The Mechanical behaviour of crowns fabricated from Multilayered Zirconia for mandibular molars in bruxers using Finite Element Analysis (FEA).

by

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ABSTRACT

The mechanical behaviour of crowns fabricated from Multilayered Zirconia for mandibular molars in bruxers using Finite Element Analysis (FEA).

Purpose: To evaluate and compare the mechanical behaviour of the crowns made from Multilayered zirconia for mandibular molars by Finite Element Analysis (FEA) using the elastic modulus obtained from nanoindentation testing, and compare it with the reference Monolithic Zirconia crowns.

Null Hypothesis

H01: The elastic modulus of multilayered zirconia will not significantly differ from each other.

H02: The maximum principal stresses of multilayered zirconia crowns will not significantly differ from each other.

H03: The maximum principal stresses of multilayered zirconia crowns will not significantly differ from those of monolithic zirconia crowns.

H04: The occlusal thicknesses of 1 mm and 1.5 mm will not affect the maximum principal stresses in the crowns.

H05: The maximum principal stress in the crowns will not significantly differ on the application of axial vs horizontal loading.

Methods: Nanoindentation testing was performed on two Multilayer Zirconia materials, ZirCAD Prime and ZirCAD Prime Esthetic, to calculate the Modulus of Elasticity and Hardness. Two rectangular samples for both multilayer Zirconia, each with dimensions of 10 mm x 16 mm x 4 mm, were milled, finished, polished, and tested for Nanoindentation using Hysitron Nanoindenter. A force of 400 µN was applied to the Multilayer Zirconia samples using the Nanoindenter tip, which comprised layers of Enamel, Intermediate, and Dentine. Ten readings per layer were measured, and the average value was calculated. The result values were input for testing Finite Element Analysis of the crowns under different loading conditions. A typodont plastic tooth, a mandibular left first molar, was used as a primary model. The tooth was prepared for a full-coverage Zirconia crown with occlusal reductions of 1 mm and 1.5 mm, axial reductions of 1 mm, and a shoulder of uniform width of 1 mm. The tooth preparation was scanned using a Primescan intraoral scanner to create an STL file. A full-coverage Crown was designed using in-lab CAD software. Three-dimensional models of tooth preparation and full-coverage crowns were created and merged using HyperView Player 2022.3 software. The 3D model was created using 848135 nodes and 4567316 elements. Once the 3D meshes were obtained, a boundary condition (zero displacement) was applied to the external surface of the root.(Ausiello et al. 2004; Shahrbaf et al. 2013). The tooth-crown complex underwent a vertical loading and horizontal loading of 540N and 200N, respectively, in the central fossa, as well as the buccal cusps, and the Maximum principal Stresses were calculated in the Zirconia crowns.

The material properties were assumed to be homogeneous, isotropic, and linearly elastic. Adhesion was simulated as perfect without voids. The features from the periodontal ligament, pulp tissue, and cement thickness layer were not considered.

Results: The 1 mm thick crowns had more tensile stresses and maximum principal stresses compared to 1.5mm crowns, irrespective of the material of the crown. Horizontal forces generate more stresses in the crowns for both the thicknesses and all the materials of the crowns. Regarding the Zirconia type, all types of Zirconia exhibited similar behaviour under the same loading conditions and crown thickness. The Maximum Principal stress values induced in crowns in all the combinations of variables ranged from 90.4 to 154.4MPa, and Maximum Principal stress values induced in the tooth structure in all the combinations of variables ranged from 1.2 to 22.9 MPa.

Conclusion: Increasing the thickness of the crown material reduces the stresses transmitted to both the crown and the tooth structure. More stresses were observed in the crown-tooth assembly under horizontal loading compared to axial/vertical forces. No statistically significant differences in Maximum principal stresses were observed between the different Zirconia crown materials. Hence, Multilayered Zirconia (ZirCAD Prime and ZirCAD Prime Esthetic) behaved similarly to Monolithic Zirconia under maximum biting forces and bruxing forces induced by Finite Element analysis.

Dedication

I dedicate this thesis to my husband, whose steadfast support, patience, and unwavering belief in my abilities have been instrumental throughout my residency. His encouragement and love have provided me with the strength and resilience to persevere through the most demanding moments of this journey.

To my family, I extend my deepest gratitude for their continuous support, positivity, and unconditional love. Their faith in me and their constant presence have been a source of comfort and motivation, enabling me to pursue my goals with confidence and determination.

Above all, the Gracious God, without His Divine presence, this accomplishment would not have been possible.

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NOMENCLATURE

3Y- TZP	3mol% Yttria stabilized tetragonal zirconia polycrystal
4Y-TZP	4mol% Yttria stabilized tetragonal zirconia polycrystal
5Y-TZP	5 mol% Yttria stabilized tetragonal zirconia polycrystal
PSZ	Partially stabilized Zirconia
Y2O3	Yttrium Oxide
ZrO2	Zirconium Oxide
Er2O3	Erbium Oxide
Fe2O3	Ferric sulphate
LTD	Low-temperature degradation
CIELAB	International Commission on Illumination
UTML	Ultra Translucent Multilayered Zirconia
STML	Super Translucent Multilayered Zirconia
ML	Multi-layered Zirconia
F-SEM	Field Emission scanning electron microscopy
OP	Opalescence parameter
AT %	Average transmittance
TP	Translucency parameter
CR	Contrast ratio
PXRD	Powder X-ray Diffraction
MWZ	Microwave-speed sintered Zirconia
FC	Full contour (FC)

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CHAPTER 1

1. INTRODUCTION AND BACKGROUND

As demand for enhanced aesthetics increases, the use of dental ceramics as a restorative material has been growing. Yttria stabilized tetragonal zirconia polycrystal (Y-TZP) has evolved tremendously in the past few decades (Schönhoff et al. 2021). The first dental Y-TZP introduced is zirconia stabilized with 3 mol% yttria content (3Y-TZP). It has excellent mechanical properties but poor optical properties due to its opaque white colour. For this reason, it was mainly used as a framework to be veneered with esthetic ceramics.

The name Zirconium comes from the Arabic Zargon (golden colour), which in turn comes from two Persian words: Zar (Gold) and Gun (Colour)(Piconi and Maccauro 1999). Zirconia ceramic is of paramount importance in dentistry due to its high fracture toughness. It is a polymorphous material that can exist in three crystal structures: monoclinic, tetragonal, and cubic. The monoclinic phase is only stable at room temperature; however, its mechanical and optical properties are insufficient for clinical use. Hence, stabilizing the tetragonal and cubic phases is required and can be achieved using other stabilizing oxides, with yttrium oxide (Y2O3) being the most widely used stabilizer. During fracturing, the predominant tetragonal particles can transform into monoclinic particles, which are larger in volume than the tetragonal particles, thereby forming compressive stresses around the crack tip and preventing the crack from propagating. This phenomenon is known as fracture toughening. The yttria content within the zirconia has a significant impact on its mechanical and optical properties. Increasing the yttria content generally increases translucency but decreases flexural strength in zirconia (Martins et al.).

The first-generation 3Y-TZP contains 3 mol% yttrium and 0.25 wt% aluminum oxide, the most robust ceramic, with flexural strength up to 1200 MPa. The second generation had a reduced 0.05 wt% aluminum oxide content with increased translucency and the same mechanical properties, but was still not suitable for esthetic sites. To further improve the translucency, a third generation, 5Y-TZP and a fourth generation, 4Y-TZP monolithic Zirconia were developed in which the yttria content was increased to 5 mol % and 4 mol %, respectively, and aluminum oxide was reduced to 0.05 wt % which result in an increase in the cubic phase and reduction in the flexural strength and fracture toughness as the stress-induced phase transformation cannot take place (Schönhoff et al. 2021).

Monolithic zirconia restorations have gained significant attention due to their reduced porcelain chipping, less tooth reduction, and lower costs. It is essential to note that “Monolithic Zirconia” refers to generations of Zirconia restorations with varying mechanical and optical properties, resulting in differing clinical performance. This includes increasing the yttria content from 3 mol% (3Y-TZP) to 4 mol% % (4Y-TZP) and 5 mol% (5Y-TZP) for higher translucency and introducing polychromatic(multi-layered) Zirconia.

The multi-layered technology (colour gradient) introduced in 2015 mimics the shade gradient of natural teeth using pigmentation only. Each blank has the same generation of zirconia, with no difference in the flexural strength between the enamel and dentine layers.

However, the colour/shade gradient multilayered zirconia did not resolve the issues with the strength and translucency of conventional dental Zirconia. To overcome this issue, a strength gradient multilayered yttria-stabilized zirconia (YSZ) has been developed, integrating different generations of zirconia in a single blank, aiming to merge the advantages of both generations of zirconia. It comprises two types of zirconia combined in a single restoration, with an intermediate zone following a gradual transition between them.

The interfacial strength of multi-layer zirconia was tested using a modified 4-point bending test aided by Finite Element Analysis. It was found that the cross-sectional multi-layer beam exhibited 30% smaller flexural strength than Enamel and Dentin. The weakness of the interface could be due to the multi-layer production process, as the layers are produced by pressing ZrO2 powder, which is likely to create interfacial defects such as large impurities and/ or poorly sintered regions (Kaizer et al., 2020). It is important to note that this interfacial weakness varies depending on the brand and formulation of zirconia used. With ongoing advancements in material processing and sintering technologies, it is anticipated that future generations of multi-layer zirconia will exhibit improved mechanical integrity and reduced interface-related vulnerabilities.

While fabricating fixed bridges using the multilayer Zirconia, the positioning of the connector should be given importance as the influence of sintering and positioning the connector in a multilayer blank ZirCAD Prime (top, centre and bottom) on the fracture forces of 3-unit fixed partial dentures showed that the highest mean fracture force was found with the longest sintering (9 hours 50 minutes) in the central position of the blank. However, these effects were not statistically significant, but should be considered during production (Rosentritt et al. 2022).

Teeth and dental restorations are subjected to mechanical, chemical, and thermal stresses, which contribute to wear and aging, ultimately leading to the failure of the restorative material. New restorative material should be selected based on clinical evidence; however, the complexity and cost of clinical trials are high. Zirconia manufactured by different manufacturers has different compositions at similar costs. Conducting clinical trials for all of them before using them clinically would be difficult, time-consuming and complex. Therefore, in vitro tests are a suitable alternative for testing and utilizing materials in clinical scenarios (Abdulmajeed et al.). On the other hand, the Finite element analysis is an innovative technique for evaluating the stress-strain relationship, and it has been widely used in dentistry. It comprises a series of computational algorithms that calculate stress and strain within a model in response to external forces, pressure, thermal, and magnetic fields, as well as other factors. It helps to estimate the biomechanical characteristics of dental prosthesis and supporting tissues, which is difficult to measure in vivo (Murakami and Wakabayashi 2014).

FEA models can precisely calculate material stress under conditions of geometry and boundaries that accurately represent clinical reality. Anatomical geometry can be obtained using various approaches, including literature data, 3D scanners, and computed tomography (CT) scans. Regardless of the image data source, the anatomic model is converted into a parameterized model by CAD software and then into a discretized model by the FEA software. With FEA, the stress distribution within a structure can be calculated by dividing the structure into elements, and their ends meet to form nodes (Dejak, Młotkowski, and Langot 2012; Syed et al. 2021).

Regarding the thickness of the Zirconia crowns, there is no consensus on tooth preparation guidelines for monolithic zirconia restorations. Manufacturers recommend a minimum reduction of 0.7 mm for full-coverage zirconia restorations without distinguishing between the different Zirconia types. Decreasing the yttria concentration and increasing the thickness of the material significantly increases the fracture load of Zirconia. The recommended minimum thickness for 3Y-PSZ is 1mm, 4Y-PSZ is 1.2mm, and 5Y-PSZ is 1.5mm (Abdulmajeed et al. 2020a).

In the present study, nanoindentation testing and FEA analysis were conducted to evaluate the mechanical behaviour of a multilayer Zirconia crown on a natural tooth under maximum biting forces of 540 N and 200 N of horizontal bruxing load. The material properties were assumed to be homogeneous, isotropic, and linearly elastic. Adhesion was simulated as perfect without voids. The features from the periodontal ligament, pulp tissue, and cement thickness layer were not considered. An axially directed load of 540 N (Calderon et al), which is equivalent to the maximum biting force, and a horizontal bruxing load of 200 N (Wicaksono et al.) were applied in the central fossa and the buccal cusps and the mechanical behaviour of the tooth-crown complex was investigated.

2. LITERATURE REVIEW

2.1 IMPROVING ESTHETICS AND MECHANICAL PROPERTIES

Metal-ceramic restorations have been the gold standard in prosthetic dentistry for decades, offering successful long-term outcomes. However, increasing demands for improved aesthetics have led to the development of tooth-coloured restorative materials. Recently, attention has focused on ceramics as the material of choice. Ceramics are classified into four categories: 1) Silicate ceramics (used for pottery, tiles, bricks), 2) Oxide Ceramics (Zirconia), 3) Nonoxide ceramics (impractical for use in dentistry), and 4) glass ceramics (Feldspathic Ceramics)(Anusavice 2013).

Achieving excellent esthetics in dentistry heavily relies on adequate translucency—the ability to allow light to pass through without significant scattering. In a natural tooth, light interaction with enamel involves various processes: reflection, absorption, transmission, and diffusion. The material's translucency or opacity is influenced by how light is transmitted through it. For Translucent materials, most light is diffusely transmitted with minimal scattering. For Opaque materials, most of the light is diffusely reflected and scattered, resulting in the material appearing opaque. Striking the right balance between translucency and opacity is crucial for creating dental restorations that seamlessly blend with the surrounding natural teeth (Reale Reyes et al. 2023).

High-strength glass ceramics, such as lithium disilicate, combine pleasing aesthetics with high strength, approximately half that of Y-TZP. There are many clinical situations where a material with adequate strength is needed to withstand loads and wear, e.g. in posterior areas and cases of bruxism. In such situations, monolithic Y-

TZP could be a promising treatment alternative, offering adequate strength and potential for improved esthetic performance (Johansson et al. 2014).

2.2 ALL CERAMICS

Modern ceramic systems can be broadly categorized into two groups: those that are translucent and those that consist of an opaque, high-strength core onto which an esthetic layer of ceramic must be applied to achieve a natural appearance. Examples of translucent materials are conventional sintered feldspathic porcelain fabricated on refractory dies, platinum foil, and pressable ceramics. Examples of opaque layered materials include non-metallic restorations made with alumina, zirconia, or lithium disilicate, as well as high-strength core materials.

While Leucite and feldspathic glass ceramics are indicated for onlays, three-quarter crowns and veneers, Lithium disilicate glass ceramics can perform successfully in the posterior segment for single crowns and 3-unit FPDPs in the anterior area. Zirconia modified alumina is ideal for posterior crowns and FPDPs, while densely sintered alumina is recommended for veneers, crowns and anterior FPDPs. The successful application of all-ceramic materials depends upon the ability of clinicians to select the material appropriately, use proper manufacturing techniques and cementation or bonding procedures so that it withstands the intraoral conditions and satisfies the esthetic demands of the patient (Conrad, Seong, and Pesun 2007).

Regarding tooth reduction, clinicians can use translucent materials with more conservative tooth preparation than opaque, layered systems. Optically, the translucent materials are usually more esthetic than the layered materials. Most translucent restorations must be bonded to improve their predictability, while layered restorations do not have this sensitivity to the choice of luting agents.

Clinicians should reserve dental ceramics with high translucency for clinical applications that require high-level esthetics. Ceramics with high strength tend to be more opaque and pose a challenge when trying to match natural tooth colour, but they can mask discoloration when present (Spear and Holloway 2008).

2.3 ZIRCONIA

Zirconia ceramic is a versatile and durable ceramic material that has gained widespread acceptance in dental applications due to its high flexural strength, fracture toughness, and dimensional stability. It is polymorphous and can exist in three crystal structures: monoclinic, tetragonal, and cubic. The monoclinic phase is only stable at room temperature; however, its mechanical and optical properties are insufficient for clinical use. Hence, stabilizing the tetragonal and cubic phases is required and can be achieved using other stabilizing oxides, with yttrium oxide (Y2O3) being the most widely used stabilizer. The yttria content within the zirconia has a significant impact on its mechanical and optical properties. Increasing the yttria content generally increases the translucency but decreases the flexural strength of the zirconia (Martins L et al, 2023) (Labetić et al. 2024).

Zirconia stabilized with a 3 mol% yttria content (3Y-TZP) is the first dental Y-TZP (yttria-stabilized tetragonal Zirconia polycrystal) with extraordinary mechanical properties and excellent biocompatibility (Kassardjian et al., 2016). This type of ZrO2-based material is known as yttria-stabilized tetragonal zirconia polycrystals (Y-TZP) and is commonly used as a structural ceramic. Stabilizing Yttria cation fine-tunes Y-TZP properties such as translucence, toughness and hardness. The same stabilizing effect can be achieved by adding Mg2+, but the resulting Mg-TZP ceramics have inferior strength and fracture toughness compared to Y-TZP (Labetić et al. 2024).

Another advantage of Y-TZP is its compatibility with computer-aided design (CAD) and computer-aided manufacturing (CAM) technology. As a result, Y-TZP has become the material of choice for fabricating all high-quality ceramic dental restorations. Additionally, CAD/CAM enables the shaping of materials in various ways, promoting the development of diverse Y-TZP frameworks (Labetić et al. 2024).

Monolithic Zirconia crowns exhibit a higher fracture load than monolithic lithium disilicate, layered zirconia, and metal-ceramic crowns. The fracture resistance of monolithic zirconia crowns with a thickness of 1 mm can be equal to that of metal-ceramic crowns. Increasing the thickness of the monolithic zirconia core to 1.5 mm increases the material's fracture resistance (Sun et al. 2014).

The fracture load of Zirconia is proportionally related to the material thickness. At 0.7mm thickness, 3Y-TZP zirconia survived masticatory simulation; however, for 4Y-PSZ and 5Y-PSZ, a minimum thickness of 1.2 mm is required to withstand mastication simulation with 1.2 million cycles at 110N load and thermal cycling at 5-55 degrees C (Abdulmajeed et al. 2020b).

The phase transition from the tetragonal to monoclinic phase of Y-TZP under hydrothermal and/ or mechanical loading is well known and is called low temperature degradation (LTD) (Cales and Drouin 1999). Sandblasting can transform the tetragonal ZrO2 phase into a monoclinic phase, which affects the mechanical properties and stability of Y-TZP materials. During fracturing, the predominant tetragonal particles in 3Y-PSZ can transform into monoclinic particles, which are larger in volume than the tetragonal particles, thereby forming compressive stresses around the crack tip and preventing the crack from propagating. One drawback of 3Y-TZP is its optical properties due to its whitish colour and high opacity. Therefore, it was mainly used as a framework veneered with translucent esthetic silicate ceramics. A common problem for veneered zirconia restorations is the chipping of the veneering ceramic (Martins et al. 2023).

2.3.a MULTI-LAYERED ZIRCONIA

New generations of monolithic zirconia have recently increased the yttria content to approximately 4 and 5 mol%. These ceramics contain a higher proportion of cubic ZrO2 polycrystals, which contribute to enhanced optical properties. The different translucency of the layers within the blank is intended to produce restorations with similar optical properties to those of a natural tooth structure (Rosentritt et al. 2022). It is essential to note that "Monolithic Zirconia" refers to generations of Zirconia restorations with varying mechanical and optical properties, resulting in differing clinical performance. This includes increasing the yttria content from 3 mol% (3Y-TZP) to 4 mol% % (4Y-TZP) and 5 mol% (5Y-TZP) for higher translucency and introducing polychromatic(multi-layered) Zirconia.

The multi-layered technology (colour gradient) introduced in 2015 mimics the shade gradient of natural teeth using pigmentation only. Each blank has the same generation of zirconia, with no difference in flexural strength between the enamel and dentin layers (Martins et al., 2023).

However, the colour/shade gradient multilayered zirconia did not resolve the issues with the strength and translucency of conventional dental Zirconia. To overcome this issue, a strength gradient multilayered yttria-stabilized zirconia (YSZ) has been developed, integrating different generations of zirconia in a single blank, aiming to merge the advantages of both generations of zirconia. It comprises two types of zirconia combined in a single restoration, with an intermediate zone following a gradual transition between them (Martins L et al., 2023; Vardhaman S et al., 2020).

2.3.b MECHANICAL AND OPTICAL PROPERTIES

The relationship between translucency and flexural strength of translucent zirconia ceramics was investigated. The biaxial flexural strength values were found to be inversely related to the translucency parameter (Reale Reyes et al. 2023).

Pigments such as Er2O3 and Fe2O3 can be added to achieve zirconia tooth-like colour. Besides the yttria concentration, differences in pigment concentrations between each layer influence the mechanical and microstructural properties of the prosthesis. 3Y-TZP specimens were shaded using infiltration or powder mixing methods and then treated in an autoclave to induce Low Temperature Degradation (LTD). The formation of the monoclinic phase was significantly lower in the tooth-coloured 3Y-TZP compared to the non-coloured material. Therefore, the tooth-coloured zirconia has comparable biaxial flexural strength but shows greater resistance to LTD, regardless of the colouring technique.

Metallic oxides such as ferric oxide, erbium, and neodymium were considered contaminants because they were found to influence the ceramic microstructure and surface hardness (Čokić et al. 2022). However, few studies have shown that colouring liquids do not affect flexural strength or hardness, but alter the optical properties (Yu and Park 2019).

Different shrinkage rates were observed in the colour gradient monolithic zirconia, with varying shrinkage rates among different colour layers. It was found that 5YZ was more resistant to low-temperature degradation (LTD) in terms of phase transformation than 3YZ. However, a certain amount of monoclinic phase was detected even in 5YZ after 50 hours of autoclaving, which decreased its flexural strength and reliability. Two distinct regions were observed in 5YZ- a dominant matrix with large grains and scattered areas with small grains. It was found that small grains were strongly affected by LTD and represented a fracture origin (Harada et al. 2020).

There are differences in susceptibility to Low Temperature Degradation (LTD) between the enamel and transition layers of multilayered zirconia (Katana Zirconia ML). The amount of phase transformation of zirconia due to LTD is highly dependent on the kind of shading. However, no significant changes in the biaxial flexural strength with LTD could be revealed (Wille et al. 2018).

The mechanical and optical properties of novel strength and colour gradient multilayered zirconia, as well as the impact of conventional and high-speed sintering, were analyzed. Three multilayered zirconia materials were analyzed: Katana Zirconia Block STML (KZC)- 4Y-TZP, Katana Zirconia Discs STML (KZL)-4Y-TZP and IPS e.max ZirCAD Prime (EZL)- a combination of 5Y-TZP and 3Y-TZP layers. The control group was the crystallized lithium disilicate ceramic IPS e.max CAD HT (ELC). KZC was high-speed sintered, KZL and EZL were conventionally sintered, and ELC was crystallized according to the manufacturer's instructions. Monolithic single molar crowns were fabricated, and half of them were aged in the chewing simulator, and fracture load was tested under the Universal testing machine. The results showed that zirconia revealed higher mechanical properties than Lithium disilicate; however, it has less translucency. High-speed sintering did not negatively influence the fracture load or wear properties of multilayered zirconia. Strength gradient multilayered Zirconia, EZL, showed a higher fracture load than colour gradient multilayered Zirconia- KZC and KZL.

Hence, the strength-gradient material was less affected by artificial aging than the colour-gradient material (Michailova et al., 2020).

The survival rates and fracture resistances of four types of zirconia were compared after undergoing thermocycling and/or thermomechanical loading. Multi- layered 5Y-PSZ/3Y-PSZ, Multi-layered 5Y-PSZ/4Y-PSZ, monolithic 4Y-PSZ and Monolithic 3Y-PSZ as control. The results of the study showed that yttria content significantly affected the fracture resistance of the crowns. The mean fracture resistance from highest to lowest: 3Y-PSZ and 4Y-PSZ, followed by the two multi-yttria-layered systems. The mean difference between the two multi-yttria-layered systems was not statistically significant. The fracture resistance of multi-layer Zirconia crowns is determined by the amount of the weaker zirconia phase at the occlusal part of the restoration rather than enforced by the stronger zirconia at the cervical part of the crown. Milling the gradient zirconia systems can result in different compositions and concentrations of each layer. The technician can control this by selecting the location where the restoration will be milled. Thus, gradient Zirconia can be used as a universal crown material, depending on the aesthetic and mechanical needs of each situation (Badr et al. 2022).

Several authors have expressed concern about the long-term degradation (LTD) or aging associated with transforming the tetragonal to the monoclinic phase. It has been demonstrated that LTD reduces the mechanical properties of stabilized ceramics over time in a humid environment through surface roughening, grain pull-out, and microcracking. Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) ceramic exhibits LTD in a humid environment between 200 °C and 3000 °C and can occur in vivo. This hydrothermal aging of Y-TZP causes phase change from tetragonal to monoclinic, which leads to a volume increase of 5%; fine microcracks start on the surface and progress inward (Flinn et al. 2012).

The effect of low-temperature degradation (LTD) on colour, translucency and hardness of layers of multilayered Zirconia, 3M Lava Esthetic was analyzed. Ninety discs were divided into six groups based on LTD treatment, no LTD treatment and inherent layer type (incisal, transition and body). The hardness of each specimen was measured using a microVickers hardness tester. The CIE lab colour coordinates were measured using a spectrophotometer. It was found that LTD did not affect the optical properties or hardness of multilayered Zirconia. The study showed that LTD treatment did not affect optical properties and hardness of multilayer Zirconia. A significant difference was found in the hardness between the layers of multilayered zirconia. LTD treatment did not change the b* or L* values but did decrease the a* values. On LTD treatment, the colour difference between the layers decreased (Park 2023).

The effect of LTD on newly developed yttria gradient multilayered zirconia with a gradient showed an increase in roughness, the occurrence of microcracks, a decrease in hardness and elastic modulus, and an increase in surface compressive residual stress. In addition, the fatigue strengths of multilayered zirconia crowns were between 3Y and 5Y, and the absolute fatigue strength decreased after LTD (Li et al. 2024).

The transformation of yttrium-stabilized tetragonal zirconia (Y-TZP) to a monoclinic phase and the change in flexural strength of the various layers of multilayered Zirconia (enamel, transition 1,2, and body layer), four layers of Katana Zirconia ML, Emax. ZirCAD and Lava Plus are compared to two conventional Zirconia ceramics before and after hydrothermal aging in an autoclave for 5,10, 15, and 20 hours. There was a significant difference in the transformation ratio for the tested materials and between the four layers of multilayered Zirconia, ranging from 9.9 vol% (LAVA Plus) to 44.2 Vol % (Katana Zirconia MI enamel layer). However, thermocycling aging or low-temperature degradation had no significant influence on the flexural strength of any material. Both test methods, the ball-on-three-ball test and the piston-on-three-ball test, provide reliable results. The flexural strength measured by the ball-on-three-balls test was 20% higher than that measured by the piston-on-three-ball test, as it is less dependent on the surface roughness, so that it might be preferred for investigations adapted to clinical reality (Wille et al. 2018).

The chemical composition and microstructure of three multi-layered Zirconia materials and their constituent layers- Ultra Translucent Multilayered Zirconia (UTML), Super Translucent Multilayered Zirconia (STML) and Multi-layered Zirconia (ML) using X-ray fluorescence, X-ray Diffraction and Field Emission scanning electron microscopy (F-SEM) were investigated. They also evaluated their low-temperature degradation (LTD) resistance and translucency properties. The results revealed that UTML (5 mol%) has 75 wt% cubic content and 4.05µm average grain size, STML (4 mol%) has 65% cubic content and 2.81 µm grain size, and ML (3 mol%) has < 50 wt% cubic content and 0.63 µm average grain size. After water aging, more monoclinic content was found in ML. UTML and STML did not show a detectable monoclinic phase. The translucency was similar among layers and between UTML and STML, which was superior to ML. The difference between each material's enamel and dentin layers is only pigment composition, resulting in colour difference but not translucency (Kolakarnprasert et al. 2019).

The flexural strength, translucency parameter, bond strength, and enamel and material wear of 5Y-TZP (Katana UTML shade A1) with 3Y-TZP (Katana HT) and Lithium Disilicate (e.max CAD). It was found that 5Y-TZP has a flexural strength and translucency parameter between those of 3Y-TZP and lithium disilicate. The short-term and long-term bond strength of 5Y-TZP and 3Y-TZP was similar to that of lithium disilicate. 5Y-TZP demonstrated no measurable material wear and opposing enamel wear similar to all other materials tested (Kwon et al. 2018).

The mechanical properties of different layers of multi-layered Zirconia materials—IPS e.max ZirCAD Prime, Optimill Multilayered 3D, and Ceramill Zolid fx multilayered were divided into two groups with and without thermal cycling and were compared. The results showed that Optimill and Ceramill showed less degradation of flexural strength than ZirCAD, but their maximum values were lower than those for ZirCAD. Also, the positioning within the blank affected the mechanical properties of multi-layered Zirconia. For ZirCAD, nesting the restorations in a cervical position as much as possible positively impacts flexural strength (Winter et al. 2022).

The difference in the post-fatigue load-bearing capacity and colour of restorations milled at different axial cutting depths in multilayer Zirconia CAD/CAM discs was evaluated. The restorations milled at different cutting depths depended on the durability and esthetic requirements. Posterior restorations can be milled from the middle or bottom of the disc to improve the load-bearing capacity. Anterior restorations can be milled from the entire thickness of the multilayer disc to meet the esthetic requirement (Li et al. 2021).

The composition, microstructure, and wear properties of a multi-layer translucent Zirconia, MULTI, IPS e.max ZirCAD Multi, were evaluated in comparison to the conventional 3Y-TZP, IPS e.max ZirCAD LT. They found that the wear pattern in multi-layered Zirconia was rougher than the control due to extensive subsurface lateral cracks, which result in localized material spalling and lead to greater volume loss and wear depth. The grain size and cubic content of multi-layered Zirconia decrease gradually from the enamel to the dentine layer; the presence of different layers does not influence its wear behaviour (Vardhaman et al. 2020).

The minimum layer thickness of 0.5mm is sufficient for 3Y-TZP to withstand chewing forces; the minimum thickness for 4Y-PSZ and 5Y-PSZ is controversial. 5Y-PSZ occupies an intermediate position between conventional zirconia and lithium disilicate as it has lower translucency and a higher flexural strength than lithium disilicate. As lithium can adhesively bond, the two materials showed comparable fracture loads (Pöppel et al. 2022).

Zirconia crowns can significantly wear down lithium disilicate restorations. Specifically, 3Y and 5Y zirconia (CERCON HT, XT) antagonists have been found to wear down lithium disilicate more than any other ceramic material tested. 3Y zirconia caused more wear than 5Y zirconia despite the similar hardness of these materials. Wear of lithium disilicate is caused by microcrack propagation along the interface between the glass matrix and lithium disilicate crystals. Therefore, if an existing zirconia crown opposes a prepared tooth, lithium disilicate might not be the ideal material to restore that tooth (Shah et al. 2024).

The relationship between monolayer and multilayer zirconia grades and their physical, mechanical, and optical properties was investigated. The microstructural and phase characteristics of five available 3Y-TZP Zirconia materials (GC ST, GC HT, Katana ML, Katana HT and Lava Plus) and three Y-PSZ materials (Katana STML, Katana UTML, and GC UHT) were tested. They found comprehensive correlations between chemical composition, microstructure, and optical properties. µRaman spectroscopy can provide information about the presence of colouring agents. All tested dental Zirconia had a Vickers hardness of around 13 GPa irrespective of the zirconia grade. 3Y-TZP materials showed lower mechanical reliability (Weibull modulus) than 4Y-PSZ, except for GC HT, but were more reliable than 5Y-PSZ and 6Y-PSZ. 6Y-PSZ and 5Y-PSZ Zirconia had a higher translucency than 4Y-PSZ and 3Y-TZP, but these values were still lower than for Lithium disilicate glass ceramics (Čokić et al. 2022).

Multilayer Zirconia with different yttria content, 3 mol%, 4 mol% (M3-4YSZ), and 4 mol%, 5 mol% (M4-5YSZ), were tested for relative density, crystalline phases, microstructure, Vickers hardness, fracture toughness, and optical properties. No statistical variation in hardness and relative density was observed in different layers of the material. Colour variations are gradual and gently perceptible, thus improving the final aesthetic effects. An increase in the tetragonal phase increased fracture toughness (De Lima Gomes et al. 2024).

The influence of yttria content on the microstructure, physical and mechanical properties of three commercially available polychromatic multilayered Zirconia: Dima Mill Zirconia ML, VITA YZ/ST multicolour and VITA YZ/XT Multicolour with 3,4 and 5 mol% % yttria, respectively, was compared. They found all zirconia specimens displayed homogeneous, dense micro and nanostructures without pores or defects. The differences in yttria content have no impact on their function in the oral cavity, both in terms of their fracture propensity and the damage that can be caused to the opposing teeth. They also concluded that milling technology is reliable for constructing fixed prosthetic restorations from this material (Dimitriadis et al. 2024).

The cross-sectional multilayer specimens have lower fracture resistance than homogeneous beams due to the weakness in the intermediate layers, resulting from the layered manufacturing process, and also due to the influence of different colouring oxides. As a result, homogeneous enamel and dentine beams fractured near the center of the beam, while multi-layer beams broke at the Enamel/ T1 interface. This could be due to the pressing technique as it creates interfacial defects making interfaces prone to fracture. The cross-sectional beam of multi-layered

Zirconia exhibited 30% less flexural strength than the Enamel and Dentin homogenous beams (Kaizer et al. 2020).

Crystallography, translucency, phase content, microstructure, and flexural strength of two commercial strength-gradient multilayered dental zirconia grades- KATANA Zirconia YML and IPS e.max ZirCAD Prime. Katana Zirconia YML comprises four layers: enamel, body 1, body 2, and body 3. IPS e.max ZirCAD Prime comprises three layers: enamel, transition, and body. These seven individual layers were investigated. They found that enamel layers exhibited a higher yttria content, higher cubic ZrO2 fraction (50wt%), higher translucency, larger grain size, and lower 4-point and biaxial flexural strength. The body layers with lower yttria content resulted in a higher tetragonal ZrO2 fraction (68-82%), lower translucency, smaller grain size, and higher 4-point and biaxial flexural strength. The interfaces between the different layers appeared gradual without sharp boundaries, and they were proven to be strong enough not to form the weakest link under flexural testing conditions (Inokoshi et al. 2023).

The flexural strength of different layers of multi-layered Zirconia, IPS e.max ZirCAD MT Multi (4Y,4/5Y, 5Y) under monotonic load application using two test devices (three-point bending and ball in hole), fatigue strength under cyclic load application, hardness and fracture toughness and Finite element analysis (using manufacturer values) were investigated. Three groups were formed based on the region in the blank- 5-YSZ: Incisal Zone, 4/5-YSZ: Transition/ Intermediate Zone, and 4-YSZ: Cervical Zone. They concluded that different layers of multi-layered Zirconia blank presented distinct mechanical properties. 4-YSZ (cervical layer) presented the highest flexural strength under monotonic and cyclic loads (fatigue) and higher fracture toughness, like the transition layer (4/5-YSZ). Hardness was similar between the layers. The ball-in-hole device performed equally well as a three-point bending device and can be used as an alternative testing mechanism (Machry et al. 2022).

The differences in surface characteristics, roughness, topography, and microhardness among multi-layered high-translucent and super-high-translucent zirconia, as well as the influence of finishing and aging on surface characteristics, were evaluated. Three multi-layered Zirconia were evaluated: STML (4Y-TZP), IPS, e.max ZirCAD CEREC (4Y-TZP+5Y-TZP), Ceramill Zolidfx ML (5Y-TZP). Ninety plate-shaped specimens were prepared; half were polished, and half were glazed. Cervical, medium, and incisal areas were evaluated for each glazed and polished surface before and after aging. They concluded that there was no significant difference among layers regarding surface characteristics. Surface treatment (glazing or polishing) significantly impacts surface roughness and microhardness; the glazed samples had higher surface roughness and lower microhardness than the polished ones. After aging, microroughness increases, but it is significant only for super translucent material 4Y-TZP (fourth generation). The super translucent 4Y-TZP material was more affected by aging than the Super-high translucent 5-TZP (third generation) (Toma et al. 2022).

The effect of 3,4,5,6,7 and 8 mol% yttria addition to Zirconia on the microstructure and mechanical properties of YSZ ceramics was investigated using conventional sintering at 1550^{0}C for 2 hours. It was found that with the increase in yttria content, the average size of grains, pores, fracture voids and cleavage facets all decreased, and a more homogeneous microstructure was observed with an increase in yttria content. The minimum flexural strength was observed for 5YSZ ceramic due to the maximum fraction of the cubic Zirconia phase. The highest level of flexural strength was found in fine-grained microstructure in 7YSZ ceramics, which contain monoclinic and tetragonal Zirconia phases (Kulyk et al. 2022).

The optical properties, related microstructures, and phase fractions of multi-layered Zirconia with and without a yttria gradient were evaluated and compared. Multi-layered Zirconia 5mol% yttria stabilized (Katana STML) and 4Y/5Y stabilized (emax MT Multi) were analyzed using the Opalescence parameter (OP), average transmittance (AT%), translucency parameter (TP), and contrast ratio (CR)—the translucency of Emax. MT Multi was slightly higher than Katana STML, assessed using AT and CR, but was similar in TP. The opalescent gradient was higher in Katana STML than in Emax. MT Multi. The gradient of pigmentation caused the translucency gradient in Katana STML; however, it was in Emax. MT Multi was caused by the difference in phase fraction and the gradient of pigmentation (Cho and Seol 2022).

Most often, the flexural strength of individual layers is measured; the overall mechanical performance of multilayer Y-TZP is rarely investigated. It has been found that different mechanical multilayer Y-TZP materials exhibited different mechanical and morphological properties. ZirCAD Prime (I), CERCON ht ML (S) and Katana Zirconia YML (K) were investigated. It was found that ZirCAD Prime had the highest flexural strength, followed by CERCON ML and Katana YML, indicating the material's ability to resist bending forces and thus being more durable in the aggressive conditions of the oral cavity. The quantitative phase analysis of each material was performed using the Powder X-ray Diffraction (PXRD). It has been found that ZirCAD Prime contains grains of uniform size, with a homogeneous grain size distribution and the highest portion of the t-ZrO2 phase, which exert a positive effect on the flexural strength. It has been found that enamel layers have higher yttria content (5 Y-TZP) while the dentine layer consists of lower yttria (3Y-TZP) content, intermediate layers include individual mixtures of 3Y-TZP and 5Y-TZP (Labetić et al. 2024).

2.3.c SINTERING

Sintering influences the mechanical properties of ceramics by influencing crystal growth, density, monoclinic phase and pore shrinking. Combining a high sintering temperature with a short time increases the flexural strength of zirconia. Ceramics are sintered for 6-8 hours and heated to 1500^0C for densification and grain growth. Such extended sintering programs are conducted in conventional furnaces; this form of heat generation is slow compared to Induction furnaces, which are more efficient and can operate at high heating rates. Thus, induction furnaces optimize workflow in dental laboratories and enable Zirconia to be used in chairside applications (Ahmed et al. 2019).

The feasibility of speed sintering programs using a conventional furnace to sinter 3 mol% to 5.4 mol% zirconia compared to a conventional sintering program was investigated. They found that using a speed sintering protocol in a conventional furnace did not significantly compromise the mechanical properties and could be used safely (Lubauer et al. 2023).

The long-term structural durability of microwave-speed sintered Zirconia (MWZ) relative to conventionally sintered Zirconia (CZ). 3Y-TZP discs were speed-sintered at 1450 °C for 15 min using an industrial microwave oven, while conventional sintering was performed using a dental furnace at 1530 °C for 2 hours. The sintering time was 105 minutes for MWZ and 600 minutes for CZ. MWZ resulted in six times faster fabrication. It was concluded that MWZ exhibited a slightly lower density, but a smaller grain size, lower cubic Zirconia content, and poorer translucency. Both materials showed similar flexural strength (Luz et al. 2021).

The impact of speed and high-speed sintering on the densification, microstructure, phase composition, translucency and flexural strength of yttria-stabilized zirconia

(YSZ) was investigated. 3YSZ (Zpex), 4YSZ (Zpex 4), and 5YSZ (Zpex Smile) powders were cold-isostatically pressed and sintered according to conventional speed (control, 12 hours), (28 min for 3YSZ, 60 min for 4YSZ, and 5YSZ), and high-speed (18 min) sintering. It was found that conventional sintering resulted in the highest density for all YSZ compositions, followed by speed and then high-speed sintering. All sintering protocols yielded similar strength values; however, speed and high-speed sintering protocols resulted in significantly lower translucency. The speed-sintered specimens had the smallest grain size, whereas the high-speed-sintered 5YSZ possessed the largest grain size (Alshahrani et al. 2024).

The effect of speed sintering on optical properties and biaxial flexural strength was assessed for multilayered zirconia IPS e.max ZirCAD Prime (ZP) and IPS e.max ZirCAD Prime Esthetic (ZPE). Both materials contain translucent, gradient, and dentin layers. It was found that the speed sintering reduced the biaxial flexural strength in all layers of ZirCAD Prime and the gradient layer of ZirCAD Prime Esthetic zirconia. On speed sintering, the translucency parameter of the gradient and dentin layers of ZirCAD Prime and the gradient layer of ZirCAD Prime Esthetic zirconia were significantly lower, so the differences may not be clinically relevant (Mohamed et al. 2025).

2.3.d FIXED DENTAL PROSTHESIS

Positioning of the connector within the multilayer blank (below, central, above) and altering the sintering time to fast (2 hours, 26 minutes), normal (4 hours 25 minutes), and long (9 hours, 50 minutes) affect the mechanical performance and fracture force of zirconia 3-unit fixed partial dentures (FPDs). Anatomic contour 3-unit FDPs were milled from ZirCAD Prime 16mm in three positions: above, central, and bottom. It has been found that all FPDs survived the thermal and mechanical loading without debonding or failure. The fracture forces were higher with prolonged sintering times, and were highest in the central and bottom positions. Therefore, the connectors of FDPs should be placed in the 3Y-TZP area. The mean fracture load of 1760 N for posterior FDPs on natural teeth made of 3Y/5Y-PSZ strength gradient zirconia was observed (Rosentritt et al. 2022).

Colour-gradient zirconia (Katana HTML, STML, UTML) and strength-gradient zirconia (Katana YML) materials with similar yttria levels in the basal layers) demonstrate comparable mechanical properties and are appropriate indications for anterior FDPs. Fracture load values for HTML (3Y-TZP) and YML (3-4Y-PSZ) were not statistically significantly different. So, the color-gradient material and the strength-gradient material showed comparable results, but there were differences within color-gradient materials. Increasing the yttria content to improve translucency leads to a reduction in the fracture load for color gradient multilayer zirconia (Pöppel et al., 2022).

The marginal and internal fit of the three-unit fixed dental prosthesis (FDPs) fabricated using CAD/CAM with two designs- full contour (FC) and Framework with cutback (FW), two Zirconia types (multi-layer- L and single-layer-W), two cement spaces (CS)- 30 µm and 45 µm. The fit of FDPs was evaluated using the replica method. The space between an abutment and FDPs in the marginal (MO), chamfer (CH), axial (AX) and occlusal (OC) areas was measured under an optical microscope. It was found that FDPs with single-layer Zirconia showed a better fit than those obtained with Multi-layer Zirconia. The reason could be due to the amount of metal oxide particles in each layer; the amount and the timing of shrinkage of each layer during secondary firing were different, which may distort the layers (Suzuki et al. 2020).

The effect of dynamic loading on the fracture resistance of gradient Zirconia fixed partial denture frameworks was evaluated. Forty frameworks were fabricated and divided into 3Y-TZP and gradient 5Y-TZP and 3Y-TZP. Conventional glass ionomer cement was used to cement the frameworks to stainless-steel master dies. Half of the samples underwent dynamic loading, and fracture resistance values were recorded using a Universal Testing machine and examined under a Scanning electron microscope. It was found that the fracture load values before and after dynamic loading were 1919 N and 1418 N for the 3Y-TZP group, 908 N and 716 N for the 5Y-TZP group, and 3Y-TZP. The relationship between the effects of Zirconia material and dynamic loading on fracture resistance was statistically significant. Dynamic loading significantly reduced the fracture resistance of all the materials. However, both frameworks showed fracture load values higher than typical masticatory values (Attia and Shokry 2023).

The effect of connector designs on the flexural strength of gradient Zirconia fixed dental prostheses compared to translucent Zirconia fixed dental prostheses. Two main experimental groups were translucent BruxZir with Sharp and round connector designs and Gradient Zirconia with sharp and round connector designs. The study results showed that the BruxZir group with a sharp connector design had the lowest mean flexural strength value (578.77), followed by the BruxZir group with a round connector design (709.10). In contrast, the highest flexural strength value was found in the ZirCAD group with a round connector (964.78). They concluded that the flexural strength of gradient 3Y and 5Y Zirconia is higher than translucent 4Y Zirconia. The round connector design showed higher flexural strength than the sharp designs (Lotfy, Mokhtar, and Hamza 2024).

In vitro performance and fracture force of different high-translucent Zirconia and their positioning within a multilayer Zirconia blank affect the performance of 4-unit

FPDs fabricated from Monolayers of 3Y-TZP, 4Y-TZP, 5Y-TZP, and 4Y/5Y-TZP, as well as different positions within the multilayer Zirconia blanks. Milling position within the multilayer blank seems to have only minor effects on the mechanical properties of the restorations. They found that 5Y-TZP is unsuitable for fabricating wide-span Fixed dental prostheses (Strasser et al. 2023).

The fracture behaviour of cantilever-fixed dental prostheses made from 3Y-TZP, 4Y-TZP, and 5Y-TZP was compared. After sintering, samples were stored in demineralized water and subjected to chewing simulation with thermal cycling. It was found that all materials can be used to fabricate permanent posterior cantilever FDPs, as no significant difference in fracture behaviour was found as a result of different yttria concentrations. However, more data is necessary to apply this concept (Schmidt 2022).

The fracture resistance and fracture modes of tooth-supported cantilever fixed dental prosthesis made from gradient Zirconia (3Y,3Y/5Y,4Y/5Y) and different thicknesses (0.7mm, 1.0mm) were fabricated, underwent thermal cycling and chewing simulation, and were then loaded statically until fracture. They concluded that cantilever FDPs fabricated from gradient Zirconia are not recommended for clinical use, whereas those made from 3Y-TZP exhibit fracture resistance above possible masticatory loads in the posterior regions (Klotz et al. 2024).

2.4 FINITE ELEMENT ANALYSIS

To estimate stresses using the linear elastic model, a force is applied to a node, element, or area of the occlusal surface to simulate occlusal loading. However, for non-linear models, the sliding and friction phenomenon during mastication is reproduced using contact elements. The forces acting on molars during mastication change in direction, location and magnitude depending upon the contact between opposing teeth. However, there are several limitations, as the surfaces move in and out of contact with each other predictably, and contact simulations require significant computer resources (Murakami and Wakabayashi 2014).

A non-linear contact simulation of the interface between the mandibular molar occlusal surface and morsels was conducted, and the stresses induced on the molar during clenching and chewing morsels of different elastic moduli were calculated. It was found that maximal stresses occurred in occlusal enamel and the cervical region of the lingual wall of the mandibular molar (Dejak, Młotkowski, and Romanowicz 2003).

Modifying the core design improved the reliability of all-ceramic crown core-veneer systems. This was tested using finite element analysis (FEA) to evaluate stress distribution on Y-TZP-supported and Metal Ceramic Restorations under simulated loads. The computer simulation results showed a shift from the veneer to the core layer when the modified core design was compared to the conventional design. Hence, the core design modification improved the reliability of all-ceramic crowns, which is comparable to metal-ceramic restorations(Silva et al. 2012).

The load-bearing capability of a prosthetic restoration depends on the material's fracture resistance and the preparation of a suitable design with adequate material thickness (Sun et al. 2014).

The tooth preparation-crown configuration was evaluated using Finite element analysis under a vertical load of 200 N and a combined vertical load of 200 N and horizontal buccal load of 100 N at different positions from the central fossa to the cusp tip over a 1 mm diameter surface area. They concluded that the combined loading of horizontal and vertical loads resulted in an increase in Maximum Principal stress values compared to vertical loading alone (Coelho et al.).

The effect of substrate core material and occlusal contact patterns on the fatigue behaviour and stress distribution of single-unit ceramic crowns made up of IPS e.max ZirCAD Zirconia, IPS e.max CAD lithium disilicate and Enamic PICN was investigated using Finite element analysis. It was concluded that the core/ substrate material did not influence the mechanical properties of Zirconia and Lithium disilicate crowns (Packaeser et al. 2025).

The 3D FEA model is a crucial tool for predicting stress distribution and supporting the structural design of dental restorations. A crown and a 3-unit fixed partial denture (FPD) made of ceramic framework (Y-TZP) and veneered with porcelain were scanned with a Micro CT scanner, and FEA models were fabricated. For the crown model, tensile stresses were concentrated in the internal surface of the core and near the applied load. For the FPD model, the highest tensile stresses were found in the framework on the cervical area of the connector and pontic (Darwich et al. 2023).

The influence of different factors on the Maximum Principal stress in the tooth-crown complex was investigated. Loads were applied over a 1mm diameter area. The major contributions are crown material, crown thickness, load position, cusp incline and supporting core. The Maximum Principal stress magnitude ranged from 73.7 to 214.0 MPa. They concluded that 1.00mm thick crowns exhibited less Maximum principal stress than 0.8mm crowns(Rekow et al. 2006).

Though the occlusal surface design may be dictated by the antagonist teeth anatomy, low cusp angulation should be observed to decrease the stress concentration in the crown. Maximum principal stress increased significantly with the addition of horizontal loading (Rafferty et al. 2010).

The stress distribution in monolithic High translucency 5Y zirconia, as investigated by FEA, concluded that the maximum loads without risk of failure are 500 N for vertical loading, 350 N for angular loading, and 200 N for horizontal loading. In all molar regions, the highest Maximum principal stress was observed under horizontal loading conditions(Ward et al. 2025).

The fatigue behaviour and stress distribution in mandibular molars with different onlays made from Lithium disilicate ceramic, polymer-infiltrated ceramic network (PICN), and Zirconia ceramics, and various preparation designs under static and cyclic conditions, using Finite Element Analysis. The study concluded that cyclic loading led to higher stresses, and Zirconia ceramic emerged as the most suitable material for onlays to withstand stresses in the oral environment. The study also found that Zirconia and lithium disilicate ceramic onlays exhibited similar stress values, while PICN was deemed unsuitable for onlays (Darwich et al. 2023).

The mechanical behaviour of the posterior crowns made from Lava Ultimate (LU) and IPS e.max. CAD using FEA analysis with different occlusal thicknesses of 1.0mm and 2.00 mm was conducted. The study found that LU crowns effectively dissipated occlusal load to the tooth structure, while LD material concentrated the load inside the crown. The study also highlighted that the higher the Young's modulus mismatch between the crown material and substrate, the higher the load stress concentration inside the material. The study concluded that crown material and occlusal thickness play a significant role in the mechanical behaviour of stress dissipation to the tooth structure, and the most favourable stress distribution was observed with 2mm thick occlusal crowns with an axial load application, primarily for LU material (Martins et al. 2023).

The higher the elastic modulus of the crown material, the lower the equivalent stresses in the tooth structure and luting cement; hence, crowns made up of stiff materials like Zirconia-based ceramic with thinner walls (0.5-0.7mm chamfer margin, axial walls with 6 degrees inclination, 1.0-1.5mm occlusal reduction) are less prone to debonding and strengthen the dental structure of teeth (Dejak et al. 2012).

The strength of a prosthetic bridge with variable geometry in the connectors in an anterior bridge with four different cross-sectional areas of the connectors: 4.0, 5.0, 5.5, and 6.0 mm^2 and three different alloys- high noble, noble and base alloys under vertical and oblique loading was investigated. It has been concluded that decreasing the connector's cross-sectional area by approximately 1 mm^2 increases the maximum stress by an average of 15%, while increasing the connector's cross-sectional area by approximately 1 mm^2 reduces the deflection of the bridge by an average of 16%. Based on the analysis of the bridge displacements, the connector's cross-sectional area affects the displacement of the prosthesis. Maximum stresses were 19% higher under oblique forces, so it must be emphasized that the incorrect assumptions of the method of loading the prosthetic bridge only by a vertical force led to a significant underestimation of the stress, which translates into an incorrect choice of the connector's cross-sectional area or material (Reimann, Żmudzki, and Dobrzański 2015).

The influence of different post-core systems and crown materials used to restore premolars was investigated using finite element analysis. Glass fibre-reinforced composite post-core and carbon fibre-reinforced composite post-core demonstrated favourable stress distribution in the dentin and crown materials. The more rigid crown material transmitted fewer stresses to the underlying core and crown (Maravic 2025).

Finite element analysis is gaining popularity due to significant advancements. Now, actual tomographic images, scans, or MRIs of jaws, soft tissues, and prosthetic and restorative elements provide more realistic volumetric data. It has been found that the Young's modulus and Poisson's ratio published in the literature do not have a reference, and they were not consistent for the material (Abu-Naba'a 2023).

So, the in-lab measurement of Young's modulus and Poisson's ratio will be more beneficial for calculating the actual mechanical behaviour of the material under Finite element analysis testing. Hardness and Young's modulus can be measured using Nanoindentation testing (Song 2019).

2.5 NANOINDENTATION

It is possible to deform ceramics plastically and measure their mechanical properties to a micrometre scale using nanoindentation techniques. Dislocation activities predominantly mediate nanoindentation-induced plasticity.(Masuda, Morita, and Ohmura 2020) Pop-in events were detected on the onset of plasticity with strain during nanoindentation (Fujikane et al. 2007).

The specific mechanical properties of YSZ were determined through nanoindentation testing and FEA analysis, and compared with ultrasonic measurements available in the literature. It was observed that there is a large anisotropy of Young's modulus and hardness measured in the elastic and plastic regions, which is affected by the indentation load velocity and time (Fujikane et al. 2007).

3. STATEMENT OF THE PROBLEM

As the yttria content increases in multilayered zirconia, the tetragonal phase decreases, which inhibits or slows crack propagation. Due to heightened aesthetic demands, it would be beneficial to provide more aesthetically pleasing zirconia crowns in the posterior region of the mouth as well. Therefore, investigating the mechanical behaviour of different zirconia crowns under various occlusal loading conditions is a pertinent topic in the field of prosthodontics.

4. PURPOSE OF THE STUDY

To evaluate and compare the mechanical behaviour of crowns made from multilayered zirconia by Finite Element Analysis (FEA) using the elastic modulus obtained from nanoindentation testing, and compare it with the reference Monolithic Zirconia.

5. OBJECTIVES OF THE STUDY

- To evaluate and compare the elastic modulus and hardness of two brands of multilayer zirconia using Nanoindentation testing.
- To evaluate and compare the maximum principal stresses in crowns made of multilayer zirconia using FEA.
- To evaluate and compare the maximum principal stresses in multilayer Zirconia crowns with two occlusal thicknesses: 1 mm and 1.5 mm.
- To evaluate and compare the maximum principal stresses in crowns under different loading conditions-Axial and Horizontal.

- To evaluate and compare the maximum principal stresses in crowns made of multilayer zirconia and monolithic Zirconia using FEA.

6. NULL HYPOTHESES

H01: The elastic modulus of multilayered zirconia will not significantly differ from each other.

H02: The maximum principal stresses of multilayered zirconia crowns will not significantly differ from each other.

H03: The maximum principal stresses of multilayered zirconia crowns will not significantly differ from those of monolithic zirconia crowns.

H04: The occlusal thicknesses of 1 mm and 1.5 mm will not affect the maximum principal stresses in the crowns.

H05: The maximum principal stress in the crowns will not significantly differ on the application of axial vs horizontal loading.

CHAPTER 2

MATERIALS AND METHODS

The present study design and methodology have been adapted from the previous study described by Abdulmajeed et al. 2020b; Bataineh and Al Janaideh 2023; Coelho et al. 2009; Martins et al. 2023).

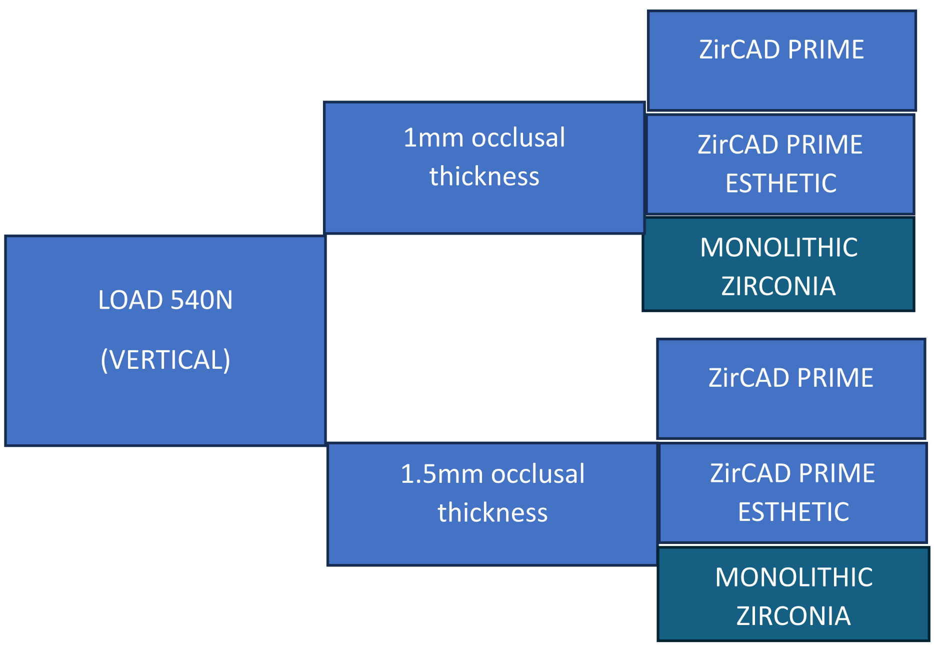


Figure 1. Methodology used in this study, showing Groups of samples subjected to 540 N Vertical loading.

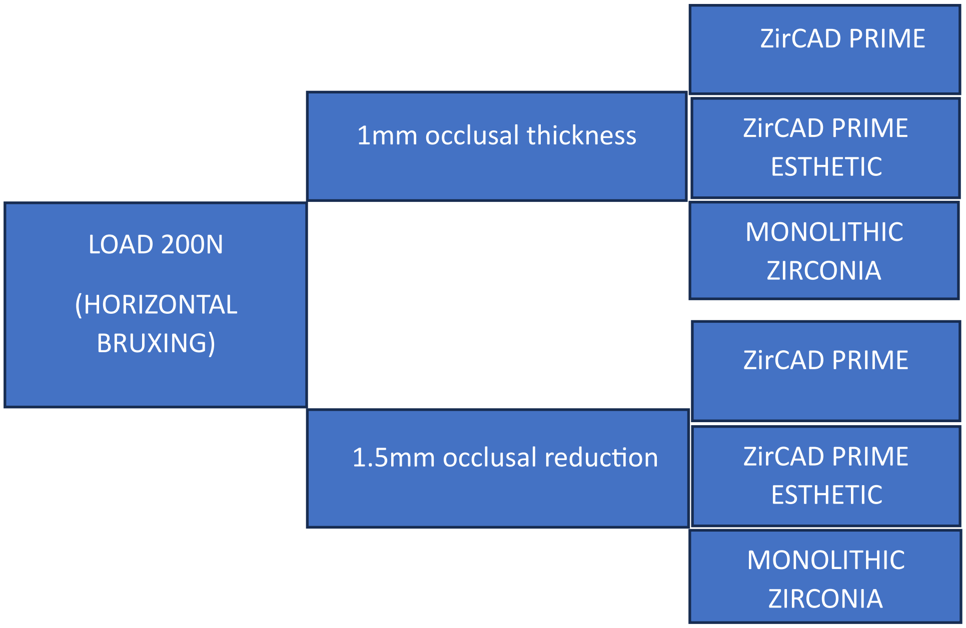


Figure 2. Methodology used in the study showing Groups of samples subjected to 200 N Horizontal loading

SAMPLE PREPARATION

The pre-sintered zirconia blocks were sectioned into 10 x 16 x 4 mm slabs using a precision diamond saw operating at low speed with gravity feed. Following the manufacturer's protocol, the slabs underwent sintering and were allowed to cool before being removed from the furnace. To stabilize the samples for polishing, they were encased in epoxy resin. Surface preparation was carried out to ensure the zirconia samples were free from any irregularities or contaminants that might interfere with subsequent tests. This involved a series of progressive grinding and polishing steps. Initially, silicon carbide abrasive papers with grit sizes ranging from 800 to 2000 were employed to address any imperfections arising from the milling and sintering processes. Then, diamond polishing suspensions with particle sizes of 1 µm and 0.5 µm were used to achieve a polished, mirror-like finish. The polishing steps were conducted under constant water irrigation to prevent overheating and potential damage to the samples. Finally, each sample was examined under a microscope to confirm a smooth surface without significant scratches.

NANOINDENTATION

The hardness and elastic modulus of the prepared zirconia samples were evaluated using a Hysitron TI750 Ubi Nanoindenter. A load of 400 µN was applied with the Nanoindenter tip to the Multilayer Zirconia samples, comprising layers of Enamel, Intermediate, and Dentine. Ten readings per layer were measured, and the average value was calculated. Hardness values were calculated based on the maximum load applied and the contact area. At the same time, the elastic modulus was determined from the stiffness of the unloading curve, considering the mechanical properties of both the sample and the indenter. The nanoindentation result values were input for testing the Finite Element Analysis of the Zirconia crowns under different loading conditions.

Figure 3. Hysitron nanoindenter used in the study.

Figure 4. Nanoindentation testing of the Zirconia sample.

3D CAD MODELING

A typodont plastic tooth, a mandibular left first molar, was used as a primary model. The tooth was prepared for a full-coverage Zirconia crown with occlusal reductions of 1mm and 1.5mm, axial reductions of 1mm, and a shoulder of uniform width of 1mm (Abdulmajeed et al. 2020b; Bataineh and Al Janaideh 2023). The tooth preparation was scanned using a Primescan intraoral scanner to create an STL file. A full-coverage Crown is designed using in-lab CAD software.

Figure 5. Prepared Typodont molar for a full coverage Zirconia crown.

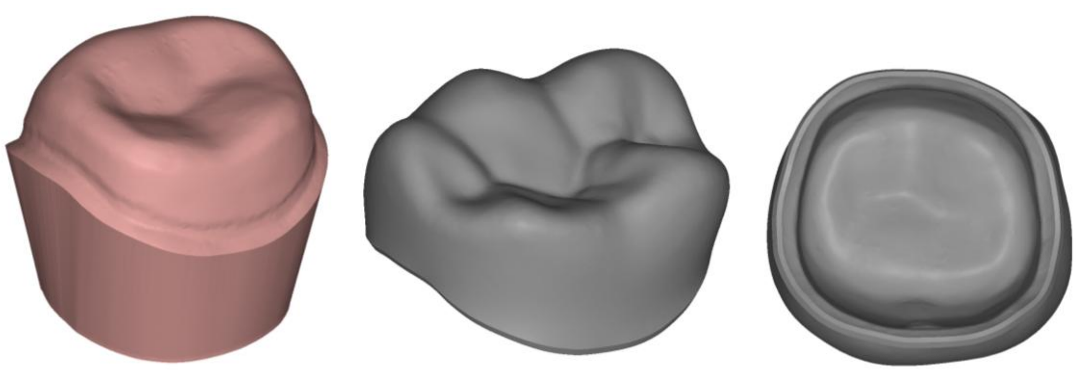


Figure 6. Digital STL files of the prepared tooth structure and crown (polished and intaglio surface).

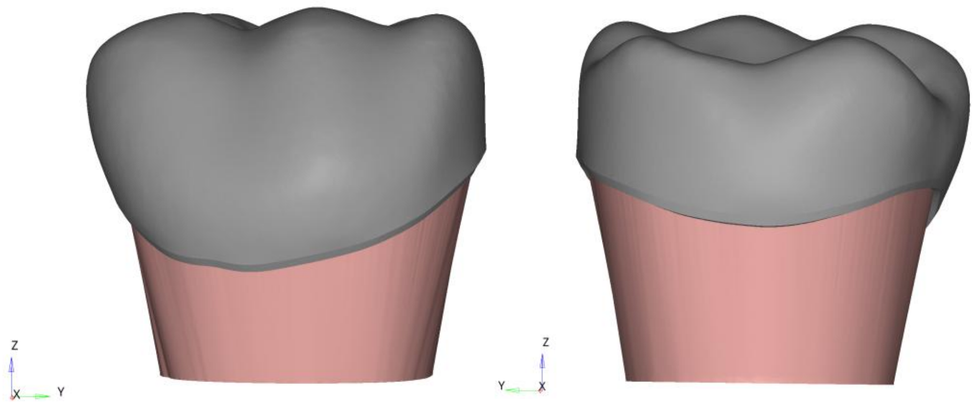


Figure 7. 3D model of the tooth and complete coverage crown assembly (buccal and lingual view).

3D models of tooth preparation and full coverage crown were created and merged using Hyper View Player 2022.3 software.
The 3D model was created using 848135 nodes and 4567316 elements. Once 3D meshes were obtained, a boundary condition (zero displacements) was set to the bottom of the root.(Ausiello et al. 2004; Shahrbaf et al. 2013). Highly refined and accurate volume meshes were constructed to replicate the natural crown-tooth complex.

FEA TESTING

The tooth-crown complex underwent a vertical loading condition of 540 N (Calderon et al. 2006) and a horizontal load of 200 N in the central fossa and buccal cusps (Wicaksono et al. 2022). The Maximum principal Stresses were then calculated in tooth structure and the Zirconia crowns.

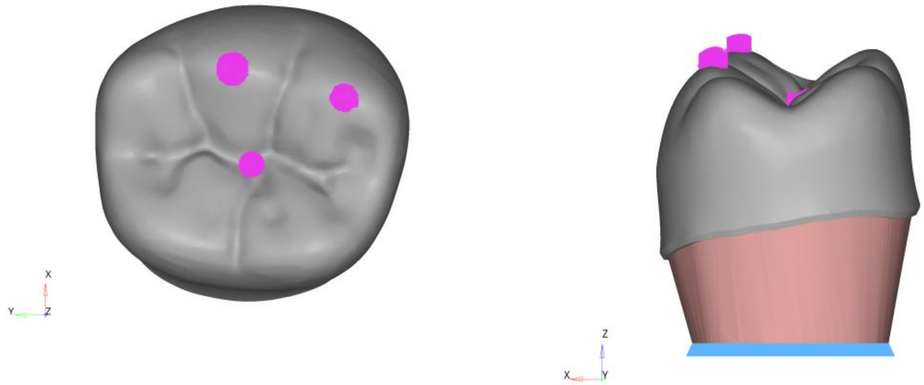


Figure 8. Occlusal and lateral view of 540N vertical load application in the central fossa and buccal cusps in the -Z direction.

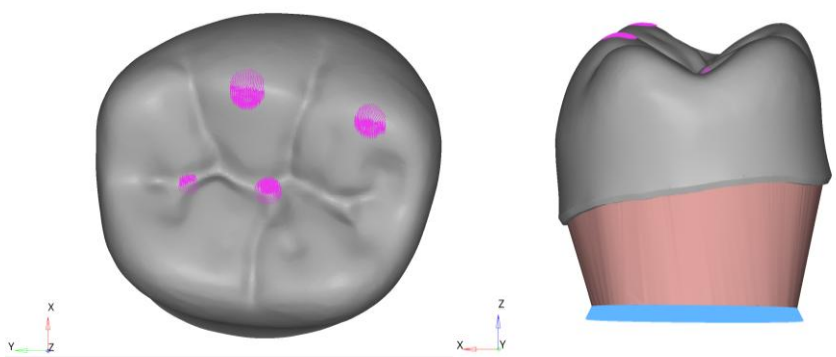


Figure 9. Occlusal and lateral views of 200 N horizontal load application in the central fossa and the buccal cusps in +X direction (Lingual to Buccal).

In the present study, the material properties were assumed to be homogeneous, isotropic, and linearly elastic. Adhesion was simulated as perfect without voids. The features from the periodontal ligament, pulp tissue, and cement thickness layer were not considered. A load of 540 N, representing the maximum biting load (Calderon et al. 2006), was applied axially at the central fossa and the buccal cusps. Additionally, a horizontal bruxing load of 200 N was applied to the central fossa and buccal cusps, simulating the bruxing forces generated by the patient (Wicaksono et al. 2022). The material properties for Multilayer Zirconia (ZirCAD Prime and ZirCAD Prime Esthetic) were calculated based on nanoindentation testing results. However, for Monolithic Zirconia, the material properties were extracted from the literature (Díaz et al., 2020; Fogaing et al., 2006; Kohal et al., 2002; Ozen et al., 2007; Pantić et al., 2019; Yamanel et al., 2009).

STATISTICAL ANALYSIS

The assumptions of normality and homogeneity of variance were tested using the Shapiro-Wilk and Levene's tests, respectively. A two-way ANOVA was performed to evaluate whether the region (incisal, transition, and dentin) affects the hardness and elastic modulus of the two zirconia brands. The ANOVA will also examine the main effects of region and brand, as well as their interaction effect. We considered an alpha level of < 0.05 to be significant for comparisons. Statistical analysis was performed using the Origin statistical software.

CHAPTER 3

RESULTS

NANOINDENTATION

Table 1. Mean values and Standard Deviation for the Elastic Modulus for each region of the 2 Zirconia brands.

Brand	Dental layer	Mean	N
	Incisal	229.6 ± 8.1	20
Esthetics	Transition	232.6 ± 14.6	20
	Dentin	237.7 ± 9.7	20
	Incisal	219.7 ± 13.4	20
ZirCAD Prime	Transition	224.9 ± 12.5	20
	Dentin	229.5 ± 8.1	20

Table 2. Mean values and Standard Deviation for the Hardness for each region of the 2 Zirconia brands.

Brand	Dental layer	Mean	N
	Incisal	17.8 ± 3.3	20
Esthetics	Transition	17.9 ± 2.2	20
	Dentin	18.1 ± 1.4	20
	Incisal	18.8 ± 1.8	20
ZirCAD Prime	Intermediate	18.5 ± 1.6	20
	Dentin	17.8 ± 0.95	20

Figure 10. Bar graph showing the mean Elastic Modulus and Hardness with Standard Deviation of each region of 2 Zirconia Brands.

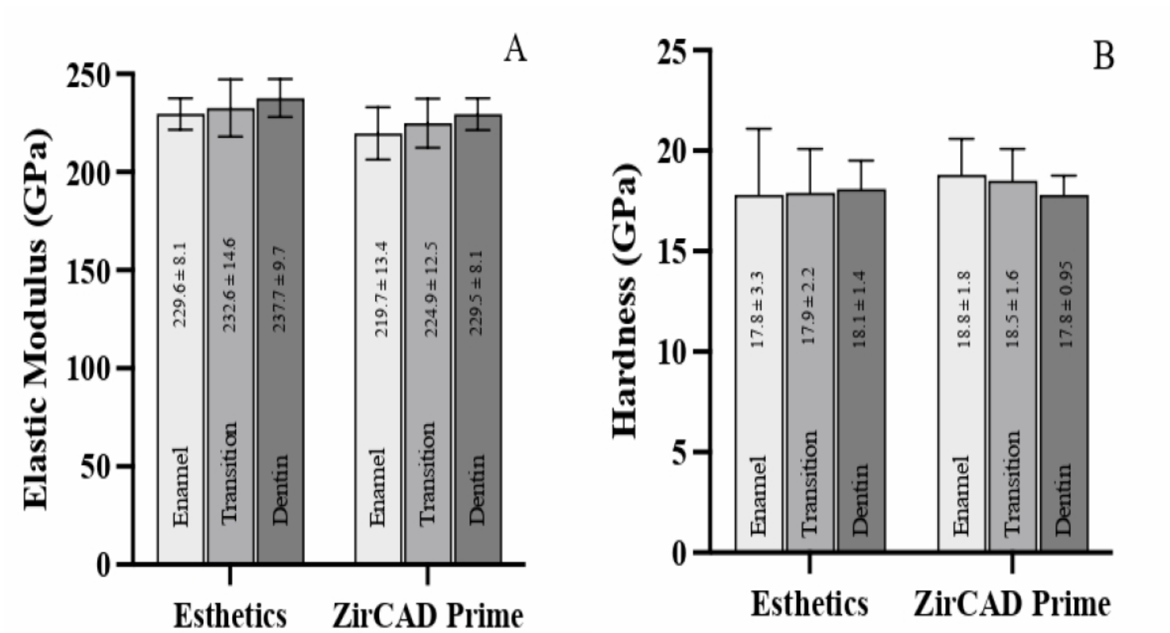
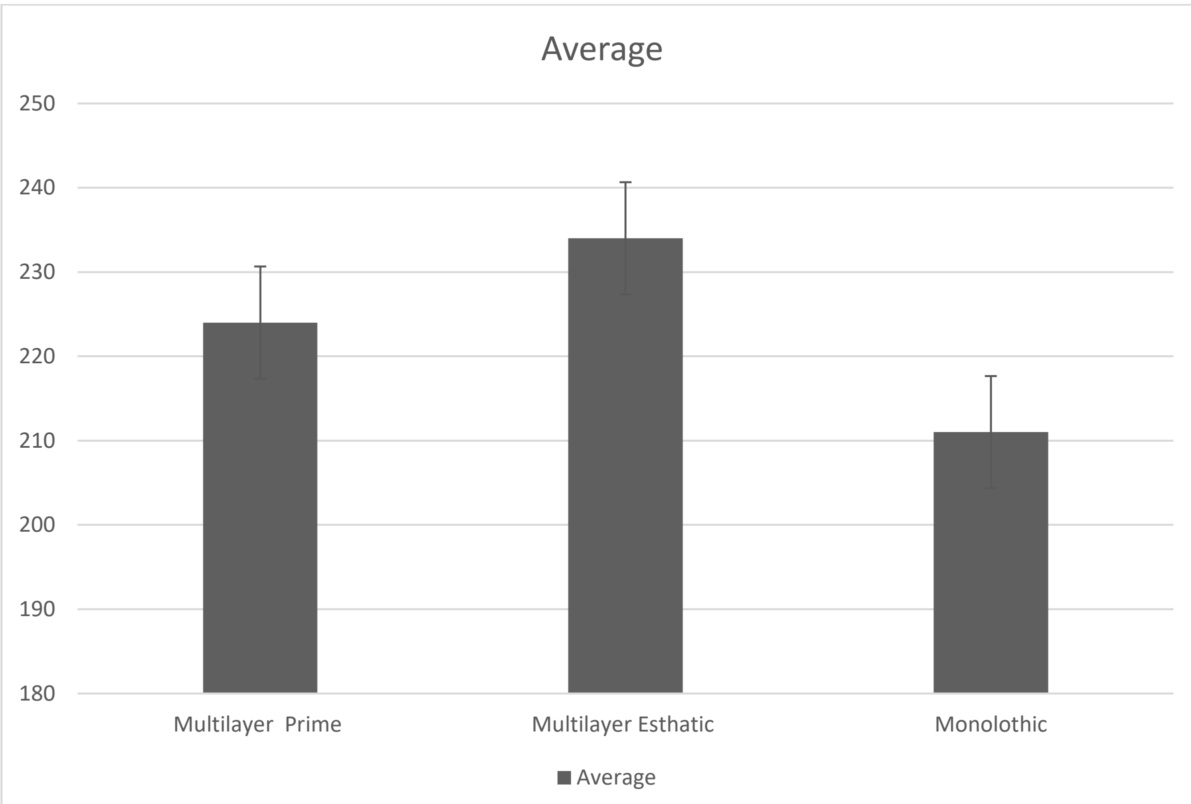


Figure 11. Bar graph showing the mean Elastic Modulus with Standard Deviation of both Multilayer Zirconia Brands in comparison to the Monolithic Zirconia (Díaz et al. 2020; Fogaing et al. 2006; Kohal et al. 2002; Ozen et al. 2007; Pantić et al. 2019; Yamanel et al. 2009**).**



Tables 1 and 3 show the mean values and standard deviations of the elastic modulus (GPa) for each region (incisal, transition, and dentin) of the two zirconia brands. The mean values for the elastic modulus of the incisal, transition, and dentin regions in the ZirCAD Prime are 219.7 ± 13.4, 224.9 ± 12.5, and 229.5 ± 8.1 MPa, respectively. The mean values for the elastic modulus for incisal, transition, and dentin regions in ZirCAD Prime Esthetics are 229.6 ± 14.9, 232.6 ± 14.6, and 237.7 ± 9.7 MPa, respectively. The average elastic modulus for the ZirCAD Prime brand was 224.7 ±

12.1 GPa, while for the ZirCAD Prime Esthetics brand, it was 233.3 ± 13.5 GPa. These overall averages suggest that the ZirCAD Prime Esthetics brand exhibits a higher average stiffness than the ZirCAD Prime brand. At the 0.05 significance level, the difference in the population means of the two brands was statistically significant. Additionally, there was a significant increase in the elastic modulus from the incisal to dentin region within each brand at the 0.05 level. However, the interaction between brand and region was not statistically significant at the 0.05 level, indicating that the increase in elastic modulus from incisal to dentin was similar for both brands, with no significant variation.

Tables 2 and 3 present the mean values and standard deviations for the hardness (in MPa) in each region for the two zirconia brands. The mean hardness values for ZirCAD Prime in the incisal, transition, and dentin layers were 18.8 ± 1.8, 18.5 ± 1.6, and 17.8 ± 0.95 MPa, respectively. The mean hardness values for ZirCAD Prime Esthetics in the incisal, transition, and dentin layer were 17.8 ± 3.3, 17.9 ± 2.2, and 18.1 ± 1.4 MPa, respectively. No significant difference was observed among the hardness values between the brands, and no significant difference was observed between each of the three regions ($p > 0.05$). Additionally, there was no interaction between brand and region regarding the hardness at the 0.05 significance level.

Table 4 show the mean values and standard deviations for the Elastic Modulus (GPa) of the two Multilayer Zirconia and Monolithic Zirconia (reference from literature-Díaz et al. 2020; Fogaing et al. 2006; Kohal et al. 2002; Ozen et al. 2007; Pantić et al. 2019; Yamanel et al. 2009). No significant difference was observed in the elastic modulus values between the different types of Zirconia.

FEA TESTING

On the contour plot of maximum principal stress, the positive values on the scale (Red) signify tensile stresses, and negative values (blue) dictate compressive stresses**.**

MODEL 1- In the 3D model with a **1mm** occlusal reduction, the vertical load of **540N** was applied to the buccal cusps and the central fossa of the Multilayer crowns and compared with the reference Monolithic Zirconia crown.

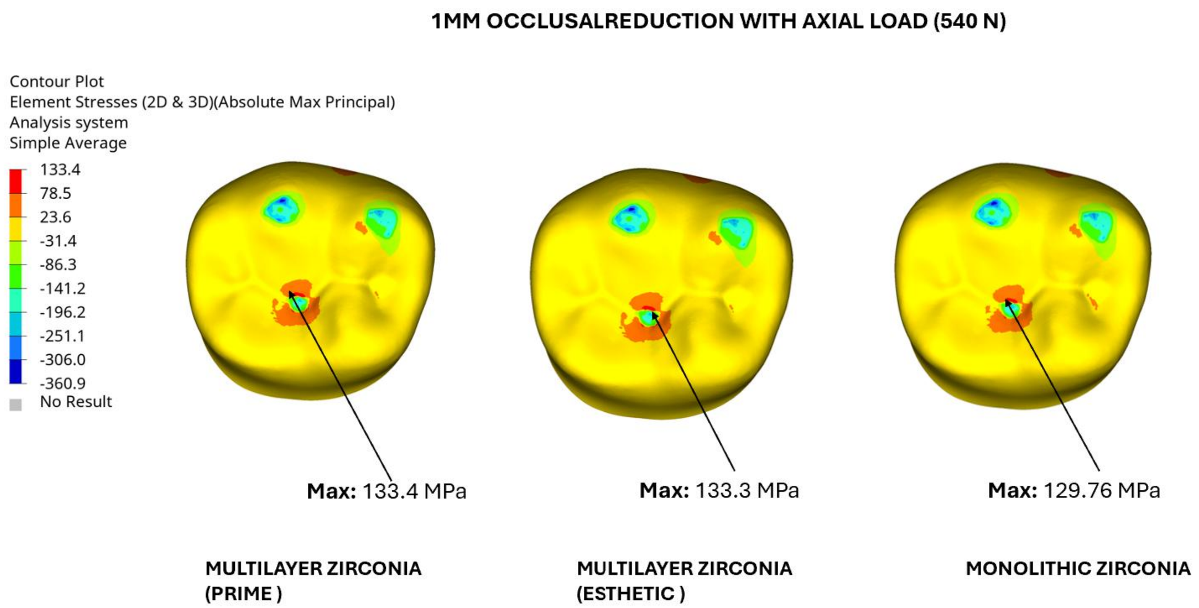


Figure 12. Maximum Principal Stresses on the Cameo Surface of 1 mm occlusal reduction Multilayer and Monolithic Crowns under a Vertical Load of 540 N.

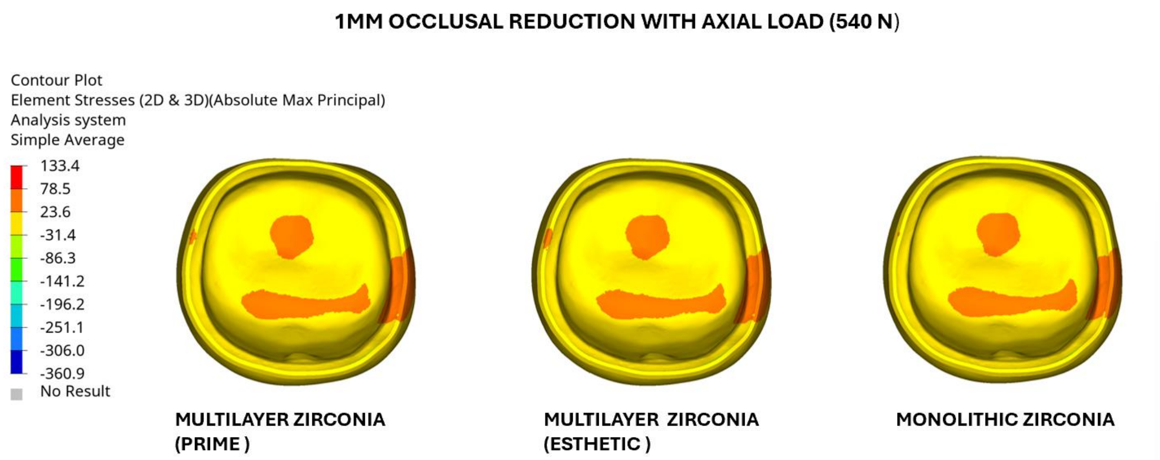


Figure 13. Maximum Principal Stresses on the Intaglio Surface of 1 mm Multilayered and Monolithic crowns under a Vertical Load of 540 N.

Crown - The Maximum principal tensile stress of 129.76 MPa, 133.4 MPa, and 133.3 MPa was observed in the central fossa and the buccal cusps on the cameo/polished surface of Monolithic, Multilayer ZirCAD Prime, and Multilayer ZirCAD Prime Esthetic. The Zones of compressive stress, ranging from 86.3 MPa to 360.9 MPa, were observed on the buccal cusps of the Monolithic, ZirCAD Prime, and ZirCAD Prime Esthetic specimens, respectively. The Intaglio surface of the crown experienced tensile stresses ranging from 23.6 to 78.5 MPa at the mesial margin and under the loading areas of the buccal cusps and central fossa.

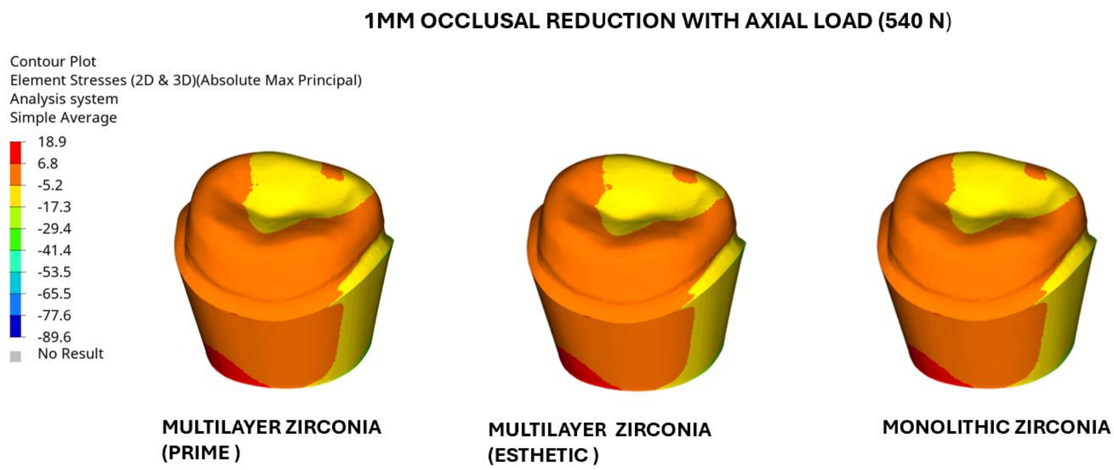


Figure 14. Maximum Principal Stresses on a 1 mm occlusal tooth preparation under Multilayered and Monolithic crowns subjected to a vertical load of 540 N.

Tooth Preparation—The tooth structure experienced a zone of compression stresses ranging from 5.2 to 17.3 MPa under the loading areas and the buccal margin for all three crowns. A transition zone was observed where stresses fluctuated from compressive (5 MPa) to tensile (6.8 MPa). Additional higher stresses were found at the boundary conditions; consequently, they were not taken into account.

MODEL 2- In the 3D model with a **1mm** occlusal reduction, the Horizontal Bruxing load of **200 N** was applied to the buccal cusps and the central fossa of the Monolithic and Multilayer Zirconia crown.

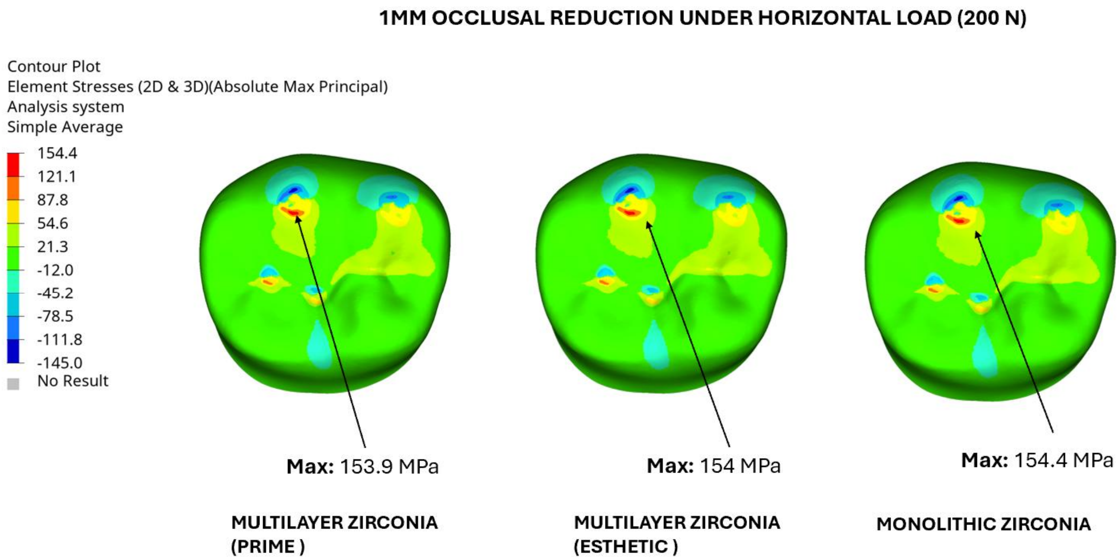


Figure 15. Maximum Principal Stresses on the Cameo Surface of 1 mm occlusal reduction Monolithic and Multilayer Crowns under a Horizontal Load of 200 N.

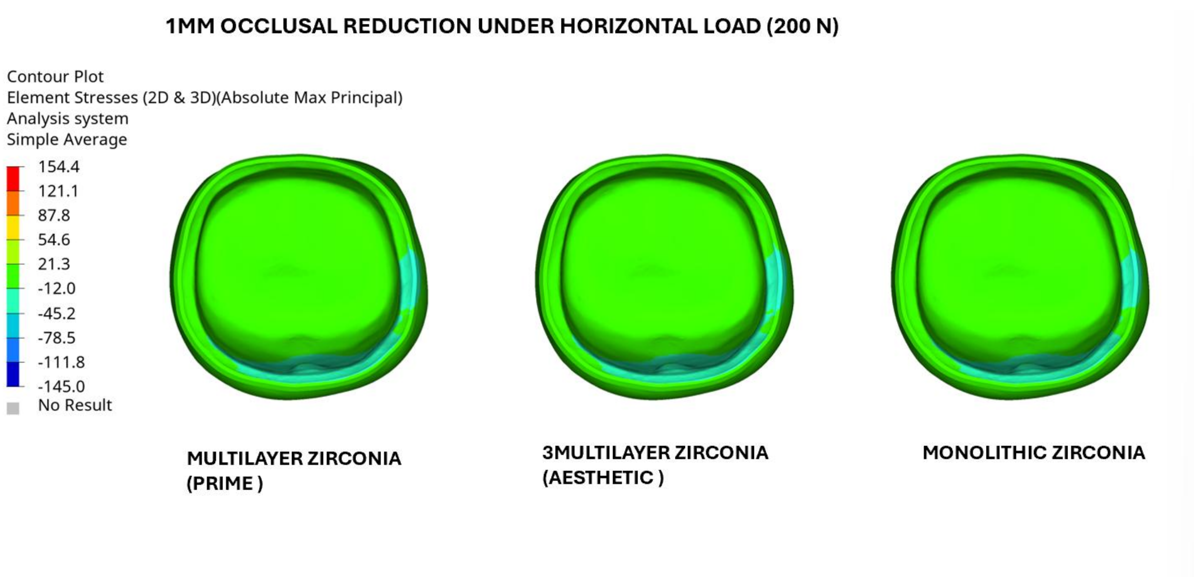


Figure 16. Maximum Principal Stresses on the Intaglio Surface of 1 mm occlusal reduction Multilayer and Monolithic Crowns under a Horizontal Load of 200 N.

Crown - The tensile stresses were observed on the lingual facing slopes of the buccal cusps and central fossa of Monolithic, Multilayer ZirCAD Prime, and ZirCAD Prime Esthetic, with the maximum principal tensile stress (Red) of 154.4 MPa, 153.9 MPa, and 154 MPa, respectively, on the cameo surface. The Intaglio surface of the crown exhibits an area of compression in the range of 12.0 to 45.2 MPa at the mesial and buccal margins.

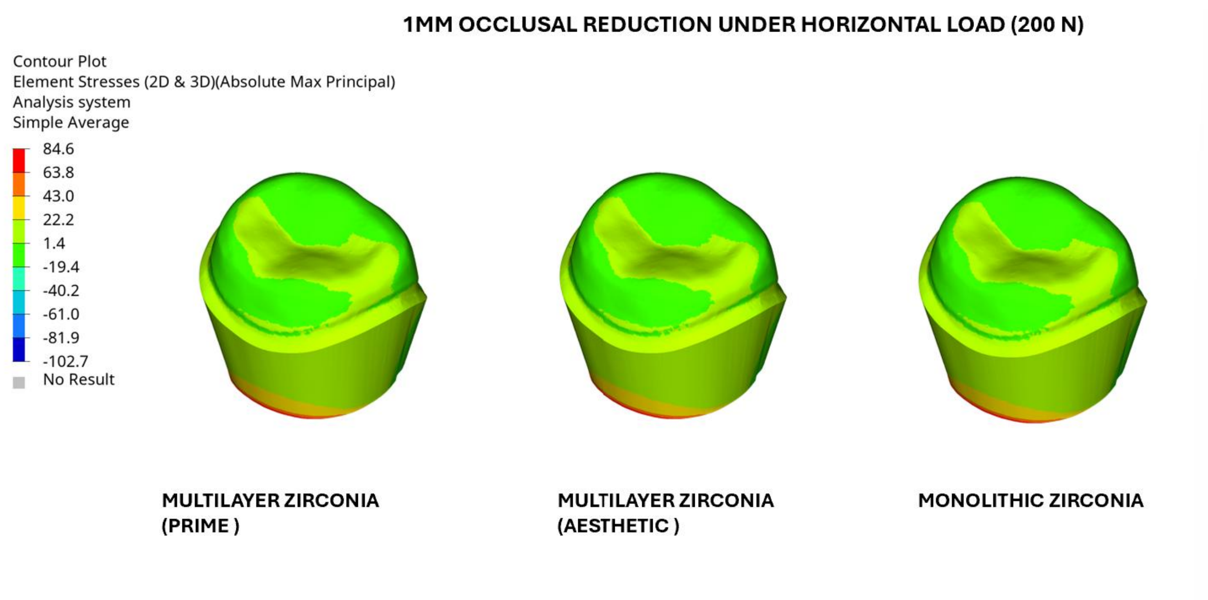


Figure 17. Maximum Principal Stresses on a 1 mm occlusal tooth reduction under Multilayer and Monolithic crowns subjected to a horizontal load of 200 N.

Tooth Preparation - Under the horizontal loading, a zone of tensile stresses was observed in the central portion of the occlusal preparation and the radicular portion of the root, with a range of 1.4 to 22.2 MPa. The margins of the tooth preparation were also subjected to tensile stresses. Results were the same for all three crowns. The tooth structure experienced fluctuating stresses, ranging from 1.4 MPa in tension to 19.4 MPa in compression, at the lingual-facing slopes of buccal cusps and the lingual aspect of lingual cusps. Additional higher stresses were found at the boundary conditions; consequently, they were not taken into account.

MODEL 3- In the 3D model with a 1.5mm occlusal reduction, the vertical load of 540 N was applied to the buccal cusps and the central fossa of the Monolithic and Multilayer Zirconia crowns.

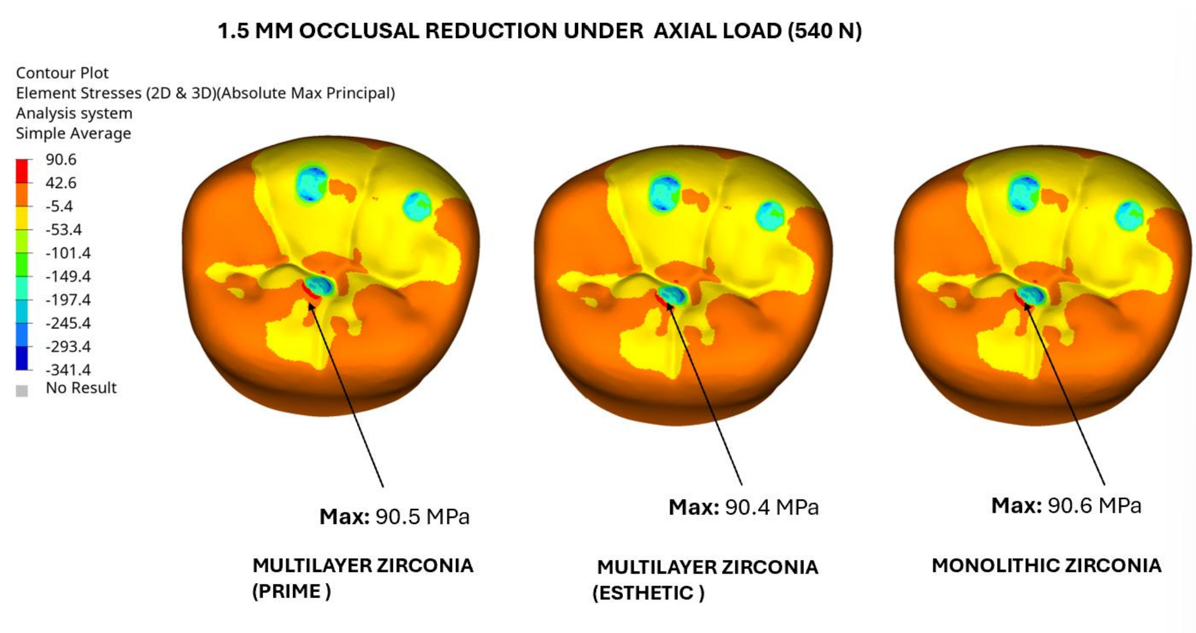


Figure 18. Maximum Principal Stresses on the Cameo Surface of 1.5 mm occlusal reduction Multilayer and Monolithic Crowns under Axial Load of 540 N.

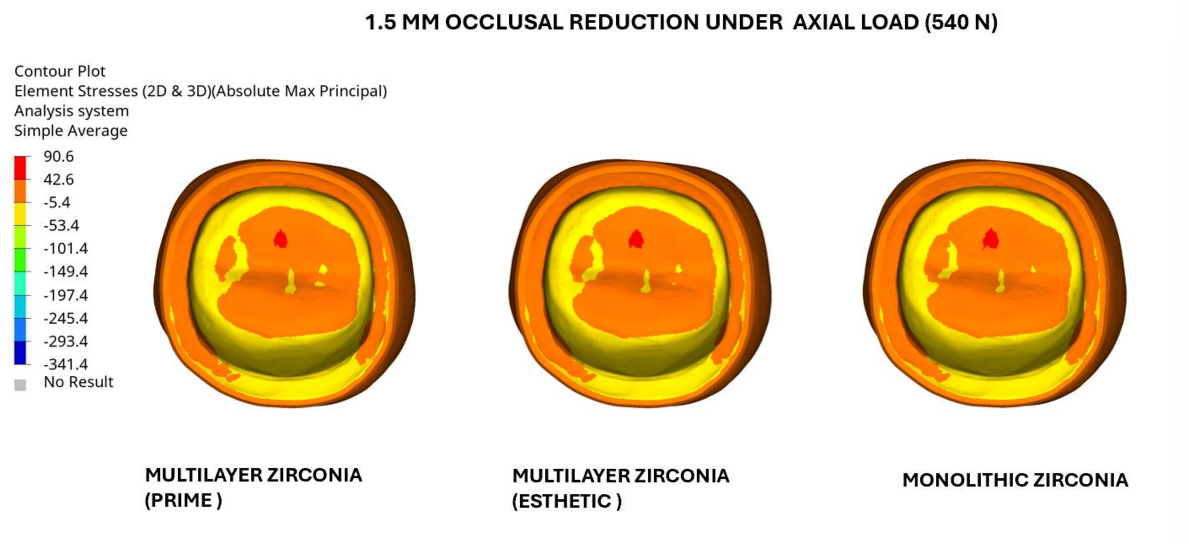


Figure 19. Maximum Principal Stresses on the Intaglio surface of 1.5 mm occlusal reduction Multilayer and Monolithic Crowns under axial load of 540N.

Crown - The maximum principal tensile stress (Red) of 90.5 MPa, 90.4 MPa, and 90.6 MPa was observed in the central fossa for the Monolithic, Multilayered ZirCAD Prime, and ZirCAD Prime Esthetic crowns, respectively. Compression zones in the range of 149.4 to 341.4 MPa were observed on the buccal cusps. The intaglio surface of the crowns exhibits a maximum principal stress of approximately 90.6 MPa, with areas of tension present on the internal aspect of the occlusal portion where loading was applied, as well as on the cervical margin of the crown.

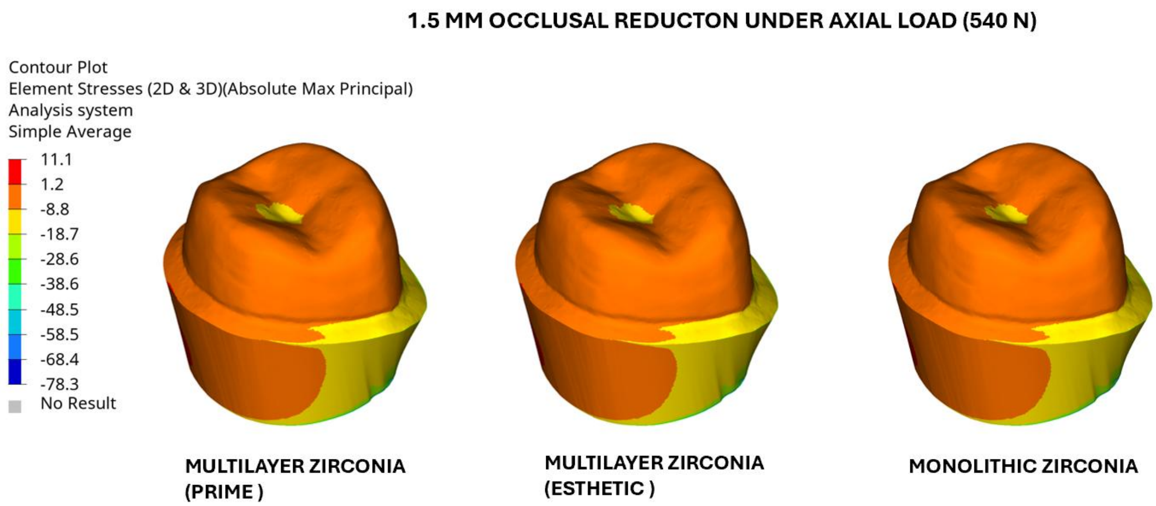


Figure 20. Maximum Principal Stresses on the 1.5mm occlusal tooth preparation underneath Multilayer and Monolithic Crown under axial load of 540N.

Tooth Preparation—The tooth structure experienced a fluctuating zone of tensile stresses of 1.2 MPa and compressive stresses of 8.8 MPa in most parts, except on the central fossa and buccal margin, and the radicular dentine on the buccal aspect, which underwent compression in the range of 8.8 to 18.7 MPa. Results were the same for all three crowns. Additional higher stresses were found at the boundary conditions; consequently, they were not taken into account.

MODEL 4-In the 3D model with a 1.5mm occlusal reduction, the Horizontal Bruxing load of 200 N was applied to the buccal cusps and the central fossa of the Monolithic and Multilayer Zirconia crown.

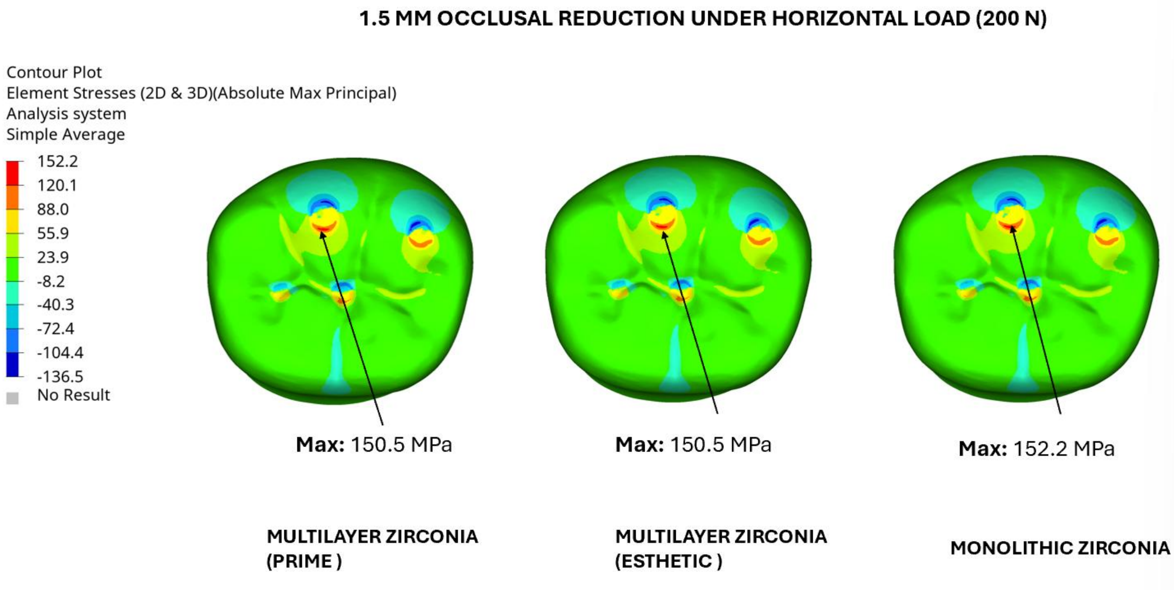


Figure 21. Maximum Principal Stresses on the Cameo Surface of 1.5 mm Multilayer and Monolithic Crowns under a Horizontal Load of 200 N.

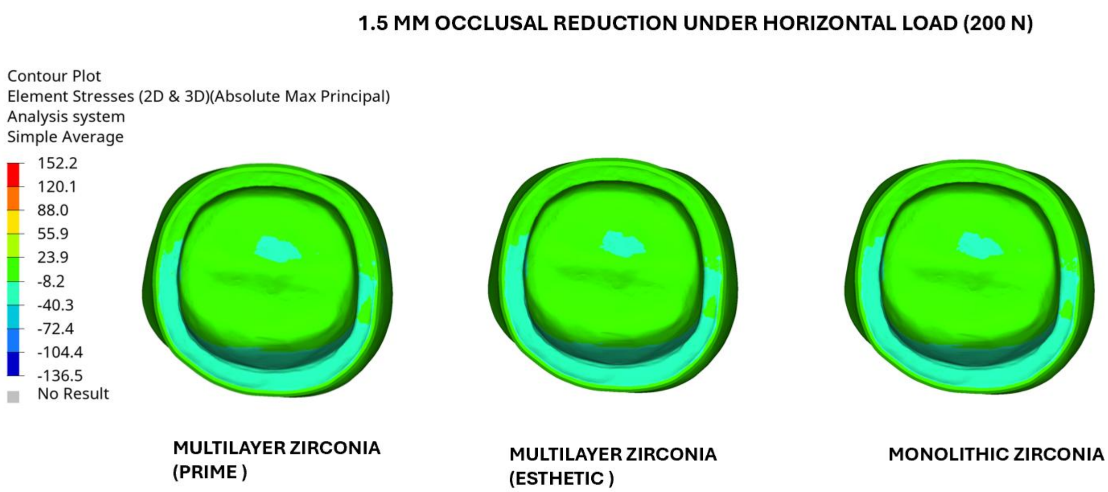


Figure 22. Maximum Principal Stresses on the Intaglio surface of 1.5 mm Multilayer and Monolithic Crowns under Horizontal load of 200N.

Crown - The tensile stresses were observed on the lingual slopes of the buccal cusps and the central fossa, with the maximum principal tensile stress (Red) of 150.5 MPa, 152.2 MPa, and 150.5 MPa on the cameo surface of the Multilayer ZirCAD Prime, ZirCAD Prime Esthetic and Monolithic Zirconia crowns, respectively. The buccal slopes of the buccal cusps exhibited compression areas, with a range of 72.4 MPa to 136.5 MPa. The intaglio surface of the crowns was found to be under a tensile load of 23.9 MPa. Additionally, areas of compression were observed on the internal aspect of the occlusal portion where loading was applied, as well as on the buccal cervical margin of the crown.

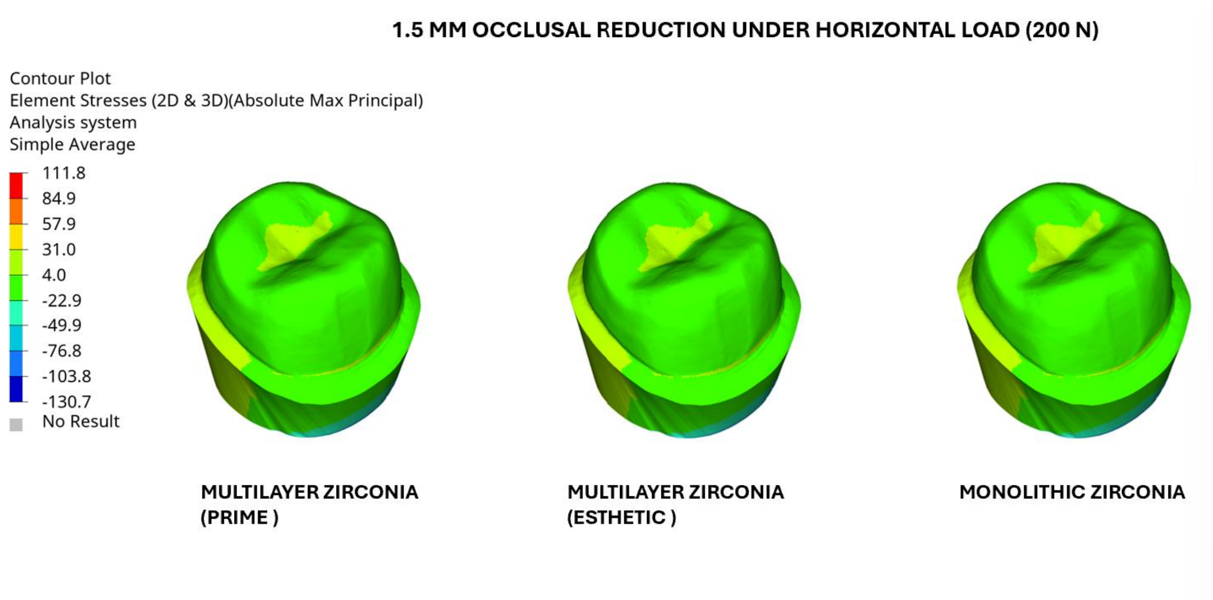


Figure 23. Maximum Principal Stresses on a 1.5 mm Occlusal Tooth Preparation Underneath Multilayer and Monolithic Crowns under Horizontal Load of 200 N.

Tooth Preparation - Most of the tooth structure is subjected to fluctuating stresses in the range of 4.0 MPa (tensile) to 22.9 MPa (compressive), with the central fossa and the margin under tensile stresses ranging from 4.0 to 31.0 MPa. Additional higher stresses were found at the boundary conditions; consequently, they were not taken into account.

Among the various combination of variables, the Maximum Principal stress values in the crown ranged from 90.4 to 154.4MPa. Regarding the thickness of the crown, 1mm thickness had more tensile stresses as compared to the 1.5mm thickness for all the materials and all loading conditions. Hence, the lowest Maximum Principal stress in the crown was for 1.5mm under the axial load of 540N. The highest maximum principal stress in the crown was for 1mm under the Horizontal load of 200N. Regarding the type of crown material, there was no difference in the Maximum principal stresses generated in the two Multilayered Zirconia and Monolithic Zirconia; hence, all materials behaved similarly under different scenarios.

CHAPTER 4

DISCUSSION

The study aimed to evaluate and compare the mechanical behaviour of Multilayer Zirconia crowns using Nanoindentation testing and Finite Element Analysis (FEA). Nanoindentation testing was performed using the Hysitron Nanoindenter (Fort Garry Campus, University of Manitoba) on two Multilayer Zirconia Samples, ZirCAD Prime and ZirCAD Prime Esthetics, and the modulus of elasticity and hardness were calculated. For Monolithic Zirconia, the values were obtained from the literature. These values were used in the Hypermesh software to calculate the Maximum Principal stress in full-coverage crowns under vertical loading of 540 N and horizontal loading of 200 N, simulating the maximum biting forces and bruxing forces.

Nano-indentation testing

The Nano-indentation testing revealed statistically significant differences in the modulus of elasticity between the two multilayered zirconia systems, ZirCAD Prime and ZirCAD Prime Esthetics. The ZirCAD Prime Esthetics demonstrated a slightly higher overall average modulus of elasticity compared to the ZirCAD Prime. Hence, the null hypothesis 1 was rejected. These findings suggest that the ZirCAD Prime Esthetics might be a slightly stiffer and more resistant material to deformation. The elastic modulus or Young's modulus values for these multilayer Zirconia are not provided in the company's brochures, nor are they easily found in the literature.

FEA testing

The posterior teeth exhibit a complex geometry, which presents a challenge for in vitro testing; sometimes, reproducing a particular set of tooth preparation variables and materials under different loading conditions is not practical. FEA can vary the material and specimen configurations in Computer-aided design (CAD) and mechanical loading software and predict stress distributions and mechanical behaviour of the crowns. In the present study, crowns of two different thicknesses (1 mm and 1.5 mm) were designed using two multilayer zirconia materials, ZirCAD Prime and ZirCAD Prime Esthetics (elastic modulus obtained from nanoindentation testing), and compared with monolithic Zirconia (data obtained from the literature). When these crowns were tested under FEA loading in both vertical and horizontal directions at 540N and 200N, it was found that the stresses induced in the two Multilayer and Monolithic crowns of the same thickness were similar. No previous studies have compared the maximum principal stresses in these Multilayer Zirconia materials with each other and with monolithic Zirconia. The similar performance of the Multilayer Zirconia and the Monolithic Zirconia provides us with information about the improvement in the strengths of the Multilayer Zirconia over the years. From a clinician's point of view, it allows us to select more aesthetically pleasing restorations in the posterior parts of the mouth, especially for more demanding patients.

Influence of the Material of the Crown on the performance of the crown

In the present study, the Maximum principal stresses induced in the crowns were the same between both multilayer zirconia and between multilayer and monolithic zirconia. Hence, Null Hypotheses 2 and 3 were accepted.

This result contrasts with a study conducted by Bakitian F, which showed that the strength gradient multilayer Zirconia with different yttria content (ZirCAD Prime) had lower mechanical strength than the shade gradient multilayer zirconia with the same yttria content in each layer (Katana HTML-3Y-TZP). The discrepancy in the results could be due to the method of testing the specimens, as in the study by Bakitian et al. the samples were separated into individual layers and each layer tested with three-point bending using the universal testing machine, as compared to the finite element analysis in the present study, where an average of elastic modulus of all the layers was used for testing the digital models in the software.

Another study by Badr et al. showed that the fracture resistance of Monolithic Zirconia was better than that of Multilayer Zirconia (ZirCAD Prime), where the crowns underwent thermomechanical loading under 110N Horizontal load, 1.2 million cycles at 5-55 degrees Celsius, simulating the normal chewing force. The reason mentioned by the author was the yttria content at the occlusal surface, which determines Zirconia's fracture resistance.

Additionally, Koo et al. investigated the transition zone of multilayer zirconia and found that it exhibited flexural strength, Vickers Hardness, phase distribution changes, and surface microstructure changes similar to those of 3Y-TZP; however, the surface roughness was lower for the transition zone compared to 3Y-TZP. Hence, my understanding is that it is beneficial to position the multilayer zirconia restoration within the blank so that the stress-bearing surface is in the transition zone, which will offer better esthetics and less surface roughness compared to a Monolithic Zirconia crown.

Influence of crown thickness on the performance of the crown

The crowns of 1.5 mm thickness performed better than 1mm as the Maximum Principal stresses induced in the crowns with 1mm thickness were more than 1.5 mm thickness for all the types of Zirconia materials under vertical axial loading of 540 N, hence thicker crowns distributed the load more favorably, reinforcing a study conducted by Rekow et al. which concluded that 1.00 mm thick crowns exhibited less maximum principal stress than 0.8mm crowns. Additionally, a study by Abdulmajeed et al. recommended a minimum thickness of 1.5 mm for all types of Zirconia crowns. Therefore, the Null Hypothesis H04 was rejected, as the thickness of crowns affects the maximum principal stress.

Influence of load direction on the performance of the crown

Another result of the present study is that the Maximum Principal Stress induced under Horizontal Loading of 200N is the highest in all crowns, regardless of the thickness of the crown; hence, Null Hypothesis 5 was rejected. This observation is following the results of a recent study by Ward et al. which showed that the maximum principal tensile stresses induced in molar teeth was highest under horizontal loading conditions, hence it is the most detrimental to the longevity of crown tooth complex, provided the Ultimate tensile strength of the material is less than the Maximum Principal stresses induced in the crown.

Another study in line with our findings is by Coelho et al., who also concluded that the combined loading of the horizontal and vertical loads increased Maximum principal stress values as compared to vertical loading alone.

The study conducted by Wicaksono et al. also aligns with the results of our present study, which indicates that the stresses induced during grinding, in the direction perpendicular to the initial contact, produce higher stresses than those induced by vertical loading.

These results confirm that the horizontal forces produced by bruxism may present a greater concern to the longevity of restorations than axial forces alone. These findings may have implications for supporting the use of splints in bruxing patients to protect the crown-tooth complex. According to Imanishi et al., when a vertical load was applied, all crowns exhibited tensile stresses concentrated around the load points on their surfaces. In contrast, when the load was applied horizontally, the tensile stress produced on the crown surface was substantially greater. This phenomenon produced under a horizontal load may be attributed to the bending and deformation of the tooth. Hence, low cusp angulation should be designed to decrease the stress concentrations in the crowns.

Resistance of tooth structure to the functional and parafunctional loading

Well-published literature on the ultimate tensile strength of tooth structures indicates a range of 33.9- 61.6 MPa, which exceeds the maximum principal stresses induced in the tooth by the maximum biting force of 540 N and the Bruxing horizontal forces of 200 N. Therefore, the tooth structure will withstand both the occlusal forces under both crowns as the higher elastic modulus of the crowns will provide sufficient strength and hence protect the underlying tooth structure.

STUDY LIMITATIONS

1. The interface between the tooth and crown was considered perfect, but this does not reflect the realistic model.
2. Nonlinear simulation of periodontal ligament properties, plastic and viscoelastic behaviours in materials, and tooth-to-tooth contact analyses were not considered.
3. The elastic modulus values for Monolithic Zirconia material is taken from reference literature.

FUTURE RECOMMENDATIONS

1. Risk of Failure of crowns can be calculated under different loading conditions, once we know the Ultimate Tensile strength of the material using Mohr-Coulomb failure criterion(Ward et al. 2025). The future recommendation is to measure the Ultimate Tensile strength of these new Multilayer Zirconia materials by in vitro tests, it will provide us with the information that crowns made up of different materials and under different loading conditions will survive or fracture especially in bruxer patients.

2. Nonlinear simulation of plastic and viscoelastic behaviours of periodontal ligament can be included in the models to compare the mechanical behaviours of both materials in intraoral conditions.
3. The testing can be continued on the incisors, canines and premolars to evaluate the difference in the impact of various loading angulations.
4. The testing can be done on the fixed dental bridges using these Multilayer Zirconia in posterior segments of mouth to evaluate their Maximal Principal stresses in complex scenarios.

CHAPTER 5

CONCLUSION

- The ultimate tensile strength of the tooth structures is around 33.9- 61.6MPa, which is more than the maximum principal stresses induced in the tooth by the maximum biting force of 540N and the Bruxing horizontal forces of 200N, so the tooth structure will withstand these occlusal loads.
- Increasing the thickness of the crown material reduces the stresses transmitted to both the crown and the tooth structure.
- More stresses were observed in the crown-tooth assembly under horizontal loading compared to axial/vertical forces.
- No statistically significant differences were observed between the different Zirconia crown materials.
- Multilayered Zirconia (ZirCAD Prime and ZirCAD Prime Esthetic) will behave similarly to the Monolithic Zirconia under the maximum biting forces and the Bruxing forces.

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APPENDIX

ANOVATwoWay (2024-04-30 08:24:01)

Descriptive Statistics

Brand

	N	Mean	SD	SEM	Variance	Missing	NonMissing
ZirCAD PRIME	58	87.83309	68.39725	8.981	4678.18365	0	58
ESTHETIC	60	156.99023	78.46901	10.13031	6157.38522	0	60

Region

	N	Mean	SD	SEM	Variance	Missing	NonMissing
Enamel	40	106.06523	83.01599	13.12598	6891.65396	0	40
Intermediate	40	156.87187	84.70623	13.39323	7175.14518	0	40
Dentin	38	105.16446	64.29131	10.42943	4133.37252	0	38

Overall

	N	Mean	SD	SEM	Variance	Missing	NonMissing
	118	122.99774	81.17679	7.47293	6589.67179	0	118

Interaction

		N	Mean	SD	SEM	Variance	Missing	NonMissing
ZirCAD PRIME	Enamel	20	51.18134	40.27058	9.00478	1621.71956	0	20
	Intermediate	20	81.09447	49.26507	11.01601	2427.04754	0	20
	Dentin	18	136.04463	84.05753	19.81255	7065.6681	0	18
ESTHETIC	Enamel	20	160.94911	78.63045	17.5823	6182.74732	0	20
	Intermediate	20	232.64927	14.56051	3.25583	212.00855	0	20
	Dentin	20	77.37231	3.29135	0.73597	10.83299	0	20

ANOVA

Overall ANOVA

	DF	Sum of Squares	Mean Square	F Value	P Value
Brand	1	134401.50791	134401.50791	47.22513	3.76032E-
Region	2	67390.73939	33695.3697	11.83966	2.16493E-
Interaction	2	239177.16364	119588.58182	42.02026	2.42573E-
Model	5	452242.47859	90448.49572	31.78121	4.91917E-
Error	112	318749.12097	2845.97429		
Corrected Total	117	770991.59957			

At the 0.05 level, the population means of **Brand** are **significantly** different.
At the 0.05 level, the population means of **Region** are **significantly** different.
At the 0.05 level, the interaction between **Brand** and **Region** is **significant**.

Means Comparisons

Tukey Test

Brand

	MeanDiff	SEM	q Value	Prob	Alpha	Sig	LCL	UCL
ESTHETIC ZirCAD PRIME	69.15714	9.82351	9.95601	3.40567E-	0.05	1	49.6931	88.62118

Region

	MeanDiff	SEM	q Value	Prob	Alpha	Sig	LCL	UCL
Intermediate Enamel	50.80664	11.9289	6.02331	1.26306E-4	0.05	1	22.47217	79.14111
Dentin Enamel	-0.90077	12.08484	0.10541	0.99694	0.05	0	-29.60564	27.8041
Dentin Intermediate	-51.70741	12.08484	6.05099	1.17177E-4	0.05	1	-80.41228	-23.00254

Interactions

Brand	Region	Brand	Region	MeanDiff	SEM	q Value	Prob	Alpha	Sig	LCL	UCL
ZirCAD PRIME	Intermediate	ZirCAD PRIME	Enamel	29.91312	16.87002	2.50762	0.48717	0.05	0	-19.00452	78.83077
ZirCAD PRIME	Dentin	ZirCAD PRIME	Enamel	84.86329	17.33229	6.92435	4.8032E-5	0.05	1	34.60518	135.12139
ZirCAD PRIME	Dentin	ZirCAD PRIME	Intermediate	54.95016	17.33229	4.48361	0.0235	0.05	1	4.69205	105.20827
ESTHETIC	Enamel	ZirCAD PRIME	Enamel	109.76777	16.87002	9.20183	4.73371E-8	0.05	1	60.85012	158.68542
ESTHETIC	Enamel	ZirCAD PRIME	Intermediate	79.85465	16.87002	6.69422	9.36449E-5	0.05	1	30.937	128.77229
ESTHETIC	Enamel	ZirCAD PRIME	Dentin	24.90448	17.33229	2.03206	0.70453	0.05	0	-25.35362	75.16259
ESTHETIC	Intermediate	ZirCAD PRIME	Enamel	181.46793	16.87002	15.21246	4.71259E-9	0.05	1	132.55028	230.38558
ESTHETIC	Intermediate	ZirCAD PRIME	Intermediate	151.55481	16.87002	12.70484	0	0.05	1	102.63716	200.47246
ESTHETIC	Intermediate	ZirCAD PRIME	Dentin	96.60465	17.33229	7.88237	2.59558E-6	0.05	1	46.34654	146.86275
ESTHETIC	Intermediate	ESTHETIC	Enamel	71.70016	16.87002	6.01063	6.21238E-4	0.05	1	22.78251	120.61781
ESTHETIC	Dentin	ZirCAD PRIME	Enamel	26.19097	16.87002	2.19559	0.63123	0.05	0	-22.72668	75.10862
ESTHETIC	Dentin	ZirCAD PRIME	Intermediate	-3.72216	16.87002	0.31203	0.99993	0.05	0	-52.63981	45.19549
ESTHETIC	Dentin	ZirCAD PRIME	Dentin	-58.67232	17.33229	4.78732	0.0123	0.05	1	-108.93043	-8.41421
ESTHETIC	Dentin	ESTHETIC	Enamel	-83.5768	16.87002	7.00624	3.77447E-5	0.05	1	-132.49445	-34.65915
ESTHETIC	Dentin	ESTHETIC	Intermediate	-155.27696	16.87002	13.01687	0	0.05	1	-204.19461	-106.35932

Brand's Grouping Letters Table

	Mean	Groups	
ESTHETIC	156.99023	A	
ZirCAD PRIME	87.83309		B

Means that do not share a letter are significantly different.

Region's Grouping Letters Table

	Mean	Groups	
Intermediate	156.87187	A	
Enamel	106.06523		B
Dentin	105.16446		B

Means that do not share a letter are significantly different.

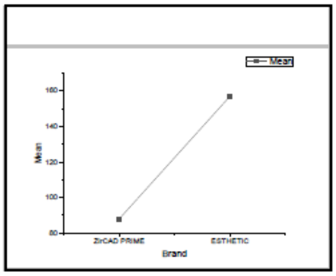
Interactions 's Grouping Letters Table

Brand	Region	Mean	Groups		
ESTHETIC	Intermediate	232.64927	A		
ESTHETIC	Enamel	160.94911		B	
ZirCAD PRIME	Dentin	136.04463		B	
ZirCAD PRIME	Intermediate	81.09447			C
ESTHETIC	Dentin	77.37231			C
ZirCAD PRIME	Enamel	51.18134			C

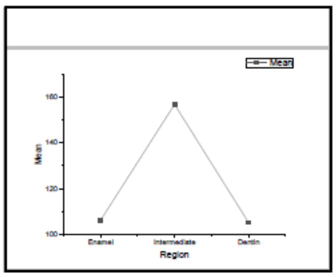
Means that do not share a letter are significantly different.

Sig equals 1 indicates that the difference of the means is significant at the 0.05 level.
Sig equals 0 indicates that the difference of the means is not significant at the 0.05 level.

Factor A



Factor B



Interaction