Tufts UNIVERSITY | School of Dental Medicine

Flexural Strength of Three-Unit Monolithic Zirconia FPD With Different Size of Connectors Sintered in SpeedFire Oven

A Thesis

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ABSTRACT

Aim of the study: The aim of this in vitro study was to investigate the flexural strength of three-unit monolithic zirconia FPD with different size of connectors sintered in the SpeedFire Oven.

Materials & Methods: Thirty-five three-unit zirconia (In-Coris TZI C Medi S A1, Sirona Dental Systems GmbH, Bensheim, Germany) FPDs with different connector sizes were milled, sintered, and glazed. Three groups were formed based on the connector size: Group A: 3x3 mm, Group B: 3x4 mm and Group C: 4x4 mm. Flexural strength was calculated in MPa with the aid of a universal testing machine (Model 5566; Instron Corp, Canton, MA) along with the determination of mode of failure. The force was applied at the center of the bridge at a crosshead speed of 0.5mm/min, until it cracked.

Results: Group A had the highest mean flexural strength (1160.71MPa with a standard deviation of 140.70). One-way ANOVA comparing the groups was statistically significant (p<0.001). Post hoc tests using the Tukey-Kramer test showed significant differences between Groups A and B (p<0.001) and Groups A and C (p<0.001), but not Groups B and C (p=0.698). 74.3% of the fractures occurred in the connector part of FPDs and 25.7% happened in the margin of restorations. Concerning the mode of failure, based on Fisher's exact test, the difference between groups was not statistically significant. (p=0.717).

Conclusions:

1- Flexural strength of three-unit FPDs with connector sizes 3x3mm sintered in the SpeedFire oven was significantly higher than that of three-unit FPDs with connector sizes 3×4 mm and 4×4 mm.

2- Group A, three-unit FPDs with connector sizes 3x3mm, that were sintered with the short sintering protocol had clinically acceptable flexural strength while the other two groups did not have acceptable flexural strength.

DEDICATION

I dedicate this dissertation to individuals who inspire my soul, add valuable meanings to my life, and motivate me to be great today. I dedicate this work to my phenomenal husband, who knew the timely words of affirmation and motivation that kept me moving forward. Your eternal devotion, relentless support, and encouragement throughout this journey were priceless and will always be recalled. I never would have had the strength to persevere and accomplish such a major goal without you.

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

Flexural Strength of Three-Unit Monolithic Zirconia FPD With Different Size of Connectors Sintered in SpeedFire Oven

Introduction

It is clearly necessary to replace any missing teeth in both the anterior and posterior areas, because the dental arch's structural unity becomes distorted and a new state of equilibrium develops following tooth loss.

The types of prostheses used commonly to replace missing teeth include an implant-supported fixed partial denture, a tooth-supported fixed partial denture (FPD), and a removable partial denture. The fixed partial denture is fixed permanently with implants or teeth, and acts as a substitute for single or multiple teeth[1].

The FPD connector is one of the prosthesis' most crucial elements, and can be either a moveable or rigid joint that connects the pontic to the abutment. The size and shape of the FPD connector is of substantial significance when any type of material is used, particularly in the case of a zirconia bridge[1].

The connector is undoubtedly a prosthesis' weakest component, and developing a restoration that demonstrates an adequate degree of durability and strength requires it to have proper width and height,[2] as unexpected fractures can be avoided in this way. Researchers have demonstrated that there is an association between prosthesis failure and fractures that initiate in the connector's gingival part. On the basis of these reports' findings, the connector's optimum dimensions are 3×3mm[2].

A variety of dental materials have been recommended for FPD restorations. However, given the growing need for restorations with esthetic qualities, researchers have made several attempts to develop a substance that demonstrates not only the proper mechanical strength to resist masticatory forces, but also has an appealing appearance[1]. Metal-ceramic systems were introduced as restorative substances in dentistry in the

1960s. These systems are based on veneering a metal substructure with a ceramic substance that has been subjected to firing and color customization. Feldspathic porcelains are used largely as veneering ceramics,[3] and in general, these are leucite-based in the case of metal-ceramic restorations. Metal-ceramic restoration has proven to be the gold standard in the dental field because of its reliability and long history of usefulness[3]. There is an increasing demand for metal-free restorative substances that demonstrate a substantial degree of translucency that provides a natural appearance,[4] and innovative dental ceramic substances have been developed during the past ten years. These include glass ceramics, zirconia-based ceramics, and poly-crystalline alumina. Advancements have been made in the processing technology as well, and this has led to the use of such computer-assisted fabrication systems as dental computer-assisted design/computer-assisted manufacturing (CAD/CAM)[1].

Polycrystalline ceramics have superior mechanical properties that include a flexural strength that ranges from 900-1200 MPa and a fracture strength of approximately 9-10 MPa (m1/2) compared to a high-strength glass ceramic (e.g., lithium disilicate), which has a flexural strength that ranges from 360-400 MPa and a fracture strength of 2.25 MPa(m1/2)[5,6].

Zirconia was introduced as a biomaterial for use in orthopedic applications in 1969[6]. Zirconium (Zr) is a metal with the atomic number of 40. Chemist Martin Klaproth discovered this grayish metal first in 1789. Zirconium's density and melting and boiling points are 6.49g/cm^3, 1852 ^{o}C, and 3580 ^{0}C, respectively, and its crystal structure demonstrates a hexagonal geometry. Moreover, it is not found in pure form in nature, but occurs as Zircon or Baddeleyite. Zircon is the silicate oxide of zirconium (ZrO_2 x SiO_2)

and Baddeleyite is a zirconium-free oxide (Zirconia, ZrO_2). The mineral forms of zirconium mentioned above cannot be used in dental applications because metal impurities may influence the substance's color. Further, the presence of natural radionuclides, such as thoria and urania, cause them to emit radiation[6]. Zirconia was introduced in the field of dentistry during the early 2000s as a substitute for metal substructures for veneered porcelain. However, a complicating factor compromises zirconia restorations' durability considerably, in that the veneered porcelain is prone to chip[2]. To address this issue, scientists devised full-contour zirconia restorations that exhibit a substantially increased flexural strength[7].

In the context of dental ceramic restoration, strength refers to clinical potential versus limitation. Moreover, a dental restoration substance's flexural strength is usually considered a reliable and efficient indicator of the ceramic substance's durability[8]. In dental appliances, bending forces generate flexural (bending) stress in one of the following two ways: (1) When a structure such as an FPD is subjected to three-point loading in which force is applied between fixed endpoints, and (2) when a cantilevered structure is subjected to a load along any section of the unsupported part, as the structure is supported at only one end. In addition to these, when an individual bites something, the anterior teeth are subjected to forces that are at an angle to their long axes, which results in the development of flexural stresses within the teeth[9]. However, substances such as zirconia demonstrate increased flexural strength that provides restorations that are less prone to fracture[10].

Manufacturing zirconia restorations involves milling and sintering, and CAD/CAM technology can be used to mill zirconia, while a sintering furnace is used to sinter it. This technology allows the zirconia structure to be converted to its complex geometrical forms. Different parameters used in the sintering process, particularly following the use of manufacturers' short sintering cycles, affect zirconia's microstructure and characteristics substantially[11]. Further, changes in the conditions used in the sintering process are known to affect zirconia frameworks' strength parameters[12]. Researchers have carried out studies with the goal to increase the strength parameters, and to do so, they shortened the process of zirconia sintering by reducing the sintered holding time and using a rapid heating rate. Although they found that these attempts did not influence the zirconia core's strength,[12] superior translucency was acquired by altering such sintering parameters as the increment in the sintering temperature or sintered holding time[13]. However, the flexural strength decreased when sintering was carried out at a very high temperature because yttrium particles moved to the border of the grain at very high temperatures[14]. When the sintered holding time was changed during zirconia sintering, the growth of the zirconia microstructure and grain size was affected considerably thereby, and altered the zirconia's translucency and strength. With an increase in grain size, zirconia is more likely to undergo spontaneous phase transformations that increase the risk of a gradual change in strength. Although a number of studies have reported the influence of altering the temperature and time used in the sintering process on zirconia core ceramics' strength, microstructure, and optical translucency, the influence of altering these parameters on monolithic zirconia's strength still needs to be investigated in detail[12,15,16].

The CEREC SpeedFire, which is capable of sintering a crown in 10-15 minutes, is the most rapid sintering furnace available on the market, and the sintering and glazing processes are combined in this ergonomic and compact instrument. Further, pre-heating and a holding temperature is unnecessary when this instrument is used because the induction technology alone is responsible for the even distribution of the temperature[9].

The research in this study was carried out with specific consideration of the size of the connector to increase the flexural strength of a zirconia FPD subjected to sintering in the SpeedFire furnace. To the best of the author's knowledge, no research has yet been published that addresses the way that superfast sintering affects the flexural strength of a zirconia three-unit FPD with different connector sizes.

Literature Review

The past three decades have witnessed revolutionary advancements in the field of dentistry. These advancements relate not only to the development of innovative techniques and entities, but scientific evidence that promotes these substances and methods' clinical applications has also been obtained.

A Parisian apothecarist, Alexis Duchateau, introduced ceramics in the field of dentistry in the 18th century, and its use in dentistry attracted immense attention, as its esthetic features compared to other substitutes for teeth made it highly popular. Duchateau used a porcelain ceramic substance to develop a complete set of dentures[15,18]. In 1903, Charles Land revolutionized dental ceramics when he used fired porcelains to develop all-ceramic onlays and inlays, as well as crown restorations[19]. Further, these innovative fired porcelains made it possible to develop porcelain jacket crowns[20]. However, these jacket crowns had certain limitations, including the substance's weakness, the complex method of production, and limited luting agents. Hence, in the mid-1960s, McLean developed the alumina-reinforced porcelain jacket crown.

Approximately a decade later, researchers began to report the efficiency of all-ceramic crowns, and during the past three decades, considerable advancements have been made in different substances and methods that have allowed the wide use of all-ceramic restorations in the field of dentistry[17]. Since that time, dental ceramics have undergone an evolution, as changes have been made in their composition, such properties as esthetic characteristics, and production, packaging, and indications procedures. Although the initial forms of dental ceramics were biocompatible and offered increasingly esthetic outcomes, a new generation of ceramic substances that demonstrate improved durability and strength remained important given the weak tensile and shear stresses in early forms of dental ceramics[18].

The current millable and pressable substances combined with CAD/CAM technology have allowed the development of ceramic restorations that are not only esthetic, but also demonstrate minimal invasiveness and greater strength[21]. As innovative ceramic substances exhibit versatility, convenient use, and impressive strength, metal-free ceramic material can be chosen specifically for a particular treatment. Ceramics are nonmetallic, inorganic solids developed by heating and then cooling raw substances that may include borides, metal oxides, carbides, nitrides, and mixtures of these substances. Accordingly, a substance that is labelled a ceramic is actually not if it contains organic constituents or is developed through other methods[18,22].

Ceramic substances can be amorphous, such as glass, or alternatively, they may be comprised of a crystalline or partially crystalline structure. As the majority of dental ceramics contain at least some crystalline constituent, some researchers have describe them as inorganic, crystalline-containing substances rather than ceramics that contain non-crystalline glasses, although glasses are ceramics[10]. It is understandable that dental ceramics are usually grouped on the basis of their microstructure. Although such grouping makes it possible to comprehend dental ceramics' chemical and structural features quite conveniently, it does not help ceramists or dentists select a substance suitable for a particular intervention significantly. Because the method used to process ceramics affects their mechanical characteristics, and accordingly, their clinical behavior, substantially, dental ceramics must be grouped on the basis of their composition and processing. Such grouping can offer improved understanding of ceramics' clinical parameters to analyze and select a ceramic for a particular clinical intervention efficiently. E.A. McLaren grouped ceramics on the basis of preserving healthy tooth structure, and these groups range from the most to least conservative[18].

CL-I (Powder/Liquid)

Class I (CL-I) liquid and powdered porcelains are developed from substances that are comprised primarily of silicon dioxide, a glassy matrix with different amounts of a crystalline phase within the matrix. This category of ceramics encompasses feldspathic porcelains that are so named because they were previously (and some are still) synthesized from feldspars found naturally. Currently, a number of feldspathic substances are available commercially (e.g., Vintage Halo, Shofu, shofu.com; VITA VM 13, VITA Zahnfabrik, vita-zahnfabrik.com). Members of this group of ceramics are produced manually, demonstrate the highest conservation, and are largely those that are most translucent. However, they have also proven to be the weakest. These substances' esthetic characteristic of increased translucence confer the appearance of a natural tooth, and liquid or powder porcelains exhibit both impressive feasibility and esthetics.

Moreover, they have been recognized as the most conservative amongst all metal-free ceramic groups because they can be layered very finely and positioned directly on enamel. A 0.2-0.3mm-thick layer of CL-I porcelain is needed for each shade change, and this group of ceramics is usually recommended for anterior restorations. However, they can also be used occasionally for bicuspids, and rarely for molars, given that all parameters demonstrate low risk[18].

CL-II (Glass Ceramics)

Just like CL-I porcelain, CL-II ceramics are comprised of a glassy matrix; however, the crystal types and glass-crystalline ratios of these two groups differ. In the case of these ceramics, crystals can be grown either within the matrix or can be added to it. The method used to fabricate this class also differs from the previous one. In particular, these ceramics are made into dense industrial blocks that can be pressed as well as machined. CL-II pressed and machined glass ceramics are categorized further into two different categories on the basis of their crystal type and clinical behavior reported.

CL-IIa

Ceramics in this subdivision are comprised of feldspathic glass that contains small to moderate amounts of leucite, and are less than 50% crystalline. They work more or less like a glass that requires bonding, and their utility is the same as that of CL-I ceramics. They can be used for bicuspids, anterior teeth, and in rare cases, molars, and studies have reported their long-term clinical efficiency in settings with increased stress and in cases of exposed dentin. Although they demonstrate increased translucence, they have been used conventionally with somewhat thicker measurements for practicality and esthetics. These ceramics exhibit impressive material strength primarily because of the method used to process them, which involves the use of a dense, industrially-synthesized block that checks the propagation of cracks. Such dense leucite- and glass-containing substances are used for thicker veneers, posterior onlays and inlays, and anterior crowns, but only in cases where a long-term bond and seal can be sustained.

CL-IIb

This subdivision includes glass or glass ceramics that contain moderate to high amounts (>50%) of crystalline. These substances' microstructure is comprised of a glass matrix that envelops a second phase of individual crystals, and are formed as homogenous glass, which is followed by crystals' nucleation and growth through secondary treatment. This treatment provides these substances with better physical and mechanical characteristics by maximizing the crystals present in them and by generating compression stress around the crystals. Lithium disilicate (e.g., IPS E-max, IvoclarVivadent) is a common example of this subdivision. This glass ceramic substance is comprised of phosphorous pentoxide, potassium oxide, alumina, lithium dioxide, and silica. Although lithium disilicate is recommended for clinical interventions that make use of other glass ceramics, it has also proven to be suitable for increased stress conditions, such as those that call for full crowns, including those on molars. Under stress conditions, lithium disilicate must be formed to a full contour with a monolithic restoration and must be placed with resin cement. The zirconia-reinforced lithium silicates ZLSs (such as CELTRA™ Duo, Dentsply Sirona; VITA Suprinity) have been added to this subdivision recently. These substances are comprised of a lithium silicate glass ceramic to which 10% of zirconia crystals are added for increased strength. The long-term clinical evidence in the literature promotes the use of lithium disilicates only as single restorations at any place within the mouth. Restorations made from substances in this subdivision[12] demonstrate impressive strength, a natural look, and fracture resistance, and are a strong and versatile substitute for a broad range of indications, including cases with increased risks, for example, when less than 50% enamel is present on a tooth, when 30% or more of the margin is in dentin, and/or when enamel is less than 50% of the bonded substrate[18]. Given their glass characteristics, adhesive bonding is suggested for these substances; however, less predictable restorations are achieved when these are bonded to dentin because of dentin's flexibility. Restorations bonded to enamel demonstrate greater predictability because enamel is much stiffer than dentin[17].

CL-III (High-Strength Crystalline)

This category, which is fabricated by industrial processes, is comprised of high-strength crystalline ceramics that are devoid of, or contain a minimal amount of, a crystalline phase. They differ from glass or glass ceramics in the way in which a sintered crystalline matrix comprised of a substance that demonstrates an increased modulus (85-100% of volume) amalgamates with the crystalline phase particles.

CL-IIIa

Substances in this subcategory are fabricated by developing a porous matrix that is made into a block, after which CAD/CAM technology is used in the substance's final processing to achieve the shape required. Thereafter, the pores present in the substance are filled by melting a second phase substance. The next step entails drawing lanthanum aluminosilicate glass (in the form of liquid or molten glass) via capillary action into all pores, which forms an interpenetrating and dense substance from the internal to external surfaces thereby. The product thus formed is an 85% crystalline mesh that contains small quantities of glass. However, this substance is not marketed often today, as polycrystalline ceramics are replacing them completely.

CL-IIIb

This subcategory includes 100% crystalline ceramics that demonstrate impressive strength. Previously, this category included alumina-based substances. However, today, this category is constituted largely of zirconia-based substances. Although alumina systems' efficiency for single units has been demonstrated, they are associated with a greater risk of failure when used in the molar region. Hence, lithium and zirconia disilicate are substituted for them. Situations in which zirconia's use is recommended include the absence of a significant tooth structure, presence of stress, and increased risk for flexure, for fixed partial dentures and posterior full-crowns, and when problems are encountered with adhesive bonding, as in cases of subgingival margins.

CL-III ceramics demonstrate impressive strength; alternatively, CL-IV metal ceramics (described below) have proven to be suitable in situations where one cannot sustain the seal and bond. These are encountered in variable bonding interfaces, increased tensile and shear stresses bonded interfaces face, and problematic moisture control. These substances' suitability in the situations abovementioned is attributable to the fact that they can be positioned through traditional methods of cementation. Still, wear on opposing dentition has proven to be a significant issue associated with full-contour zirconia.

Whether zirconia- or alumina-based, the strength these substances exhibit is greater than that CL-I, as well as CL-II substances, exhibit. Accordingly, these substances can be used to develop a core substructure that substitutes for metal. Nevertheless, because of their increased crystalline concentration, they exhibit greater opacity and hence, unsatisfactory esthetics overall. Accordingly, porcelain is used to layer them so that greater strength is achieved as well as better esthetic characteristics. The substrate's color influences the thickness that CL-III high-strength ceramics require, and usually, a layer 1.2-1.5mm thick is used. Today, forms with greater translucence are used as monolithic all-zirconia restorations in the posterior area[18].

CL-IV (Metal Ceramics)

This group includes metal ceramics that are basically class-I substances combined with an increasingly supportive metal substrate that allows them to be used in conditions that involve increased stress and call for esthetics and traditional crowns. These substances have proven to be most suitable in cases in which there is minimal or no tooth structure. These metal ceramics exhibit substantial strength; however, their esthetic characteristics are unsatisfactory. The thickness required to achieve natural esthetics is at least 1.5mm. These substances' qualities resemble those of CL-III zirconia-based restorations, but their sensitivity to thermal firing differs from that of zirconia.

Collectively, ceramic substances that belong to CL-I and CL-II exhibit impressive esthetics, but their strength is limited. Ceramics of all kinds have weak resistance to shear and tensile stresses compared to compressive stresses. However, weak substances can be used effectively provided that stresses are controlled. On the other hand, ceramic substances in CL-III and CL-IV are sufficiently strong, but exhibit limited esthetics. For esthetic purposes, the substructure can be veneered with porcelain in situations in which functional stresses are uncontrollable and high-strength substances like metal, alumina or zirconia are used[18.]

Zirconia

The three phases of zirconia at low pressure that remain stable at room temperature are the cubic, tetragonal, and monoclinic phases. This particular feature is of great significance in zirconia's mechanical characteristics[22]. According to Daou's study, temperature at normal pressure affects the crystallographic structure of unalloyed zirconia. Its structure at room temperature is monoclinic and when the temperature is increased to 1170 ^{0}C. Further, it demonstrates a tetragonal structure within the temperature range of 1170-2370 ^{0}C. Finally, it demonstrates a cubic structure from temperatures over 2370 ^{0}C up to its melting point. A considerable volume increment (~4.5%) is apparent when the tetragonal (t) form transforms back to the monoclinic (m) form through cooling, at which time catastrophic failure occurs. When zirconia alloys that contain CeO_2, Y_2O_3, MgO, or CaO are added, the t structure is retained at room temperature. In this way, the *t→m* transformation-induced stress can be controlled. Further, compressive stresses produced close to the tip of a crack check crack propagation and result in increased toughness[23.] Zarone et al. conducted a systematic review that indicated that this characteristic is the reason why such substances have attracted many biomedical researchers' attention during past years, as it is able to induce a substantial increment in the material's fracture toughness by impeding (not preventing) the spread of a crack actually at the tip of crack. The concentration of tensile stress causes transformation and leads to a positive compressive stress that serves to limit cracks. This phenomenon, together with the substance's grain size, causes zirconia to demonstrate the greatest fracture toughness and flexural strength compared to the remainder of the ceramic substances[24].

Gargari et al. conducted a systematic literature review and found that micro-deficiencies in the structure of ceramic expand over time and advance to prostheses' fracture and failure, and occurs in all forms of ceramics except zirconia. Zirconia is able to address this issue through phase transformation (t → m), referred to also as transformation toughening. As mentioned earlier, transformation results in a volume increase (3-4%) that causes instant amplification of zirconia's fracture resistance[1]. Zirconia is amongst the few restorative substances that require fabrication in the laboratory or dental setting through digital technology. The fabrication system generally includes a sintering oven, milling machine, compatible zirconia milling blocks, and a scanner[25].

Methods Used to Fabricate Zirconia

Two different methods can be used to fabricate zirconia dental frameworks, soft machining of pre-sintered blocks and hard machining of fully-sintered blocks. The soft milling method is one of the simplest and most time-efficient methods to fabricate 3Y-TZP zirconia. Milling pre-sintered bocks serves as the basis for this method and these blocks are sintered completely during the final step. These zirconia blocks, which demonstrate the "green state," are generated through a method that uses the isostatic cold pressing phenomenon to compact zirconia. As a consequence, components are distributed evenly within the blocks and the pore size becomes extremely small, i.e., 20-30nm[26].

A pre-sintering temperature appropriate to process zirconia has proven to be an important factor, as it influences the blocks' roughness, machinability, and hardness. It is understandable that manufacturers are interested in convenient fabrication methods; however, in this context, machinability and hardness serve as opposing factors. A sufficient degree of hardness is important for safe manipulation of the 3Y-TZP blocks; however, excessive hardness has been proven to affect machinability negatively. Further, blocks with rough surfaces form at increased pre-sintering temperatures,[27] and increased temperature is used to complete the sintering process. The final mechanical characteristics are acquired when the zirconia framework experiences an approximately 25% linear shrinkage in volume and regains its impressive parameters thereby. This form of processing has been shown to generate highly stable cores that possess substantial amounts of tetragonal zirconia with surfaces that lack the monoclinic phase[26]. Still, some amounts of cubic zirconia can be found because of the heterogeneous distribution of yttrium oxide. The amount of stabilizing oxides in the cubic phase is greater than that in tetragonal crystals, and this difference can prove to be detrimental to the material's stability[28]. Accordingly, most manufacturers tend to prefer soft machining. However, the primary disadvantage of this approach is the issue of matching the framework's shrinkage attributable to sintering with the degree of enlargement software determines with the greatest precision possible[29]. Conversely, the hard machining method uses hot isostatic pressing to achieve dense sintering of the 3Y-TZP. This pressing is executed at high pressure and temperature that ranges from 1400 and 1500 ^{0}C in a setting that contains inert gas, and homogeneous, dense, and extremely hard blocks of completely sintered zirconia are formed in this way[30]. In vitro tests have demonstrated that increased flexural strength and fracture toughness can be achieved with different methods that involve both hot and cold isostatic-pressed zirconia blocks. However, in comparison to soft-machining, the hard-milling method requires more time, which also calls for cutting instruments that exhibit toughness and wear resistance. Compared to alumina sintered densely and zirconia blocks sintered completely, the completely-sintered 3Y-TZP blocks exhibit substantial hardness and less machinability. Therefore, the milling time is lengthier and the fabrication process is costly. Reports have indicated that grinding these blocks also produces different types of surface micro-cracks, as well as defects of a ductile or a brittle nature[31]. Deep defects can be encountered in the case of surface grinding that influence strength and toughness negatively. If these processing defects are exposed to moisture, more damage can occur, and reduce durability because of low-temperature degradation that is attributable to the slow transformation from the tetragonal to monoclinic phase in a moist environment[32]. Predictable stability is achieved for the framework with the soft machining method, provided the surface is not damaged following sintering. The surface's state following processing continues to be a controversial issue, particularly in the case of hard machining. Still, there is a general consensus that the main causes of fatigue damage and failure include micro-cracking because of processing defects or occlusal adjustments[26]. Residual stress is another factor that has a greater significance in inducing low temperature degradation (LTD) than the factor of final surface roughness. For example, when a ceramic substance with a dissimilar coefficient of thermal expansion (CTE) is used, stress is exerted during veneering or when zirconia is subjected to high temperature firing followed by rapid cooling. In addition to these, the existence of huge cubic phases affect zirconia's ability to resist aging and LTD negatively[33].

Sintering Zirconia

Closed and open system sintering are two different technologies used to sinter zirconia. Closed system sintering involves using a particular scanner that functions specifically with the manufacturer's milling machine that accepts zirconia milling blocks from the same manufacturer[34]. Later, researchers developed open CAD/CAM systems that allow the use of design files from a single scanner that varying milling units can read. Such open systems are able to use milling blocks from exterior sources and thus allow international suppliers to offer their zirconia milling blocks[34]. Conventional furnaces are used most widely to sinter zirconia. Sintering in these furnaces occurs at 1350-1400 ^{o}C over a period of 2-4 h[7]. However, Erosy et al. claimed that sintering zirconia takes several hours[11], and per Helvey's report, the sintering process requires nine hours[25].

The conditions used in sintering are known to affect zirconia's crystal size and content, stability, and mechanical characteristics. Thus, conditions must be regulated strictly throughout the process of generation[24]. For example, research has shown that the holding time used for sintering influences zirconia's grain growth. There is also an inverse relation between zirconia's grain size and its stability. The transformation stops completely if the grain size is less than 0.2um and its rate is low with smaller size grains (0.2-1um)[12,35]. On the other hand, as the grain size increases, the substance tends to undergo phase transformations that result eventually in a gradual decrease in strength. In addition, differences in the parameters used to sinter the substance can affect zirconia's characteristics and microstructure negatively[13]. The degree of this effect has attracted the attention of a large number of researchers, particularly following the introduction of ovens that support short sintering cycles. Many studies have been conducted to investigate the influence of various sintering parameters on zirconia's grain size, translucency, and biaxial flexural strength. Nonetheless, these effects still are not understood completely. Heating and cooling at a slow rate (typically 5-10 ^{0}C/minute) are involved in the general Y-TZP sintering process, which may require many hours and influence the material's strength adversely. At present, dental researchers are attempting to develop methods that allow sintering at a superfast rate[25]. To achieve this goal, researchers have devised super- and novel-speed sintering methods to address such requirements as cost- and time-effectiveness, and one-visit, chairside, CAD/CAM-generated restorations to prevent Y-TZP grain growth and ensure improved translucency.

Kaizer et al. studied the effect of sintering dwell time on Y-TZP's wear and mechanical characteristics. They produced monolithic molar crowns using translucent Y-TZP (InCoris TZI, Sirona), and used three different sintering methods, the long-term method (LT, 120 min holding time, 1510 ^{o}C), Speed method (S, 25 min holding time, 1510 ^{o}C) and Super-speed method (SS, 10 min holding time, 1580 ^{o}C). The authors observed a reduction in flexural strength with an increase in holding time,[22] but the crowns' hardness still remained the same regardless of the method used[7].

Hjerppe et al. carried out a study during which they compared the mechanical characteristics of Y-TZP samples obtained with two different sintering methods. In the first, Y-TZP discs were subjected to sintering at 20-1500 ^{o}C with a rise time of three hours and finally held for two hours at 1500 ^{o}C. In the second method, the-TZP discs were subjected to sintering at 20-1500 ^{o}C with a rise time of 100 min and finally held for one hour at 1500 ^{o}C. The differences in the micro-hardness and biaxial flexural strength of the specimens obtained from the two different methods were statistically insignificant[12].

As mentioned earlier, changes in the sintering conditions affect zirconia frameworks' strength substantially. Numerous researchers have attempted to reduce the time required for the sintering process through rapid heating and decreasing the holding time. However, these changes have had no effect on the zirconia's strength[36]. Increasing the sintering temperature or holding time leads to improved translucency because of additional cubic phase development, but flexural strength decreases if sintering is carried out at temperatures above 1600 °C because yttrium particles move toward the border of the grain,[36] and the heterogeneous distribution of yttrium-stabilizing ions causes undesirable cubic phases to develop[26].

Bogna et al. synthesized zirconia specimens (Ceramill ZI, Amann Girrbach) that were sintered in part to study sintering temperature's influence on grain size, contrast ratio, and flexural strength. The specimens obtained after sintering were subjected to random division to form nine groups (N = 198; n = 22 per group). The next step was final sintering carried out at various temperatures—1,300, 1,350, 1,400, 1,450, 1,500, 1,550, 1,600, 1,650, or 1,700 °C—and a holding time of 120 min. ISO 6872: 2008 was used to determine the three-point flexural strength, and scanning electron microscopy was performed to study the specimens' microstructures and grain sizes. According to their findings, final sintering temperatures that ranged from 1400 to 1550 °C resulted in the greatest flexural strength. Moreover, increments in the final sintering temperatures beyond 1300 °C amplified the zirconia's contrast ratio. When sintering temperatures were raised beyond 1300 °C, zirconia's microstructure demonstrated large sized grains. Similarly, increments in sintering temperature beyond 1600 ^{0}C caused grain growth and formation of zirconia with holes[14,26].

Stawarczyk et al. also reported increases in zirconia's grain size with an increase in the sintering temperature. The researchers demonstrated that sintering temperature is correlated negatively with flexural strength, and concluded that zirconia must be sintered at a temperature less than 1550 °C[14]. However, Ersoy[11] reported findings that contradicted those of Stawarczyk et al.[14] Ersoy studied the effect of sintering duration and temperature on zirconia's grain size and flexural strength, and used three different sintering methods with 120 In-Coris ZI and In-Coris TZI samples. Method I involved sintering at 1510 °C for 120 min; Method II involved sintering at 1540 °C for 25 min, and Method III involved sintering at 1580 °C for 10 min. Thereafter, the three-point flexural strength was determined. The results showed that samples sintered with Method III demonstrated the greatest flexural strength. However, the difference in the flexural strengths the samples sintered by Methods I and II was found to be insignificant, and the grain sizes were also found to be the same for the two test groups[11].

In the context of dental ceramic restoration, clinical potential versus limitation refers to strength. A ceramic substance's durability can be analyzed efficiently using flexural strength, which has proven to be a reliable indicator for such analysis. Substances that demonstrate increased flexural strength provide restorations that demonstrate less vulnerability to fracture[36].

According to Zarone et al.'s systematic review, zirconia exhibits mechanical characteristics far better than the remaining ceramics that can be used for dental interventions. Zirconia's compression resistance, flexural strength, and fracture toughness are 2000 MPa, 900–1200 MPa, and 6–10MPa/m1/2, respectively. It has also been shown that zirconia restorations exhibit an average load-bearing capacity of 755N, and reports have demonstrated fracture loads of 706N, 2000N, and 4100N. Studies also have indicated that zirconia dental restorations withstand greater fracture loads than do lithium or alumina disilicate. Researchers have studied zirconia FPDs in vitro recently and reported failure loads in the range of 379-501MPa, which is greater than the average human bite force. This verifies that these frameworks' serviceability is considerable[24].

Daou concluded from his research that zirconia frameworks exhibit a mechanical strength up to three times that all other ceramics exhibit. Moreover, they are able to withstand physiological occlusal forces faced in the posterior area[23].

Kelly et al. have shown through in vitro, as well as in vivo tests, that the only failure mode seen in all ceramic FPDs is connector fracture, and the results of a number of clinical studies that have focused on the analysis of all-ceramic FPDs are consistent with their findings. Hence, it can be stated that the primary cause of failure in all-ceramic FPDs differs from that found in metal ceramic FPDs. These failures in all-ceramic FPDs can be avoided by using connectors with an adequate width and height. Strength, and accordingly, the minimal parameters for connectors depend exclusively on the kind of ceramic substance used in the core material[37].

In FPD restoration, the connector's size and shape have proven to be crucial factors that affect zirconia-based fixed dental prostheses' mechanical characteristics, and connectors' fracture has proven to be the most frequent failure mode noted in all ceramic FPDs. 70-80% of cracks arise from the interface between the ceramic and the core[38].

A number of clinical trials have reported an association between the fractures of zirconia FPDs and inadequate height of the connector, as this type of prosthetic part that connects the pontic and retainers, shows a minor resistant to load. Sufficient flexural strength is required so that occlusal loads can be tolerated[24]. Thus, connectors' parameters have proven to be important factors that affect zirconia FPDs' efficiency. In vitro analyses have suggested that the minimum diameters for 3-, 4-, and 5-unit zirconia FPDs are 3-6mm, 4-6mm, and 5-6mm, respectively,[24] and the majority of manufacturers also recommend these diameters. An in vitro study has also shown that the radius of curvature at the connector's gingival embrasure affected the fracture resistance all-ceramic FPDs exhibited substantially. Accordingly, the gingival embrasure must have the greatest width possible,[24] as a larger radius of curvature at the gingival embrasure can reduce the magnitude of tensile stresses. However, sharp occlusal embrasures do not influence an FPD's fracture resistance[23]. Oh et al. conducted fractographic and finite element analyses and reported that connector fracture initiated at the gingival embrasure. They concluded that using a larger radius of curvature at the gingival embrasure decreased the magnitude of tensile stress and improved the FPD's fracture resistance thereby[39]. Further, researchers have suggested that connectors' lingual and gingival parts must be made only from the framework substance to facilitate acquisition of suitable connector dimensions without damaging the supporting tissues[38]. According to Daou's study, metal ceramic FPDs' long-term efficiency can be ensured through the use of connectors with a buccolingual width of 2.5mm, occlusogingival height of 2.5mm, and surface area of 6.25mm^2. Further, these parameters can be used in posterior as well as anterior regions.

As discussed above, a fracture usually extends from the connector's gingival surface to the pontic, and zirconia-based FPDs' fracture strength is amplified by 20% if a 3×3mm connector is used. The parameters needed for the connector can be even smaller compared to those needed for the remainder of the all-ceramic substances, and some reports have recommended the use of a 4×4mm connector[23]. It has been established that the framework's shape and thickness must be customized and optimized to achieve a veneered ceramic with even thickness as well as satisfactory support for the ceramic[24]. Campbell and Sozzi[40] and Kelly[37] studied FPDs that failed clinically. Their results showed that contact or Hertzian failure, which is the propagation of localized contact damage crack systems (Hertzian stress state) was not a clinical failure mode, and did not lead to framework failure; instead, most were from the gingival side at interdental connectors[40,37].

The three-point bending test is among the tests used most extensively to compute the modulus of rupture or transverse flexural strength a rectangular beam composed of a brittle substance exhibits. The Procera AllCeram Bridges system (Nobel Biocare, Goteborg, Sweden) framework's substance exhibits a transverse flexural strength that ranges from 500 to 650MPa. The minimum connector dimensions have a surface area of $6mm^2$ and $3mm^2$ occlusal/gingivally[41]. The In-Ceram Zirconia system (Vita Zahnfabric) framework substance exhibits a transverse flexural strength that ranges from 600 to 800MPa. Moreover, the connectors must be 3mm buccal/lingually and 4mm occlusal/gingivally[42].

Substances based on yttrium tetragonal Y-TZP are contemporary core substances available for all-ceramic FPDs. According to the findings of an in vitro study, the flexural strength Y-TZP bars demonstrated were 900-1200MPa[42]. Another in vitro study analyzed Y-TZP FPDs with varying connector dimensions after subjecting them to a static load, and found that they exhibited a fracture resistance of 1800-2000N. The fracture resistance exhibited by posterior 3-unit bridges supported with glass ionomer cement under the influence of cyclic load that simulates a 5-year clinical load was 1457N, i.e., much greater than the 1000N required[43]. Because of their physical and mechanical characteristics, frameworks based on YTZ-P may have a smaller connector area (7-16mm^2) compared to earlier forms[44]. Given the bite forces in the molar area, it appears that the use of yttrium oxide, partially stabilized zirconia ceramic is quite feasible and can be used as the core substance for 3-unit posterior FPDs with a connector size of 3×3mm.

Objective and Hypothesis

Goal

This study's goal was to investigate the flexural strength of three-unit monolithic Zirconia FPDs with connectors of varying sizes sintered in a SpeedFire oven.

Hypothesis

Smaller zirconia FPD connectors demonstrate a lower flexural strength than do larger zirconia FPD connectors.

Clinical Significance

The findings of this research will improve our understanding of the effect of short sintering cycles (SpeedFire) and the FPD connector's size on fractural strength. Moreover, the results may indicate that the factors above can save time for both patients and clinicians. The influence of the speed sintering method will also be addressed with respect to such requirements as cost- and time-effectiveness, and one-visit chairside appointment.

Materials and Methods

Sample Size Calculation

A power calculation was performed using nQuery Advisor v. 7.0. Based on the effect size obtained in a pilot study that included n = 3 per group, a sample size of n = 13 per group was found to be adequate to obtain a power greater than 99% with $\alpha = 0.05$. Groups A, B, and C included FPDs with connectors 3×3 mm, 3×4 mm, and 4×4 mm thick, respectively.

Master Model Fabrication

Abutment teeth (#18 and #20) were prepared on a typodont model (D95SDP-200, Kilgore, MI) with a coarse diamond chamfer bur accompanied by a water coolant. After the initial preparation of the teeth, an axial reduction of approximately 1mm with a 1mm chamfer finish line, a taper of 16 degrees, and an occlusal clearance of 1.5mm was carried out. A fine bur (KD7W6 Brasseler, Savannah, GA) was used to smooth the model further. The final step in the procedure involved scanning the model with a lab scanner calibrated according to the manufacturer’s instructions (Activity 880 scanner, Smart Optics, Bochum, Germany) after the model was drenched in an indicating spray (Quickcheck Indicating Spray, white, Vacalon Company Inc, Pickerington, OH). The teeth prepared with “metal die” fabrication (Cobalt-Chrome, Fusion Dental USA, Michigan, MI) using Selective Laser Melting technology (Fusion Dental USA service, Dexter, MI) with the aid of an STL file delivered to the lab. The end product was a replica of a typodont model (D95SDP-200, Kilgore, MI) that required teeth at positions 18 and 20 to be used for abutment and the tooth at 19 to be detached and used as a pontic.

Scanning and Designing Zirconia FPDs

Zirconia (In-Coris TZI C Medi S A1, Sirona Dental Systems GmbH, Bensheim, Germany) was selected as the material for the procedure after registering and viewing a new patient file in the Cerec Bluecam (CEREC AC Bluecam, Sirona, Charlotte, NC). Studying the file revealed that tooth no. 19 was to be placed as the pontic, with teeth no. 18 and 20 serving as abutments. The abutment teeth and the edentulous space were intended to replace tooth no. 19. The metal die was covered in indicating spray first (Quickcheck Indicating Spray, white, Vacalon Company Inc, Pickerington, OH), as the bluecam requires a thin layer of powder to scan. Then the metal die was scanned at an angle of 45^0 from all three aspects: occlusal, buccal, and lingual in the respective order. A dual system was used to ensure the cervical margins' clarity, first by an automated process and then manually by an operator. The FPD was then oriented carefully in such a way that its direction was in line with the prepared teeths' path of insertion. 36, 3-point fixed prostheses (teeth 18-×-20), each with an occlusal thickness of 1.0mm, axial thickness of 1.0mm, and marginal thickness of 0.5mm,[59] were designed using the CEREC computer software (CEREC 4.4, Sirona, Charlotte, NC). The connector dimensions used for this particular study were: 3×3mm, 3×4mm, and 4×4mm) for groups A, B, and C, respectively. The connector was designed and measured with the intelligent and user-friendly CEREC Omnicam software.

The connector cross-sections were a minimum of 9mm per CEREC's instructions. The scale factor of how much the zirconia had shrunk was calculated accurately by entering the bar code in the software. Alternatively, it could be entered in the milling unit display.

Zirconia blocks (In-Coris TZI C Medi S A1, Sirona Dental Systems GmbH, Bensheim, Germany) are generally used for dry milling, and calibration is performed beforehand to ensure that there is minimal deviation from the standard values. The block was dry milled within a Sirona CAD/CAM milling machine (CEREC MC XL Sirona Dental Systems GmbH, Bensheim, Germany). A diamond bur with a straight headpiece was used to separate the restoration and block at the sprue point. Air spray (Dust Off compressed air spray in a can-12 oz.) was blown across the pre-sintered restoration to remove any remaining specks of dust from the milled zirconia, and the dust was then removed using a size 10 sable brush.

Sintering

The CEREC software sends a request to the oven with all of the information required. A sintering chamber can be used for a maximum of 3 restorations at one time. Its working principle involves placing the restoration with its occlusal surface facing downward on the top layer adjacent to the opening of the sintering chamber. The restoration retains its position until the timer goes off at zero minutes, after which it is removed from the insulation layer using a pair of tweezers and repositioned at the fan-area of the CEREC SpeedFire to cool. It usually requires approximately 2 minutes before it has cooled sufficiently for use. The criterion for sintering is set using the mass ratio the manufacturer provides.

Glazing

A glazing spray, CEREC SpeedGlaze, was used to cover the restoration surface. The spraying process continued 10cm away from the restoration until it appeared to be of a uniform color. Before the glazing spray can be used, CEREC SpeedPaste is used as a filler in the restoration. The specimen with the firing pin is placed in the insulating layer of the firing chamber, which is programmed for glazing, and stops when the timer reaches zero.

Flexural Strength Determination

Flexural strength was calculated successfully with the aid of computer software and displayed as MPa together with the determination of the causes of failure. This was achieved by applying force at the center of the bridge (sphere diameter of $15mm^2$) at a crosshead speed of 0.5mm/min until it cracked. Before the force was applied, the restorations were fixed to the metal master model using Temp-Bond NE (Kerr, 50 g Base, 15 g Accelerator). This is a characteristic method to test strength as mandated by the ISO (#6872.770), and is carried out in a universal testing machine (Model 5566; Instron Corp, Canton, MA). Bluehill 2 software (Instron, Canton, MA, USA) was used to calculate the flexural strength.

Statistical Analysis

SPSS v. 26 was used in all analyses. Descriptive statistics (means, standard deviations, minima, and maxima for continuous variables, and counts and percentages for categorical variables) were obtained. Based on the Shapiro-Wilk test, there was no significant evidence of non-normality of the flexural strength data ($p > 0.05$). Based on Levene's test, there was no significant evidence that the assumption of homogeneity of variances was violated ($p > 0.05$). Therefore, the flexural strengths of the connectors with different thicknesses were compared via one-way ANOVA, with the Tukey-Kramer test used in post hoc comparisons. Fisher's exact test was used to compare the groups with respect to their mode of failure (connector failure or marginal failure).

Results

39 samples were milled, sintered, and glazed. One of the specimens broke during the flexural strength testing because of a technical error. Another two broke during the laboratory procedure while separating the metal sprue from the milled zirconia block, and one was lost during the sintering procedure as a result of operator error. A power calculation was conducted again using the final sample sizes (n =11 for the 3×3 mm connector thickness; n = 12 for the 3×4 mm connector thickness, and n = 12 for the 4×4 mm connector thickness), and based on these final sizes, it was confirmed that the power of the study was still greater than 99% with $\alpha = 0.05$.

Group A with connectors 3×3 mm thick had the highest mean flexural strength (1160.71 MPa, SD = 140.70) and Group C with connectors 4×4 mm thick had the lowest mean flexural strength (509.92 MPa, SD = 203.37 (Table 1). Post hoc tests using the Tukey-Kramer test showed significant differences between Groups A and B (p<0.001) and Groups A and C (p<0.001), but not Groups B and C (p=0.698). Side-by-side boxplots are shown in Figure 1.

In Group A, 8 samples broke because of connector failures ,and 3 fractured within the margins. The mode of failure in Group B differed slightly, as 10 samples broke via connector failure, and 2 fractured throughout the gingival margins. In Group C, 8 samples fractured across the connector site, and 4 failed because of a marginal fracture (Table 2). Accordingly, 74.3% of the fractures occurred in the FPDs' connector and 25.7% of the failures occurred in the restorations' margin. Based on Fisher's exact test, the difference between groups was not statistically significant. (p = 0.717).

Discussion

The purpose of this in-vitro research was to examine the effect on flexural strength of fast or short-cycle sintering of 3-unit zirconia FPDs with various connector sizes. Sirona's advertisement of the SpeedFire oven indicates that a high-quality restoration will be achieved after zirconia is sintered and glazed in the SpeedFire furnace. Although the literature lacked evidence that supported this claim, Almuwash et al.[46] considered this to be true and mentioned that the flexural strength of short-cycle sintered restorations is greater than that of those that use a long sintering protocol. However, previous studies have failed to investigate the way the new sintering protocol can affect zirconia restorations' mechanical strength. Accordingly, to address the previous claims, this study used Sirona SpeedFire sintering protocols to assess one of the zirconia FPDs' (with varying connector sizes) mechanical characteristics. The results showed significant differences between Groups A and B ($p<0.001$) and Groups A and C ($p<0.001$), but not Groups B and C ($p=0.698$). The samples with a connector size of 3×3 mm were found to have the greatest flexural strength (mean ± SD flexural strength 1160.71 MPa). Use of the new sintering protocol to fabricate zirconia FPDs with a larger connector size may not have outcomes as predictable as those of FPDs with smaller connector dimensions. However, because of this study's limitations, further investigations are required in this respect.

Given the modes of failure found in the results, it appears that one of the reasons for structural failure could be presence of immature crystal structures and grain boundaries' defects development resulted from the sintering cycle that consistent with Juntavee et al.'s work[45].

This research used the same material to fabricate each restoration, and the same STL file, milling, and sintering machine for milling and sintering. In addition, a similar universal testing machine was used to standardize the testing. Accordingly, the connector dimensions were the only factors that varied. Apart from that, there were also the considerations of the different maximum forces required before each sample failed.

Upon the recommendation of CEREC's manufacturer,[47] the 9 mm^2 dimension category showed the greatest flexural strength, while there was a slight decrement in the flexural strength when the connector dimensions were larger than 9 mm. The reason why the thicker connectors had lower flexural strength may be attributable to certain changes that may occur in zirconia's transformational phase during the sintering cycle. In the sintering process, the surface of the material absorbs the oven's heat and mature sintered zirconia is achieved through thermal conduction[45]. The increase in the connector dimension could interfere with the zirconia particles' maturation and impede their full shift from the monoclinic to tetragonal phase. This could be attributed to the fact that the shorter sintering cycle does not allow some of the heat necessary to penetrate into the specimen's bulk. Also, the yttrium particles migrate to the grain boundaries in response to very high temperatures. For connectors with larger dimensions and a greater surface area, a greater number of particles could migrate to the grain boundaries and increase the grain size, thereby adding the cubic phase to the structure and reducing the modulus of rupture. With the enlargement in the grain size, zirconia particles transform from their vulnerable tetragonal phase to the monoclinic phase that could alter the strength gradually[45]. However, given the limitations of this study, further investigations using Scanning Electron Microscope analysis will be needed.

With respect to the FPD, the margin is considered the weak component, which likely is the reason that the restorations' margin experienced 25.7% failure. This area breaks easily as it has the lowest restoration thickness. Øilo et al.'s research also supported this statement, as they found that the margin is the main weak point in a single-tooth restoration and the design and preparation of the restoration influences the fracture modes and rates greatly[48].

The flexural strength of three-unit FPD with connector sizes 3x3 mm sintered in the SpeedFire recorded in this study was much greater than the maximum bite force documented in different studies, and the findings of different studies support the result of this study. Abe et al.'s study demonstrated that the maximum bite force between the right lower and upper first molars was 76N (men) and 85N (women)[49]. D'Souza's research indicated that the normal masticatory bite force was 600N (axial MBF) and 225N axial[50]. Al-Omiri et al. reported a maximum bite force of 595.1 for the dentate side and 577.9N for the prosthesis side the implant supported[51]. Helkimo et al. found that there was an MBF of 382N (39Kg) in the molar area and 176N (18Kg) in the incisor area for men, while these values were 216N (22Kg) and 108N (11Kg) for women[52].

This research involved the use of an FPD design, and thus differed from prior studies that used the bar design. Unlike the FPD design, which is characterized by embrasures on the samples' occlusal, gingival, lingual, and buccal surfaces, the bar design is uniform. Accordingly, the connectors would become the locus of the forces. Onodera et al. assessed the way the connector's surface area affects a 3-unit FPD's flexural strength. In their research, they considered samples with 1:1, 3:4, and 2:3 shapes, and 9.0, 7.0, and 5.0 mm^2 surface areas, respectively. The 1:1 shape with a $9mm^2$ surface area demonstrated the maximum flexural strength values[53].

This research involved the use of In-Coris TZI C medi S A1, commercially available zirconia blocks that have been sintered already. CEREC also reports that they have a 98.1N Vickers indentation, 6-8 MPa fracture toughness, and 800-1000 MPa flexural strength[26]. All specimens in this study were subjected to 3-point bending tests to determine flexural strength.

The zirconia samples' flexural strength was assessed through static loading in this research. However, although this is not equivalent to the masticatory system, which involves dynamic loading, Itinoche et al. demonstrated that static loading is correlated with dynamic loading and found no significant differences between flexural strength in both static and dynamic loading tests[54].

With respect to the use of various techniques for fast sintering, Kim et al. assessed the relation between sintering and zirconia's translucency and grain size[22]. Their research involved 1, 2, 10, and 40h of microwave sintering with a holding time of 20mins. Their results showed that zirconia achieved greater translucency and a small grain size with short sintering[22]. Nevertheless, their research differed from this study, in that it used different technology and sintering durations.

Erosy et al. used the ultrafast speed oven and zirconia bars, and found the maximum flexural strength was 904.2 ± 115.7 MPa with the minimum sintering time (1580 ^{0}C for 10mins), while the other categories with longer sintering durations showed lower flexural strength[11]. However, theirs differed from this research, because this study used the SpeedFire oven and examined its influence on a 3-unit FPD's (with varying connector dimensions) flexural strength. This study's results have more clinical relevance as an FPD design rather than a bar was chosen for the investigation.

Hjerppe et al. investigated whether sintering duration affected zirconia discs' biaxial flexural strength. They used two different sintering conditions: (1) 1h holding time, 1h 40mins rise time, and 20-1500^{0}C temp, and (2) 2h holding time, 3h rise time, and 20-1500^{0}C temp. The results indicated that there was no great difference between the conditions, as the mean flexural strengths were found to be 1096.6 MPa (short cycle group) and 1074.7 MPa (long cycle group)[12]. This may have been the case because although these two sintering conditions differed, neither of them involved the same technology and conditions as in this research or in Ersoy et al.'s study[11]. Compared to the short cycle protocol Hjerppe et al.[12] suggested, this research involved a shorter sintering duration. Moreover, their samples had a disc design rather than the 3-unit bridge used here.

According to Almuwash et al.'s research, a 3-unit FPD has a flexural strength of at least approximately 505± 211MPa[46]. The outcomes of their study were not consistent with those in this research, although they considered using the same sintering protocol and identical zirconia (In-Coris TZI C medi S A1) as current study. However, in contrast to their conclusion, the outcome of this analysis supported using the SpeedFire furnace to fabricate a full zirconia restoration with connector size of 3x3 mm only.

All of the specimens used in the study were glazed after sintering, which may or may not affect their strength, as this subject is controversial. Manawi et al. claimed that glazed zirconia showed significantly higher flexural strength (385.4 ± 45.4 MPa) and fracture toughness (6.07 ± 1 MPa.m½) values than did ground, finished, and polished zirconia (302.4 ± 47.6 MPa and 2.14 ± 0.5 MPa.m½)[55]. However, Yener et al. disagreed, and asserted that glazed specimens' mean flexural strength did not differ statistically significantly[56]. Accordingly, further studies should be performed to investigate these discrepant findings.

This research had several limitations, and thus, the outcomes will not apply to all zirconia types, as the tests involved only one specific zirconia type, TZI In-Coris. In addition, this in-vitro research involved static loading, and there is greater accuracy and likelihood to reproduce the masticatory process with dynamic fatigue tests. Moreover, restorations may fail in the future, because saliva influences zirconia's aging greatly. Lastly, this research did not replicate the oral cavity by using a wet environment to assess zirconia's mechanical characteristics. To do so, future research should attempt to replicate the oral environment with a dynamic and wet environment. Based on the outcomes of this research, it can be concluded that using the SpeedFire oven is clinically acceptable, yields better results in FPD sintering, and is beneficial in formulating a monolithic zirconia bridge with connector size of not more than of 3x3 mm chairside. Moreover, future investigations using Scanning Electron Microscope analysis would explicate further the short sintering cycle's effects on zirconia restorations' mechanical properties.

Conclusions

1- The flexural strength of three-unit FPDs with connector sizes 3x3mm was significantly higher than that of three-unit FPDs with connector sizes 3×4 mm and 4×4 mm.

2- Three-unit zirconia FPDs with connector sizes of 3×3 mm that were sintered with the short sintering protocol had clinically acceptable flexural strength.

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APPENDICES

Appendix A: Tables

Appendix B: Figures

Appendix A: Tables

Connector Thickness	n	Mean	SD	Minimum	Maximum	*p*
3x3 mm	**11**	**1160.71**	**140.70**	**888.11**	**1438.80**	**<0.001**
3x4 mm	**12**	**565.20**	**145.91**	**325.16**	**850.38**	
4x4 mm	**12**	**509.92**	**203.37**	**180.80**	**785.95**	

Table 1: Flexural strength (MPa) by group.

Connector Thickness	n	Connector failure (Count; %)	Marginal Failure (Count; %)	*p*
3x3 mm	**11**	**(8; 72.7%)**	**(3; 27.3%)**	**0.717**
3x4 mm	**12**	**(10; 83.3%)**	**(2; 16.7%)**	
4x4 mm	**12**	**(8: 66.7%)**	**(4; 33.3%)**	

Table 2: Mode of failure by group.

Heating rate C^0/min	**Holding temperature C^0**	**Holding time min**
99	**750**	**0**
99	**1100**	**0**
50	**1510**	**30**
99	**800**	**5**

Table 3: Sintering protocols for zirconia specimens (Short cycle).

.

Appendix B: Figures

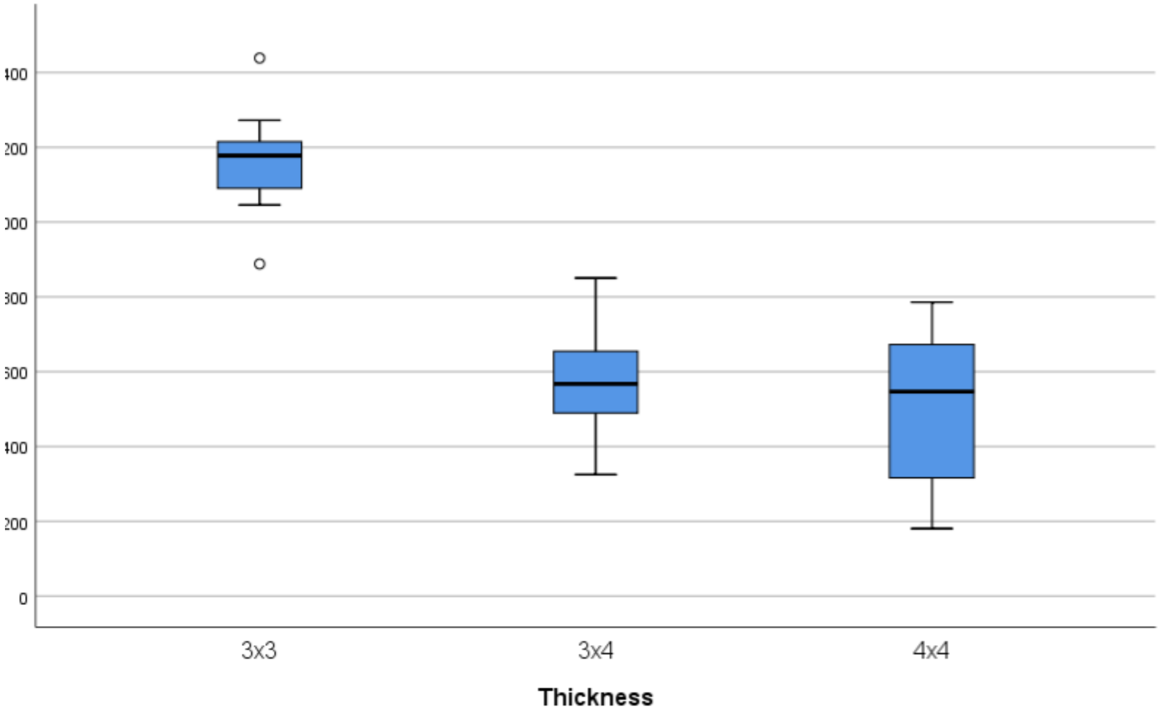

Figure 1: Side by Side Boxplots

Figure 2: Typodont Model Teeth # 29 and 31

Figure 3: Indicating Spray and Typodont Model

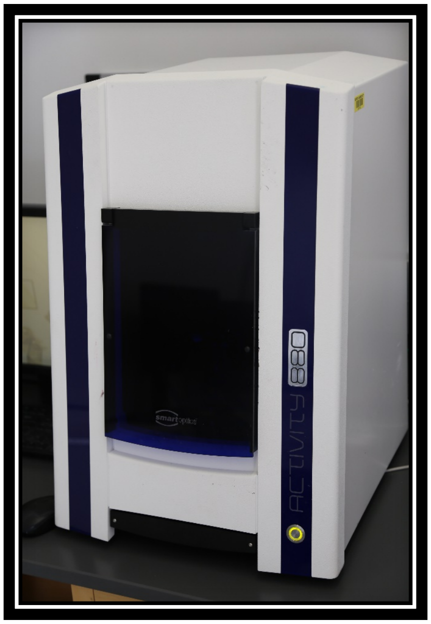

Figure 4: Activity 880 Scanner from Smart Optics.

Figure 5: Printed Metal Die.

Figure 6:CEREC Omnicam (left) and AC Bluecam from Sirona (right).

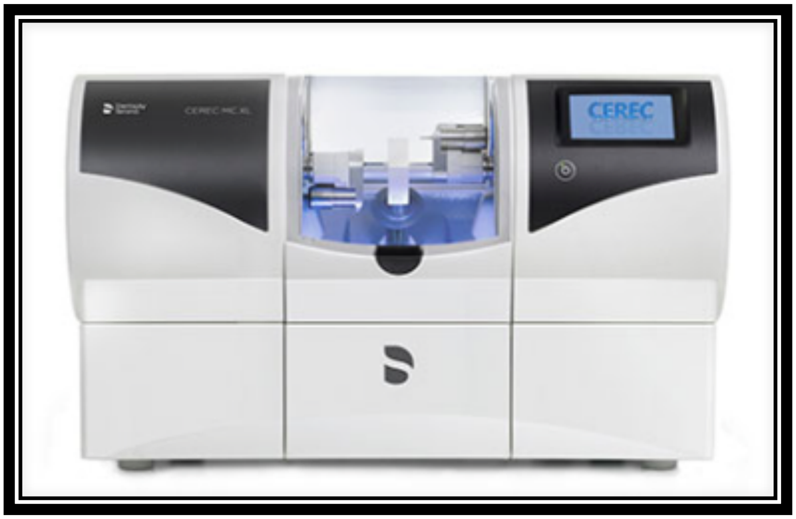

Figure 7: Sirona CAD/ CAM milling unit.

Figure 8: Milled and Sintered FPD

Figure 9: SpeedFire Furane

Figure 10: Marginal Fracture

Figure 11: Connector Fracture

Figure 12: Milled Block of TZI Incoris Zirconia

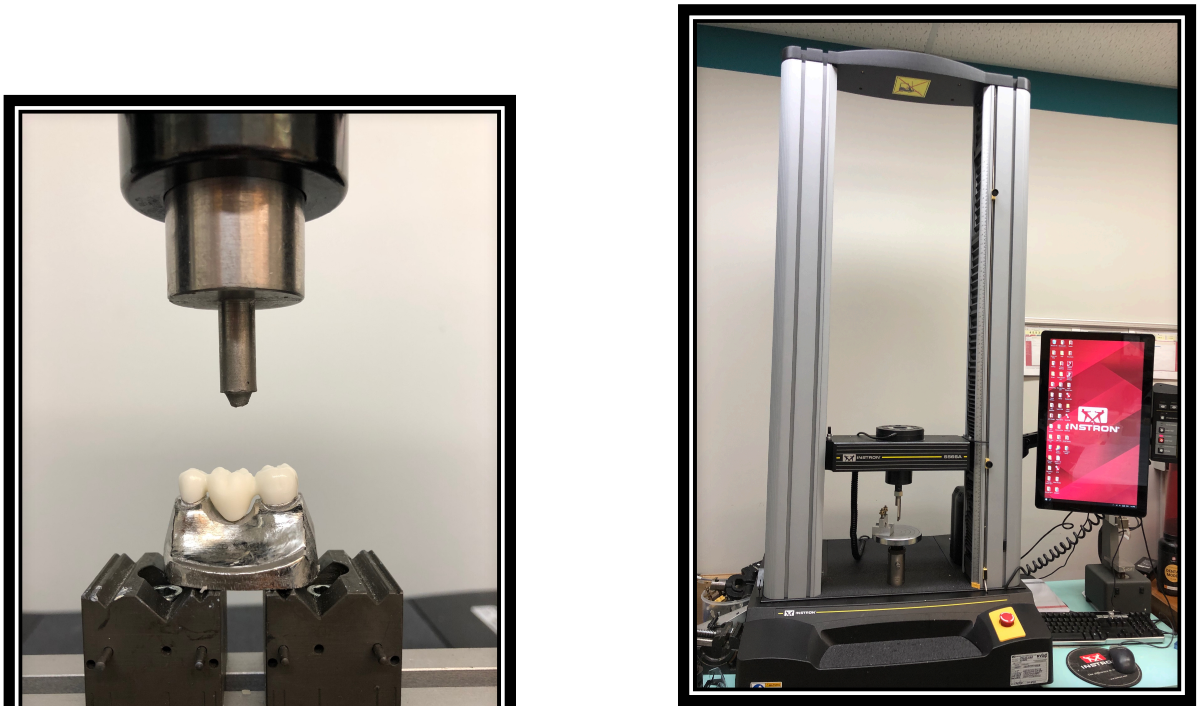

Figure 13: Universal Maschine