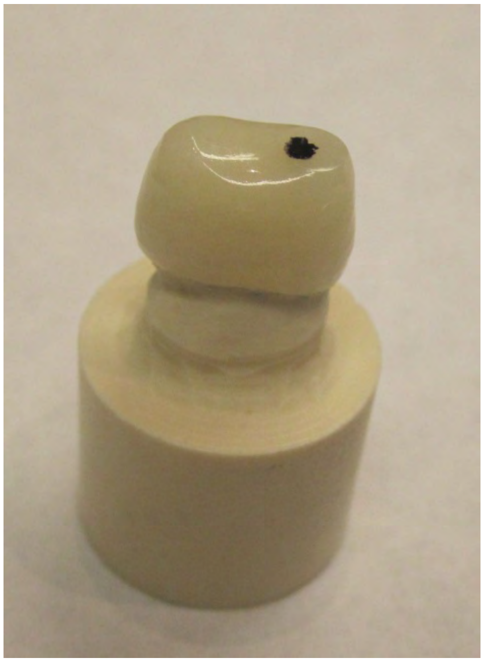


Figure 22. Anatomical all-ceramic crown specimen

Figure 23. Flat all-ceramic crown specimen

Neither porcelain firing cycles nor the glaze cycles have been found to affect the marginal fit of the zirconium-oxide-based ceramic CAD/CAM systems[14]. For this reason, all specimens were gradually heated up to 900 °F (below their original pressing temperature) over a 30 minute period then held at that temperature for 5 minutes in a ceramic oven (Whipmix Pro 100 Porcelain Oven, USA – Figure 24) for natural glazing.

Figure 24. Whipmix Pro 100 ceramic oven, USA.

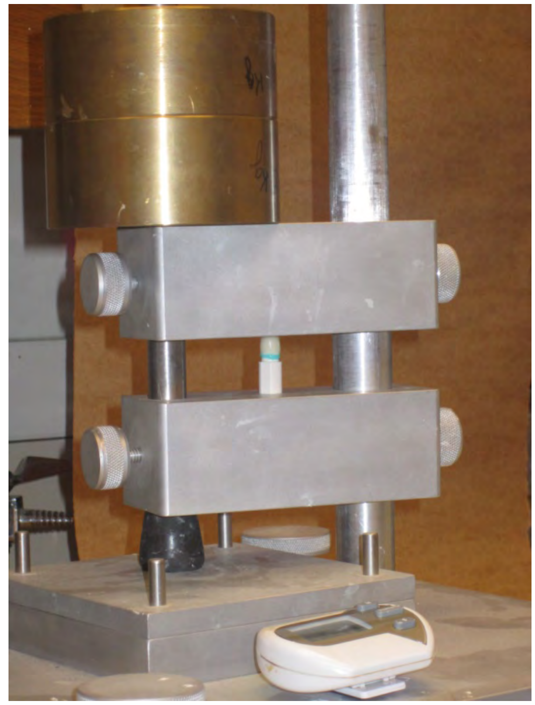


Figure 25. 5 kg Static loading jig.

Air abrasion at a distance of approximately 10 mm for 10 s using 50 µm Al_2O_3 at 80 psi (Pure Blast, Macro Cab, Danville Engineering Inc., USA) was used to roughen the internal surface of the zirconia core in preparation for bonding. This was cleaned with distilled water and air dried.

Total etching resin luting cement has been found to be an ideal option to the bonding of zirconia ceramics for strong reliable bonding[15]. Furthermore, the strongest bond to zirconia was obtained with Panavia 21 when bond strength has been studied against other luting cements such as Rely X Unicem, Ketac-Cem, Nexus , Superbond C&B and Panavia F[16]. Although, conditioning the high-strength ceramic surfaces with silica coating and silanization provides higher bond strengths of the resin cement than with airborne particle abrasion with 110 µm Al_2O_3 and silanization[17] other studies suggest resin cements increase bond strengths of air abraded surfaces[16]. Internal surfaces were primed, etched with 40% phosphoric acid and silanated followed by bonding with Panavia 21. Luting was performed with firm finger pressure then placement under a 5 kg static load while excess cement was removed (Figure 25).

5.2. Measurement of Fracture Load and Location

Specimens were stored in 100% humidity at 37 °C for 24 hours. Specimens were mounted and stabilized in a metal device and a 1.0 mm diameter stainless steel rod was placed 2 mm from the external edge over the occlusal surface of each specimen (Figure 26). An Instron universal testing machine (Model 5560, Instron Corp., Canton MA, USA- Figure 27) was used to load specimens axially and record the maximum fracture loads. Specimen segments were examined for analysis of fracture mode. Statistical analyses were performed to determine significance between the different coping designs.

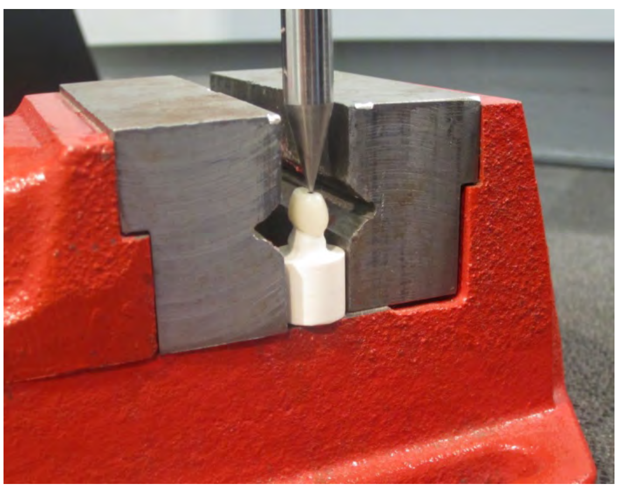


Figure 26. Metal loading device & 1.0 mm diameter stainless steel rod 2 mm from external edge of occlusal surface.

Figure 27. Instron universal testing machine model 5560, Instron Corp., Canton MA, USA.

6. SUMMARY OF STATISTICAL ANAYLSIS

Means and standard deviations were calculated for each group tested and the mean fracture strength values were analyzed using independent student t-tests to determine the effect of cutting back the zirconia core with either occlusal design. Statistical analysis was carried out using statistical software (SYSTAT Software Inc., Evanston, IL, USA) at $\alpha=0.05$. Appendix I shows a detailed statistical analysis of mean maximum and fracture loads.

7. RESULTS

All results refer to the following groups:

GROUP 1 - anatomical crown with cut-back core
GROUP 2 – flat crown with cut-back core
GROUP 3- anatomical crown with normal core
GROUP 4 – flat crown with normal core

CORE AND RESTORATION THICKNESS

Appendix II shows the zirconia core, waxed crowns and final core plus veneering porcelain thickness values recorded in mm using a dial caliper. One reading was taken for each surface (mid-buccal, mid-lingual, mid-mesial, mid-distal and occlusal center) of each group of crowns. The total average core thickness was 0.6 ± 0.1 mm and the total average crown thickness was 1.5 ± 0.1 mm axially and 2.0 ± 0.1 mm occlusally.

MEAN MAXIMUM AND FRACTURE LOADS

The mean maximum load value represents the highest load recorded by the universal testing machine during the full length of the loading process. The fracture load is usually obtained shortly after the maximum load recording and represents the final load recorded before catastrophic crown failure. Both mean maximum and fracture loads obtained in this study were higher for flat restorations with both normal cores and cut-back core designs. The highest overall mean maximum and fracture load values were reflected by the normal core design group. The lowest overall mean maximum and fracture loads were observed in the anatomically designed crown with a cut-back core design. Mean fracture loads were an average of 36 N less than mean maximum loads recorded. The mean maximum loads in Newtons for anatomical and flat all-ceramic crowns with full coverage zirconia cores were 1300.1±365.0 and 1572.2±140.8 respectively. The fracture loads were significantly less for both anatomical (1013.1±159.3) and flat crowns (1243.0±242.6) with the modified copings (Independent t-test, $p<0.05$). For both coping designs, the fracture loads were significantly greater with the flat preparation design (Independent t-test, $p<0.05$).

Table 3. Summary of the means and standard deviations of fracture and maximum compression loads for groups 1, 2, 3 and 4.

Group Number (# of samples)	Design	Mean Max Load/N	Mean Fx Load/N
1 (10 samples)	Anat. Crn / Cutback Core	1013.1 ± 159.3	980.4 ± 177.5
2 (11 samples)	Flat Crn / Cutback Core	1243.0 ± 242.6	1191.5 ± 253.2
3 (11 samples)	Anat. Crn /Normal Core	1300.1± 365.0	1271.9 ± 369.0
4 (11 samples)	Flat Crn / Normal Core	1572.2 ± 140.8	1542.7 ± 137.6

Table 4. Fracture and maximum compression loads for all samples in group 1.

Cutback/Anatomical Crown Sample No.	Fx. Load /N	Max. Load /N
1	1132.18	1147.18
2	1112.86	1145.22
3	623.11	733.64
4	892.40	912.31
5	918.39	950.66
6	1031.66	1049.60
7	1015.18	1040.49
8	1214.94	1240.05
9	1073.24	1098.64
10	790.32	813.27
Mean	980.4	1013.11
Standard Deviation	177.5	159.31

Table 5. Fracture and maximum compression loads for all samples in group 2.

Cutback/Conical Crown Sample No.	Fx. Load /N	Max. Load /N
1	905.94	922.41
2	1205.73	1236.24
3	1425.79	1440.79
4	1191.12	1209.85
5	1082.46	1095.50
6	977.33	1356.65
7	1062.65	1076.77
8	1028.62	1065.39
9	1666.54	1666.54
10	995.28	1013.52
11	1565.14	1589.17
Mean	1191.5	1242.99
Standard Deviation	253.2	242.65

Table 6. Fracture and maximum loads for all samples in group 3.

Control/Anatomical Crown Sample No.	Fx. Load /N	Max. Load /N
1	834.45	864.55
2	1328.21	1397.15
3	902.68	830.33
4	1410.49	1425.00
5	1016.07	1045.88
6	1836.59	1836.59
7	1579.16	1621.43
8	1157.87	1188.86
9	877.99	891.03
10	1350.87	1389.60
11	1795.79	1810.60
Mean	1271.9	1300.09
Standard Deviation	369.0	365.98

Table 7. Fracture and maximum loads for all samples in group 4.

Control/Conical Crown Sample No.	Fx. Load /N	Max. Load /N
1	1552.78	1618.59
2	1285.65	1301.40
3	1600.44	1645.36
4	1550.82	1565.93
5	1553.77	1567.00
6	1545.23	1576.42
7	1661.54	1704.89
8	1287.12	1316.74
9	1712.73	1729.30
10	1575.63	1601.82
11	1643.69	1666.44
Mean	1542.7	1572.17
Standard Deviation	137.6	140.79

MODE OF FRACTURE

Classifications for fracture mode for each restoration were formulated and applied in four different ways. The first classified the mode of fracture for the entire restoration and is based on the fact that the testing probe could not fracture underlying layers without first penetrating overlying layers. The order of penetration of the probe starts most superficially with the veneering porcelain of the bilayered ceramic, followed by the zirconia core, which is in turn cemented to a die (the final underlying layer of each specimen). More specifically, any fracture of the core cannot occur without fracture of the veneering porcelain and specimens categorized only in the veneering layer did not display any core fracture.

The term open fracture refers to a fracture in which a piece of the sample is lost and results in a void or negative space within a specified layer. Alternatively, the term closed fracture refers to a hairline fracture or groove in which the mass of the specified layer remains present but is no longer attached following loading to fracture. The numbers of specimens which fall into each category of overall fracture mode are listed in Table 8. This shows that the majority of fractures occurred within the veneering layer of the bonded all-ceramic, pressed to zirconia restorations. Furthermore, between both groups with a normal zirconia core design, core fracture occurred only in one specimen. This was also the only design in which die fracture was seen. The groups with cut-back cores displayed the most variable fracture modes distributed between the veneering porcelain as well as both open and closed core fractures. No correlation was found between mode of failure and maximum fracture loads.

Table 8. Overall fracture mode rating for each all-ceramic restoration.

Rating	Open Veneer Fracture	Closed Core Fracture	Open Core Fracture	Die Fracture	Total Number of Samples
Group 1	3	4	3	0	10
Group 2	2	4	5	0	11
Group 3	9	1	0	1	11
Group 4	10	0	0	1	11

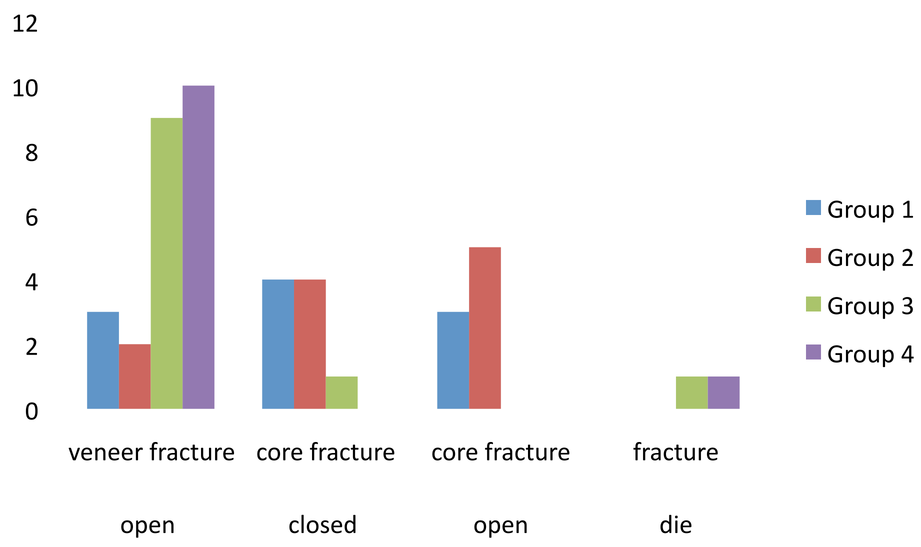


Figure 28. Overall fracture mode rating for each all-ceramic restoration.

The three other classifications for fracture mode were specific for each layer of the bonded restoration. Subcategories were developed to reflect the extent of fracture within each layer. The extent of fracture was rated from 0-4 for both veneering porcelain and zirconia core layers and 0-3 for the die. The rating system for each category is listed in the following table.

Table 9. Fracture mode classification within each layer of the bonded restoration.

Porcelain (P)	Coping (C)	Die (D)
0- no evidence of fracture	0- no evidence of fracture	0- no evidence of fracture
1-fracture line	1-crack line (closed)	1-horizontal
2-chip not extending ≥1/2 crown	2-fracture line (open)	2-vertical fracture
3-chip extending ≥1/2 crown	3-chip	3-combination fracture
4- chip infringing on the shoulder		

The highest overall frequency of failure mode within the pressed porcelain layer was chipping extending to the shoulder of the restoration for the cut-back groups tested and a less than 50% chip for the crowns with normal cores. None of the specimens exhibited a lack of fracture or hairline fractures of the pressed veneer. This is shown in Table 10 and illustrated in Figure 29 below.

Table 10. Fracture mode rating within the pressed veneering porcelain layer of the bonded restoration.

Rating	0 No Fracture	1 Crack Line Closed	2 Chip < 50%	3 Chip > 50%	4 Chip to Shoulder	Total # of Samples
Design 1	0	0	2	0	8	10
Design 2	0	0	1	0	10	11
Design 3	0	0	8	3	0	11
Design 4	0	0	9	2	0	11

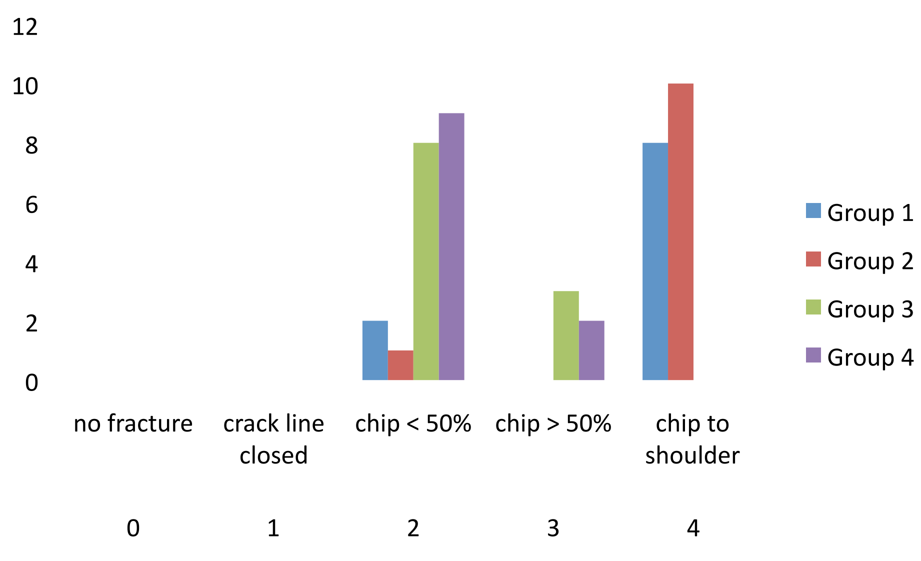


Figure 29. Fracture mode rating within the pressed veneering porcelain layer of the restoration.

The highest frequency for failure mode within the zirconia core layer was an absence of fracture altogether. When fracture of the core occurred, it was either a hairline fracture or a large chip of core which broke off the specimen and did not involve any open fracture lines within the core itself. However, chipping only occurred in the cut-back designs and not in the normal core specimens. This is shown in Table 11 and illustrated in Figure 30.

Table 11. Fracture mode rating within the zirconia core layer.

Rating	0 No Fracture	1 Crack Line Closed	2 Fracture Line Open	3 Chip	Total Number Of Samples
Design 1	3	4	0	3	10
Design 2	1	5	0	5	11
Design 3	10	1	0	0	11
Design 4	11	0	0	0	11

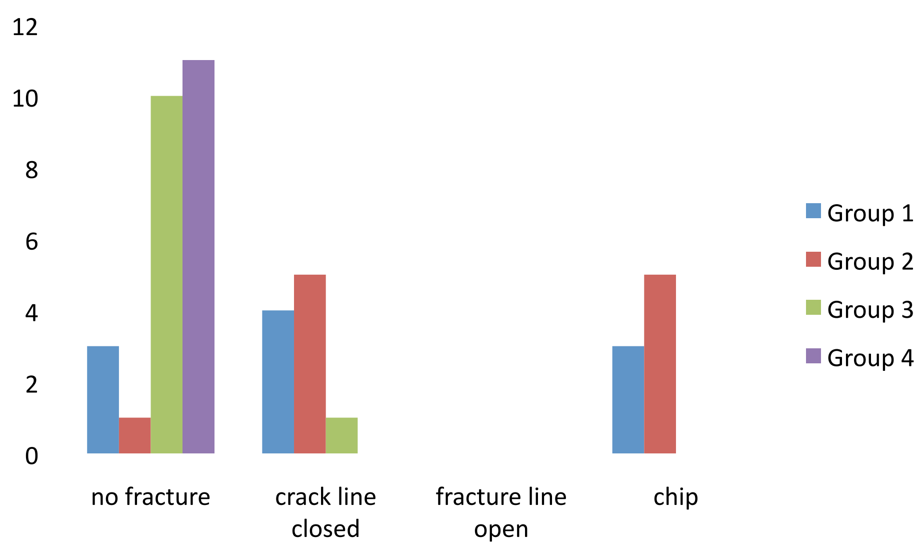


Figure 30. Fracture mode rating within the zirconia core layer.

There was no die fracture for the majority of specimens. Only two dies fractured and this occurred only in the groups of restorations with full length cores. This is shown in Table 12 and illustrated in Figure 31 below.

Table 12. Fracture mode rating within the die of the bonded restoration.

Rating	0 No Fracture	1 Horizontal	2 Vertical	3 Both	Total Number of Samples
Design 1	9	1	0	0	10
Design 2	11	0	0	0	11
Design 3	10	0	0	1	11
Design 4	10	0	0	1	11

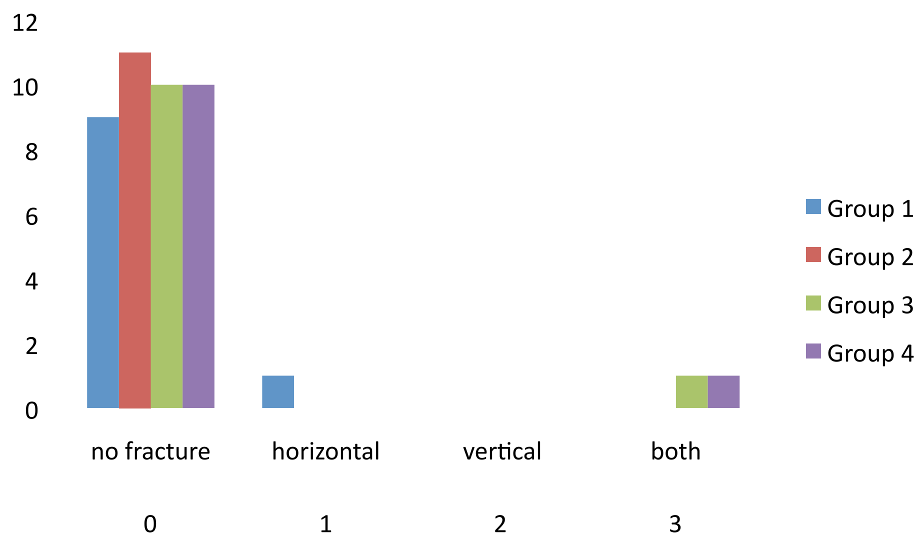


Figure 17. Fracture mode rating within the die of the bonded restoration.

8. DISCUSSION

The aim of this in vitro study was to determine the effect of a modified coping margin design on fracture resistance and location of heat pressed Procera® All-Zircon crowns. The importance of obtaining clinically applicable data was addressed by using an anatomical crown design for comparison against the flat crown design traditionally used in physical crown fracture tests. The goal of the substructure modification was to improve the natural appearance of the crown since core materials have been found to generally lack fluorescence.[7] However, improvements in the esthetics of dental restorations should not compromise their strength to withstand maximum bite forces which are around 1031 N for the partially dentate and 1243 N in patients with full a dentition.[6]

Pressable ceramics usually have application only as core and framework materials. Pressable veneering materials, such as IPS e.max, ZirPress (Ivoclar-Vivadent) are available, but the depth of layering for esthetics may be limited when using pressable ceramics for veneering purposes.[18] Numerous techniques have been utilized to improve the natural appearance of pressed ceramics. A modified layering technique has been found to enhance fluorescence within the veneering ceramic and provide a better esthetic appearance of glass-infiltrated aluminum oxide ceramic restorations.[8] However, knowledge of the optical properties of available ceramic systems can enable the clinician to make appropriate choices when faced with various esthetic challenges.[9] Heffernan determined a range of core translucencies at clinically relevant core thicknesses with dentin as a standard. In increasing order this was found to be; In-Ceram Spinell, Empress, Procera, Empress 2, In-Ceram Alumina, In-Ceram Zirconia then lastly, metal alloy.[3]

A study by Rapti showed In-Ceram Spinell to allow better transmission than PFM restorations and IPS Empress to possess superior light transmission.[19] The glazing cycle has also been shown to decrease opacity for ceramic cores except the completely opaque In-Ceram Zirconia and metal-ceramic specimens.[4] This study employed a core design modification to zirconia which essentially replaced the more opaque ceramic core at the buccal shoulder of the restoration with the more transparent veneering porcelain. Glass ceramic crowns investigated by Friedlander and Doyle exhibited highest fracture resistance for a preparation design with a total convergence angle of 10°, 1.2 mm shoulder finish line and sharp axiogingival line angle. All-ceramic single tooth restorations with a minimum thickness of 1.6 mm and a load bearing capacity of 2000 N were found to be a suitable restoration. Whereas any decrease in minimal thickness was associated with the choice of ceramic capable of withstanding higher loads.[20] This justifies the shoulder preparation modification in this study in which the veneering ceramic at the buccal shoulder had an axial thickness of 1.5 mm to maximize fracture resistance. Although average maximum loads obtained in this study were smaller for modified cores, these loads remained within the range of average load for natural dentition. Further studies need to be performed to address the effect of the core cut-back in applications where there is a narrower shoulder width available for preparation. This will conclude the extent to which the all-ceramic shoulder can be modified without compromise to strengths comparable with the natural dentition.

Studies have indicated that the shear bond strength between zirconia core and veneering ceramics are not affected by thermocycling. [21] Furthermore, the marginal fit of pressed-to-metal (PTMs) and pressed all-ceramic crowns (PCRs) is similar to that of traditional metal-ceramic crowns.[12] Holden et al also found no significant difference in marginal adaptation between PTMs and PCRs. [22] However, for adequate strength and rigidity, a thickness of 0.3-0.5 mm of metal substructure is required.[23] This systematic review of the literature on the physical properties of all-ceramic restorations justifies the choice of all-ceramic restorations over metal-ceramics restorations. The results of this present study are in agreement with current literature. The maximum fracture load attained in this study for a pressed to zirconia unaltered core was 1729 N. Therefore the argument for replacing pressed-to-metal crowns (PFMs) with more esthetic all-ceramic crowns (PTCs) is not only valid for improvement of core esthetics but also for adequacy of strength.

Researchers and manufacturers have developed advanced formulas to prevent crack propagation mainly by using yttrium-tetragonal zirconia polycrystals (Y-TZP), commonly known as zirconia.[24-25] There is abundant evidence which has proven that zirconia-based ceramics are stronger and tougher than glass ceramics.[26] The zirconia core reinforces the overall strength and toughness of the restoration. In fact, the influences of these properties are so significant that it tolerates zirconia core modification within certain design parameters. The extent to which a zirconia core can be altered without compromising the integrity of the restoration has not been fully explored. In this study the fracture loads exceeded 1000 N by the zirconia core all-ceramic crown irrespective of its core design. This proves both the reliability of both conventional pressed-to-zirconia core ceramic restorations as well as those restorations with up to 3 mm of glass ceramic at the shoulder of one aspect of the restoration in place of the zirconia core.

Many clinicians have not established the trust of heat-pressed to zirconia restorations over traditional veneering. This opinion prevails despite the fact that the strength of traditional powder liquid restorations have been found to be comparable with heat pressed all-ceramic restorations. However, the higher crystalline content and lack of porosity of heat pressed all-ceramic restorations does not increase fracture resistance when compared with traditional fabrication techniques.[18] In fact, the benefits of pressing versus traditional veneering are less technique sensitivity for labs, consistency and consequently cost effectiveness. Although the pressing process is less technique sensitive than both powder condensation and slip casting techniques, proper preparation of an appropriate investment material, burnout and pressure setting for compressed air are required for a successful press. It has been proven that the bonding of manufacturer recommended veneering ceramic to the zirconia substructure varies for different types of zirconia.[27] Also the manufacturer's recommended thickness of zirconium copings which support veneering porcelain ranges from 0.5-0.8 mm with the coping shoulder covering the margins of the tooth preparation. Both the flexural strength of the veneering porcelain and ceramic core must be compatible and the thickness of veneering porcelain should not exceed that of the core. In other words, overlay cannot exceed the thickness of its core and the thickness of the substructure material must be adequate to prevent cracking under pressure in the press oven. The manufacturer's chemical composition and flexural strength compatibility guidelines for ceramic core materials were used to choose the pressable veneering porcelain for this study.

In contrast to manufacturers design specifications, the inherent nature of a cut-back core and the current CAD/CAM technique employed for processing ceramic cores produces a thinned out core margin. Since the core margin is not coincident with the restoration margin in the modified core samples, the result was a larger than recommended veneer-core thickness ratio at the level of the core margin. Since the overall axial wall thickness of 1.5 mm is maintained this veneer-core thickness discrepancy lies hidden within the axial wall and above the level of the restoration margin. This could be a possible explanation for a number of core fractures which occurred during pressing. In the press oven, eleven modified core samples developed hairline core fractures while the veneering porcelain was pressed onto it and remained intact with no evidence of veneer voids or fracture. These hairline fractures were only visible under magnification and light illumination at an angle which demonstrated a change in direction of light at the fracture line. These samples were discarded and not used for maximum load and fracture mode testing. It should be noted that fractures of this nature, similar to hairline vertical tooth fractures, are not easily clinically detected and these restorations may possibly be cemented without knowledge of the core defect. This illustrates the importance of thorough inspection of the cores of all-ceramic restorations by shinning a light through them before final cementation. These findings are in agreement with previous studies that also conclude that the mechanical properties and microstructure of core materials are crucial to the clinical long-term performance of all-ceramic dental restorations. [28]

The most frequent mode of failure found in this study was within the veneering layer of porcelain only. Finite element analysis studies on single-cycle-loaded to failure crowns have indicated high stress levels below the load origins. Like this study, these mouth motion sliding contact fatigue loads resulted in veneer chipping, reproducing clinical findings.[29] When more catastrophic fracture occurred, it was at the porcelain-core interface and left the core ceramic intact. These findings are consistent with previous surface fracture analyses. For example in a study by Van Der Zel et al, the crack initiation site was typically located on the inside of the coping at the glass-zirconia interface. [30] Coelho et al in a cyclic loading study of all-ceramic crowns found in fatigue, failure occurred by formation of large chips within the veneer originating from the contact area without core exposure and concluded that LAVA and CERCON ceramic systems present similar fatigue behavior. This was consistent with this study as well as clinically observed failure modes.[31]

The cementation protocol implemented in this study involved both etching and sandblasting with 50 µm aluminum oxide. However, Zhang et al suggested that aluminum oxide abrasion may propagate crack formation consequently decreasing fracture toughness.[32] Consequently, the methodology employed in the study could have possibly introduced inherent micro-fractures in the zirconia core and impacted the overall test results. This was the rationale behind using silanization and Panavia 21 (containing phosphate) cement which have been shown to significantly increase the bond strength.[16] Conditioning the high-strength ceramic surfaces with silica coating and silanization provides higher bond strengths of the resin cement than with airborne particle abrasion with 110 µm Al_2O_3 and silanization.[17] Furthermore, standard resin cements (e.g. RelyX ARC, Panavia F, Variolink II, Compolute) have been shown to produce higher bond strengths than a self-adhesive universal resin cement (e.g. RelyX Unicem).[17] In fact, in a study by Luthy et al the strongest bond to zirconia was obtained with Panavia 21 when tested among four other adhesive resin cements (Rely X Unicem, Superbond C&B, Panavia F, and Panavia 21).[16] This particular cement has a low film thickness and can penetrate microscopic porosities and microcracks which can increase the bond strength of the restoration. This intimate bonding provides a 'die-crown' unit effect which allows partial stress factors to be transferred to the die and reduces stress creating failures of the restoration.[33]

Many studies have been performed to test the reliability and clinical applicability of physical test methods of dental restorations. The present study did not replicate physiological tooth mobility which has been demonstrated to influence resistance to fracture. More specifically, periodontal support can provide as much as a threefold resistance to force.[34] This data suggests that the total range of fracture loads recorded among all individual samples (733-1729 N) should translate to a threefold intraoral resistance to force of approximately 2200-5200

N. Also in this study, samples were axially loaded unlike the direction of loading for natural dentition. However, Bonfante et al concluded that reliability was not significantly different between axial and off-axis mouth-motion fatigued veneer pressed over Y-TZP cores.[35] For this reason it can be concluded that the axial loading in this study adequately represented load data for the typical off-axis loading of natural dentition. In a follow-up study Bonafante also concluded that fractures originated from the contact area regardless of the cusp loaded and no significant difference on fatigue reliability was observed between the DF compared to the ML cusp. [36] Since fracture of one cusp did not affect the other, it can be assumed that buccal cups loaded produced values representative of the strength of the restoration as a whole.

This in-vitro study illustrates the importance of sample design in the production of clinically applicable data. The data obtained also provides clinicians with a basis to limit the extent to which zirconia cores can be modified without compromise to the strength of all-ceramic restorations. More clinically relevant specimen geometry, surface finish, and mechanical loading are being applied to in vitro studies. Therefore, in vitro studies are becoming more reliable indicators of the clinical performance of ceramic prostheses. Regardless of these improvements, clinicians should exercise caution when extrapolating from the laboratory data to clinical cases.[5] Clinical failures can be simulated by blunt contact loading, cyclic fatigue loading, and loading in an aqueous environment.[37] Regardless of these efforts, physical testing does not guarantee a clinically relevant mode of failure. Specimens should be prepared and loaded using a clinically applicable method that reproduces clinical modes of failure. Flat crown specimens versus anatomically designed specimens are not clinically applicable to meet the morphological limitations of the human dentition. In this study the average fracture loads and maximum loads were higher for the flat specimens when compared with the anatomical designs. This demonstrates that overall surface specimen geometry may influence final maximum load and fracture resistance of crowns tested. This illustrates the limitations of specimen design and mode of in-vitro testing in the production of clinically applicable data. Clinicians should be able to decipher whether physical models accurately reflect physical attributes of the dynamic clinical environment.

9. CONCLUSIONS

Within the limitations of this study, the following conclusions can be drawn from the heat pressed all-ceramic, zirconia core crowns tested:

a.) Flat samples have an average 18% higher mean maximum load than anatomically designed all-ceramic crowns.

b.) Modification of core design by cutting back its facial length decreases the overall maximum load resistance of the all-ceramic crown by 21-22%.

c.) Regardless of core design, porcelain fracture occurs mainly within the veneering layer for all groups tested.

d.) Both the manufacturer's recommended zirconia core extension and the modified cut-back cores have the potential to withstand average physiological occlusal forces.

10. CLINICAL RELEVANCE

This in-vitro study illustrates the importance of sample design in the production of clinically applicable data as well as providing clinicians with a basis to limit the extent to which zirconia cores can be cut-back short of margins without compromise to the strength of all-ceramic restorations.

This study introduces a promising outcome for future modifications to margin design of heat pressed to zirconia restorations. The zirconia coping modification in this study improved esthetics while producing a restoration capable of withstanding normal physiologic loads. However, clinical trials are necessary to accurately determine the performance of such changes in the physiologic environment of the oral cavity.

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12. APPENDIX I

DESCRIPTIVE STATSISTICAL ANALYSIS

<u>MEANS AND STANDARD DEVIATIONS</u>

THE FOLLOWING RESULTS ARE FOR: GROUP = 1 TOTAL OBSERVATIONS: 10

	MAXLDKG	MAXLDN	FRACLDKG	FRADLDN
N OF CASES	10	10	10	10
MEAN	103.308	1013.106	99.976	980.428
VARIANCE	263.920	25380.699	327.454	31491.568
STANDARD DEV	16.246	159.313	18.096	177.459

THE FOLLOWING RESULTS ARE FOR: GROUP = 2 TOTAL OBSERVATIONS: 11

	MAXLDKG	MAXLDN	FRACLDKG	FRADLDN
N OF CASES	11	11	11	11
MEAN	126.749	1242.985	121.500	1191.509
VARIANCE	612.220	58877.462	666.845	64130.182
STANDARD DEV	24.743	242.647	25.823	253.239

THE FOLLOWING RESULTS ARE FOR: GROUP = 3 TOTAL OBSERVATIONS: 11

	MAXLDKG	MAXLDN	FRACLDKG	FRADLDN
N OF CASES	11	11	11	11
MEAN	132.573	1300.093	129.691	1271.834
VARIANCE	1392.780	133944.177	1415.738	136150.528
STANDARD DEV	37.320	365.984	37.626	368.986

THE FOLLOWING RESULTS ARE FOR: GROUP = 4 TOTAL OBSERVATIONS: 11

	MAXLDKG	MAXLDN	FRACLDKG	FRADLDN
N OF CASES	11	11	11	11
MEAN	160.335	1572.172	157.309	1542.673
VARIANCE	204.998	19822.659	196.882	18934.274
STANDARD DEV	14.318	140.793	14.031	137.602

GROUPS 2 VS 4

INDEPENDENT SAMPLES T-TEST ON MAXLDKG BY GROUPS

GROUP	N	MEAN	SD
2	11	126.749	24.743
4	11	160.335	14.318

SEPARATE VARIANCES T = -3.897 DF = 16.0 PROB = 0.001

POOLED VARIANCES T = -3.897 DF = 20 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON MAXLDN BY GROUPS

GROUP	N	MEAN	SD
2	11	1242.985	242.647
4	11	1572.172	140.793

SEPARATE VARIANCES T = -3.892 DF = 16.0 PROB = 0.001

POOLED VARIANCES T = -3.892 DF = 20 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON FRACLDKG BY GROUPS

GROUP	N	MEAN	SD
2	11	121.500	25.823
4	11	157.309	14.031

SEPARATE VARIANCES T = -4.041 DF = 15.4 PROB = 0.001

POOLED VARIANCES T = -4.041 DF = 20 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON FRADLDN BY GROUPS

GROUP	N	MEAN	SD
2	11	1191.509	253.239
4	11	1542.673	137.602

SEPARATE VARIANCES T = -4.041 DF = 15.4 PROB = 0.001

POOLED VARIANCES T = -4.041 DF = 20 PROB = 0.001

GROUPS 1 VS 3

INDEPENDENT SAMPLES T-TEST ON MAXLDKG BY GROUPS

GROUP	N	MEAN	SD
1	10	103.308	16.246
3	11	132.573	37.320

SEPARATE VARIANCES T = -2.366 DF = 13.9 PROB = 0.033

POOLED VARIANCES T = -2.287 DF = 19 PROB = 0.034

INDEPENDENT SAMPLES T-TEST ON MAXLDN BY GROUPS

GROUP	N	MEAN	SD
1	10	1013.106	159.313
3	11	1300.093	365.984

SEPARATE VARIANCES T = -2.366 DF = 13.9 PROB = 0.033

POOLED VARIANCES T = -2.286 DF = 19 PROB = 0.034

INDEPENDENT SAMPLES T-TEST ON FRACLDKG BY GROUPS

GROUP	N	MEAN	SD
1	10	99.976	18.096
3	11	129.691	37.626

SEPARATE VARIANCES T = -2.339 DF = 14.7 PROB = 0.034

POOLED VARIANCES T = -2.267 DF = 19 PROB = 0.035

INDEPENDENT SAMPLES T-TEST ON FRADLDN BY GROUPS

GROUP	N	MEAN	SD
1	10	980.428	177.459
3	11	1271.834	368.986

SEPARATE VARIANCES T = -2.339 DF = 14.7 PROB = 0.034

POOLED VARIANCES T = -2.267 DF = 19 PROB = 0.035

GROUPS 1 VS 2

INDEPENDENT SAMPLES T-TEST ON MAXLDKG BY GROUPS

GROUP	N	MEAN	SD
1	10	103.308	16.246
2	11	126.749	24.743

SEPARATE VARIANCES T = -2.588 DF = 17.4 PROB = 0.019

POOLED VARIANCES T = -2.537 DF = 19 PROB = 0.020

INDEPENDENT SAMPLES T-TEST ON MAXLDN BY GROUPS

GROUP	N	MEAN	SD
1	10	1013.106	159.313
2	11	1242.985	242.647

SEPARATE VARIANCES T = -2.588 DF = 17.4 PROB = 0.019

POOLED VARIANCES T = -2.537 DF = 19 PROB = 0.020

INDEPENDENT SAMPLES T-TEST ON FRACLDKG BY GROUPS

GROUP	N	MEAN	SD
1	10	99.976	18.096
2	11	121.500	25.823

SEPARATE VARIANCES T = -2.228 DF = 17.9 PROB = 0.039

POOLED VARIANCES T = -2.190 DF = 19 PROB = 0.041

INDEPENDENT SAMPLES T-TEST ON FRADLDN BY GROUP

GROUP	N	MEAN	SD
1	10	980.428	177.459
2	11	1191.509	253.239

SEPARATE VARIANCES T = -2.228 DF = 17.9 PROB = 0.039

POOLED VARIANCES T = -2.190 DF = 19 PROB = 0.041

GROUPS 3 VS 4

INDEPENDENT SAMPLES T-TEST ON MAXLDKG BY GROUPS

GROUP	N	MEAN	SD
3	11	132.573	37.320
4	11	160.335	14.318

SEPARATE VARIANCES T = -2.304 DF = 12.9 PROB = 0.039

POOLED VARIANCES T = -2.304 DF = 20 PROB = 0.032

INDEPENDENT SAMPLES T-TEST ON MAXLDN BY GROUPS

GROUP	N	MEAN	SD
3	11	1300.093	365.984
4	11	1572.172	140.793

SEPARATE VARIANCES T = -2.301 DF = 12.9 PROB = 0.039

POOLED VARIANCES T = -2.301 DF = 20 PROB = 0.032

INDEPENDENT SAMPLES T-TEST ON FRACLDKG BY GROUPS

GROUP	N	MEAN	SD
3	11	129.691	37.626
4	11	157.309	14.031

SEPARATE VARIANCES T = -2.281 DF = 12.7 PROB = 0.040

POOLED VARIANCES T = -2.281 DF = 20 PROB = 0.034

INDEPENDENT SAMPLES T-TEST ON FRADLDN BY GROUPS

GROUP	N	MEAN	SD
3	11	1271.834	368.986
4	11	1542.673	137.602

SEPARATE VARIANCES T = -2.281 DF = 12.7 PROB = 0.040

POOLED VARIANCES T = -2.281 DF = 20 PROB = 0.034

GROUPS 1 AND 2 (5) VS GROUPS 3 AND 4 (6)

INDEPENDENT SAMPLES T-TEST ON MAXLDKG GROUPED BY GROUPEDV

GROUP	N	MEAN	SD
5	21	115.587	23.849
6	22	146.454	31.028

SEPARATE VARIANCES T = -3.667 DF = 39.3 PROB = 0.001

POOLED VARIANCES T = -3.645 DF = 41 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON MAXLDN GROUPED BY GROUPEDV

GROUP	N	MEAN	SD
5	21	1133.519	233.881
6	22	1436.132	304.319

SEPARATE VARIANCES T = -3.666 DF = 39.2 PROB = 0.001

POOLED VARIANCES T = -3.643 DF = 41 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON FRACLDKG GROUPED BY GROUPEDV

GROUP	N	MEAN	SD
5	21	111.250	24.538
6	22	143.500	31.108

SEPARATE VARIANCES T = -3.783 DF = 39.6 PROB = 0.001

POOLED VARIANCES T = -3.762 DF = 41 PROB = 0.001

INDEPENDENT SAMPLES T-TEST ON FRADLDN GROUPED BY GROUPEDV

GROUP	N	MEAN	SD
5	21	1090.994	240.636
6	22	1407.253	305.060

SEPARATE VARIANCES T = -3.783 DF = 39.6 PROB = 0.001

POOLED VARIANCES T = -3.762 DF = 41 PROB = 0.001

DESCRIPTIVE STATS FOR GROUPS 1 & 3 (7) VS 2 & 4 (8)

THE FOLLOWING RESULTS ARE FOR: GROUPANA = 7 TOTAL OBSERVATIONS: 21

	MAXLDKG	MAXLDN
N OF CASES	21	21
MEAN	118.637	1163.432
VARIANCE	1039.456	99964.241
STANDARD DEV	32.241	316.171

THE FOLLOWING RESULTS ARE FOR: GROUPANA = 8 TOTAL OBSERVATIONS: 22

	MAXLDKG	MAXLDN
N OF CASES	22	22
MEAN	143.542	1407.578
VARIANCE	684.591	65857.364
STANDARD DEV	26.165	256.627

T-TEST OF GP7 VS GP8

INDEPENDENT SAMPLES T-TEST ON MAXLDKG GROUPED BY GROUPANA

GROUP	N	MEAN	SD
7	21	118.637	32.241
8	22	143.542	26.165

SEPARATE VARIANCES T = -2.774 DF = 38.5 PROB = 0.008

POOLED VARIANCES T = -2.787 DF = 41 PROB = 0.008

INDEPENDENT SAMPLES T-TEST ON MAXLDN GROUPED BY GROUPANA

GROUP	N	MEAN	SD
7	21	1163.432	316.171
8	22	1407.578	256.627

SEPARATE VARIANCES T = -2.773 DF = 38.5 PROB = 0.009

POOLED VARIANCES T = -2.786 DF = 41 PROB = 0.008

2 WAY ANOVA FOR KGF; ANATOMY VS CUTBACK

ANALYSIS OF VARIANCE

SOURCE	SUM-OF-SQUARES	DF	MEAN-SQUARE	F-RATIO	P
CORE	7034.181	1	7034.181	11.209	0.002
CORE2	10598.258	1	10598.258	16.888	0.000
CORE*CORE2	50.108	1	50.108	0.080	0.779
ERROR	24475.257	39	627.571		

13. APPENDIX II

DIAL CALIPER READINGS FOR THE THICKNESS OF ZIRCONIA CORES BEFORE AND AFTER WAXING AND OF FINAL ALL-CERAMIC CROWNS

Table 13. Thicknesses of Zirconia Cores Before and After Waxing and of All-Ceramic Crowns in GROUP 1

Sample	Width of Zirconia Coping					Width of Coping & wax-up					Width of Final Crown				
Number	B	L	M	D	O	B	L	M	D	O	B	L	M	D	O
1	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.1	1.4	1.5	1.5	1.5	2.0
2	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
3	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
4	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
5	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
6	0.6	0.6	0.6	0.7	0.7	1.5	1.5	1.5	1.5	2.1	1.4	1.5	1.5	1.5	2.0
7	0.6	0.6	0.6	0.6	0.7	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
8	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
9	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
10	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0

Table 14. Thicknesses of Zirconia Cores Before and After Waxing and of All-Ceramic Crowns in GROUP 2

Sample	Width of Zirconia Coping					Width of Coping & wax-up					Width of Final Crown				
Number	B	L	M	D	O	B	L	M	D	O	B	L	M	D	O
1	0.3	0.6	0.6	0.6	0.7	1.5	1.5	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.0
2	0.3	0.6	0.6	0.6	0.7	1.6	1.6	1.6	1.7	2.1	1.5	1.6	1.5	1.6	2.0
3	0.3	0.6	0.6	0.6	0.6	1.7	1.7	1.6	1.6	2.1	1.5	1.6	1.5	1.5	2.0
4	0.3	0.6	0.5	0.5	0.7	1.7	1.6	1.6	1.7	2.2	1.5	1.5	1.5	1.5	2.1
5	0.3	0.7	0.7	0.7	0.7	1.5	1.7	1.5	1.5	2.1	1.5	1.6	1.5	1.5	2.0
6	0.5	0.7	0.7	0.7	0.7	1.6	1.6	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.0
7	0.2	0.7	0.6	0.7	0.7	1.6	1.6	1.6	1.7	2.2	1.5	1.5	1.5	1.5	2.0
8	0.5	0.6	0.7	0.6	0.6	1.6	1.7	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.1
9	0.4	0.6	0.6	0.6	0.7	1.7	1.7	1.7	1.6	2.1	1.6	1.5	1.5	1.5	2.0
10	0.4	0.7	0.6	0.7	0.6	1.6	1.7	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.0
11	0.4	0.5	0.6	0.6	0.7	1.6	1.7	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.0

Table 15. Thicknesses of Zirconia Cores Before and After Waxing and of All-Ceramic Crowns in GROUP 3

Sample	Width of Zirconia Coping					Width of Coping & wax-up					Width of Final Crown				
Number	B	L	M	D	O	B	L	M	D	O	B	L	M	D	O
1	0.6	0.6	0.6	0.5	0.6	1.5	1.5	1.5	1.4	2.0	1.5	1.5	1.5	1.5	2.0
2	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
3	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.4	1.5	1.5	1.5	2.0
4	0.6	0.5	0.6	0.5	0.6	1.5	1.4	1.5	1.4	2.0	1.5	1.5	1.5	1.5	2.0
5	0.6	0.6	0.6	0.5	0.6	1.5	1.5	1.5	1.4	2.0	1.5	1.5	1.5	1.5	2.0
6	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.4	1.5	1.5	1.5	2.0
7	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
8	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
9	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0
10	0.6	0.6	0.6	0.6	0.6	1.5	1.5	1.5	1.5	2.0	1.5	1.5	1.5	1.5	2.0

Table 16. Thicknesses of Zirconia Cores Before and After Waxing and of All-Ceramic Crowns in Group 4

Sample	Width of Zirconia Coping					Width of Coping & wax-up					Width of Final Crown				
Number	B	L	M	D	O	B	L	M	D	O	B	L	M	D	O
1	0.7	0.7	0.6	0.7	0.7	1.5	1.5	1.6	1.6	2.1	1.5	1.5	1.5	1.5	2.0
2	0.7	0.7	0.7	0.7	0.7	1.6	1.6	1.6	1.7	2.1	1.5	1.6	1.5	1.6	2.0
3	0.6	0.6	0.6	0.6	0.6	1.7	1.7	1.6	1.6	2.1	1.5	1.6	1.5	1.5	2.0
4	0.6	0.6	0.6	0.6	0.7	1.7	1.6	1.6	1.7	2.2	1.5	1.5	1.5	1.5	2.1
5	0.6	0.6	0.6	0.6	0.7	1.5	1.5	1.5	1.5	2.1	1.5	1.5	1.5	1.5	2.0
6	0.7	0.6	0.6	0.7	0.7	1.6	1.5	1.6	1.6	2.2	1.5	1.5	1.5	1.5	2.0
7	0.7	0.7	0.7	0.7	0.7	1.6	1.6	1.6	1.7	2.2	1.5	1.5	1.5	1.5	2.0
8	0.6	0.6	0.6	0.6	0.6	1.6	1.7	1.6	1.6	2.3	1.5	1.5	1.5	1.5	2.1
9	0.6	0.6	0.6	0.6	0.7	1.7	1.7	1.7	1.6	2.3	1.5	1.5	1.5	1.5	2.0
10	0.7	0.7	0.7	0.7	0.7	1.6	1.7	1.6	1.7	2.2	1.5	1.5	1.5	1.5	2.0
11	0.6	0.6	0.6	0.6	0.7	1.6	1.7	1.6	1.6	2.3	1.5	1.5	1.5	1.5	2.0