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Finite Element Analysis of Monolithic PEEK and Zirconia Fixed Dental Prosthesis (*In vitro Study)*

Essam Faisal[1], Tamer Hamza[2], Ashraf Mokhtar[3]

ABSTRACT

Background: Difference in the elastic modulus between zirconia and dentin may cause unfavorable stresses on dental abutments. A high-performance polymer has been recently introduced with similar elastic modulus in attempts to enhance stress distribution. ***Aim of the study:*** To compare the stress distribution of CAD/CAM milled PEEK on abutment teeth versus Zirconia Fixed Dental Prosthesis. ***Materials and methods:*** Two three-unit bridges were fabricated to replace a lower 2nd premolar on a stainless steel master model. A finite element analysis was done by 3D modeling of the two fixed dental prosthesis. Eight FDP models were simulated models and divided into two equal groups (n=4) according to the type of material used: Group Zr, monolithic zirconia and Group P, monolithic ceramic reinforced PEEK. The simulated models received a compressive load of 100N and an oblique load of 50 N at the pontic. The resultant stress ratios and deformations were analyzed by using **ANSYS ®. *Results:*** PEEK showed high von misses stresses and total deformation in the cement layer. Both restoration materials showed extreme stress values at the dentin interface under oblique loading. ***Conclusions:*** From a biomechanical perspective, PEEK transfer more stresses to the underlying cement layer. Under oblique loading, PEEK is generally safer to the underlying dentin than zirconia. ***Clinical implications:*** According to the current study, both materials can withstand chewing forces in the posterior region. However, PEEK restorations showed enhanced stress transfer which could lead to a shorter lifetime for the underlying cement.

Keywords: Monolithic, Zirconia, PEEK, Stress distribution, Finite element analysis.

INTRODUCTION

Ceramics were introduced owing to their superb biocompatibility and high aesthetic prospect. Dental ceramic types available on the dental market are of feldspathic, glass, glass-infiltrated alumina, and zirconia. Recently, an aesthetic posterior FDP with

1.Postgraduate Researcher, Fixed Prosthodontics Department, Faculty of Oral and Dental Medicine, Misr International University, Cairo, Egypt.
2.Professor of Fixed Prosthodontics, Fixed Prosthodontics Department, Faculty Dental Medicine, Al-Azhar University, Cairo, Egypt.
3.Professor of Fixed Prosthodontics, Fixed Prosthodontics Department, Faculty of Dentistry, Cairo University, Cairo, Egypt.

larger ceramic restorations has been achieved by the introduction of high-strength oxide ceramic.[1]

The new high-performance ceramic contains yttrium oxide (Y_2O_3) added to pure zirconium dioxide (Y-TZP). This innovation significantly increases its strength characteristics rendering it suitable as a framework material at the posterior region. 3Y-TZP exhibits excellent mechanical properties as its flexural strength reaches 900 to 1200 MPa and fracture strength about 9 to 10 MPa. It acquires compression strength of 2000 MPa. Tetragonal Zirconia Polycrystals has a modulus of elasticity of 210 GPa.[2]

One apparent drawback is the original color of zirconia which is white to ivory. Its opaque nature requires the material to be veneered by a translucent feldspathic porcelain in order to match the natural tooth color.[3] One of the main causes of failure in bilayered zirconia is chipping of the veneering porcelain. The rate of this cohesive failure is reported at 2% to 9% for single crowns after 2 to 3 years and at 3% to 36% for FDPs after 1 to 5 years.[4] This cohesive failure is attributed to the stresses induced by the difference in coefficient of thermal expansion between the zirconia core and the veneering ceramic.[5]

Full contour monolithic zirconia restoration was fabricated in an attempt to eliminate the cohesive failure in the veneering ceramic. Furthermore, it is of clear advantage in cases with reduced occlusal clearance as it could be milled in thinner sections.[6-8]

Manufacturers have altered the crystalline phase, supplied pre-colored zirconia ceramics, color liquids and liners in order to provide for esthetically pleasing monolithic zirconia restoration.[9] An increase in the cubic phase results in an increase in translucency as it reduces the tetragonal phase. In order to increase the cubic phase, a greater yttria content is required to stabilize zirconia which is approximately 7-9 wt %.[10] This alteration in the crystalline content diminishes the mechanical properties of conventional zirconia frameworks.[9]

Ozer et al.[11] evaluated the flexural strength of monolithic zirconia in an attempt to identify the recommended minimum thickness in load bearing areas. This in-vitro study was performed using 21 monolithic zirconia disks with different thickness under simulated masticatory stresses. The study reported that the mean flexural strength 0.8 mm thick monolithic zirconia surpasses the average masticatory forces, which renders the material clinically acceptable.

A new high performance polymer Poly ether ether ketone (PEEK) has been introduced to the dental market and has been investigated as a dental implant material. PEEK resembles human bone in terms of modulus of elasticity which is around 4 GPa. This property allows for even stress distribution of shear forces on dental restorations. Mastication forces are anticipated to be gently transferred along the material to the cancellous bone due to the similarity in elastic modulus.[12]

PEEK restorations acquire shock absorption potential which favors the theory to produce less stresses to the underlying bone and implant. A finite element study by **Tekin et Al.**[13] evaluated the difference in stress distribution between PEEK crown and post system and metal crown and glass fiber post. The study concluded that the use of PEEK material as both post and crown material reduces the stresses occurring in the post cement area. The study also shows that crowns fabricated from PEEK reduced stress peaks in the crown-cement interface.

Another in vitro study by **El Shahawy et al.**[14] evaluated fracture resistance of PEEK and zirconia frameworks. The authors concluded that ceramic reinforced PEEK (BioHPP) frameworks acquire fracture resistance that surpasses that of zirconia frameworks. This claims that BioHPP is a reliable material and can withstand fracture loads in the posterior region. Nevertheless, the author did not specify the type of zirconia used in the control group which limits the credibility of the study.

Therefore, the study aims to evaluate the stress distribution of three-unit monolithic PEEK FDP in comparison to monolithic zirconia. The study is conducted by a finite element stress analysis to accurately investigate load distribution on abutment teeth and supporting structures.

MATERIALS AND METHODS

A total of two fixed dental prosthesis were fabricated and simulated into eight 3D dimensional models and divided into two equal groups (n = 4), according to the type of material used. Group **Z** (monolithic Zirconia Katana STML), and Group **PC** (monolithic PEEK Bio.HPP). To evaluate the stress distribution of the materials, the models were subjected to static and oblique loading until failure. The stress distribution was evaluated using a computer software (ANSYS® program).

A specially designed model containing two stainless steel dies and a polymer base was fabricated by computer numerical control to simulate a mandibular first premolar and first molar. The dies were prepared having flat occlusal surfaces and were cylindrically shaped with 6° taper and 1 mm circular shoulder finish line. The premolar and molar dies had a diameter of 7 and 8 mm with a uniform height of 5 mm.

The distance between the centers of the holes to receive the dies in the base were 16.5 mm. **(Figure 1)**

Figure (1): Stainless steel Model.

Scanning, Designing and Construction of Restorations

The stainless steel model was sprayed by an antireflection scan powder (SHERA-Scan spray, extra oral CAD/CAM application, Germany) to remove the optical highlights from the surface of the dies and scanned by a desktop scanner (Identico hybrid scanner, Medit, Korea).

The acquired STL (standard tessellation language) file evaluated for clarity and precision by a dental design computer software (Exocad Dental CAD2.2 software, Valletta, Germany) and a 3D prosthesis model was designed**.** To create the restoration, the cement space was set to 100 μm. An occlusal thickness of 2 mm from the center of the abutments was set to standardize the production of the bridge. Connector size was done according to manufacturer's instructions with $4.3\times5.1mm^2$ for the molar abutment and $4.2\times4.8mm^2$ for the premolar abutment. After thorough editing and verification of the seating and margin design, the acquired data was subsequently sent to the milling machine unit (**Roland** DWX-**52D**, **Roland** DGA Corporation, Irvine, CA, USA).

The Monolithic Zirconia (Katana STML) and Monolithic PEEK (Bio.HPP) blocks were inserted and fixed into the milling machine by special clamps (**Roland** DWX-**52D**, **Roland** DGA Corporation). The restorations were milled using special burs for gross milling, final adjustments, and finishing. PEEK FDPs were dry milled using special milling burs for thermoplastic materials (breCam.cutter) with speed of 19000 rpm at a feed rate of 15mm/s. The diameter of the tool was 2 mm with a feed depth of 0.5 mm. The milling chips were collected during the milling processes. The restorations were separated from their corresponding disks and the projections where the sprues were attached were finished and smoothened by finishing stones. Polishing of the PEEK FDPs was done by a special silicone polisher (Ceragum Wheel, Bredent, Senden, Germany) and polishing paste (Abraso-Starglanzasg, Bredent, Senden, Germany) for three minutes. Sintering of zirconia FDPs was performed by using ceramic furnace (inFire HTC speed, Dentsply Sirona, Germany). The temperature is increased by 10°C/minute to reach 1550°C which is sustained for a two-hour holding time. The temperature is then reduced by a rate of 10 °C/minute which takes seven hours **(Figure 2)**. The restorations were visually evaluated for accuracy before seating on their corresponding models.

Figure (2): Zirconia CAD/CAM milled Fixed dental prosthesis.

Finite elemenet analysis:

A Three dimensional (3D) finite element model was constructed by acquiring the geometry of the FDP using a laser scanner (Geomagic Capture, 3D Systems, Cary, NC, USA). The scanner produced data file containing a cloud of points coordinates which requires an intermediate software (Rhino 3.0 - McNeel inc., Seattle, WA, USA) to trim a newly created surface by the acquired points. Subsequently, the bridge geometry was exported to finite element program as STEP file format.

Cement layer of 100 µm was created by using a set of vertical and horizontal planes and applying a set of Boolean operations (scale, divide, cut, add, subtract ... etc.) to keep the cement layer separated from the restoration. Bone geometry was simplified and simulated as two solid cuboid shapes. The inner one represents the spongy bone (38 mm width, 23 mm height, and 18 mm depth) that fills the internal space of the other cuboid shape (shell of 1mm thickness), representing cortical bone (40 mm width, 25 mm height, and 20 mm depth) **(Figure 3).** Another set of Boolean operation was used to generate roots cavity inside bone.

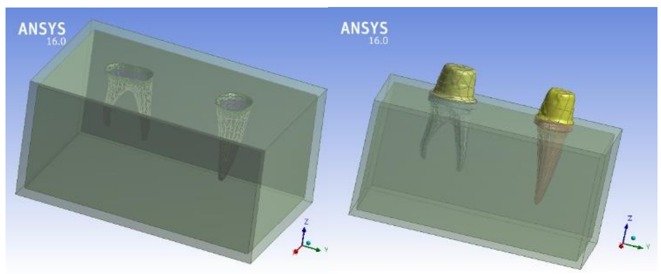

Figure (3): Simplified bone and cement layer.

All materials that used in this finite element study were assumed to be homogenous, isotropic and possess linear elasticity. Their properties were listed in (**Table 1).** The elastic modulus and Poisson's Ratio were used for defining the linear part of the stress strain curve of the isotropic materials. Parabolic tetrahedral element was used for meshing the model. Mesh density of all model components is presented in (**Table 2).**

Table (1): Material's properties of assembly components.

Material	Modulus of elasticity in MPa	Poisson's ratio
Cortical bone	**14,600**	**0.30**
Cancellous bone	**1,400**	**0.30**
Dentine	**18,600**	**0.31**
Cement (100µm)	**8,000**	**0.30**
Bridge (Zirconia)	**200,000**	**0.31**
Bridge (PEEK)	**4,550**	**0.37**

Table (2): Mesh Density.

Material	Number of nodes	Number of elements
Cortical bone	**72,384**	**45,174**
Cancellous bone	**158,571**	**112,737**
Dentine	**200,728**	**143,057**
Cement (100µm)	**35,086**	**21,310**
Restoration (Bridge)	**105,985**	**74,625**

After meshing the model, two load cases were applied to each bridge material. Loads were applied to the central fossa of the pontic. A 50 Newton compressive or oblique 45º as test loading was applied on each bridge material to estimate each model component behavior. Linear extrapolation was used to estimate the value of the compressive load that causes failure in the models. The resultant stresses (MPa) and deformations (mm) under each loading condition were calculated by ANSYS® program.

Thus, the total of eight runs were performed on the model (four per bridge material). The base of hollow cuboid representing the cortical bone was set to be fixed in place as a boundary condition. The solid modeling and finite element analysis (linear static analysis) were performed on a Workstation (HP Z820, with Dual Intel Xeon E5-2660, 2.2 GHz processors, 64GB RAM).

RESULTS

Each figure (showing one type of stress, strain, or deformation) appeared as a color distribution on the component as presented in **Figure 4.** The distribution ranged from maximal resultant values in red color, while the minimal values were represented in blue color. The analysis provided information on the distribution of vertical deformation (Uz), total deformation (Ut), Von Mises stress (equivalent stress), shear stress, maximum principal stress "tensile", and minimum principal stress "compressive" on each component of the case studies of Zirconia and PEEK restorations under vertical and oblique loading at central fossa of the pontic. Tabulating and comparing the extreme values can lead to conclusion and recommendations for each component behavior.

Results comparisons:

Figures 5 to **9** compared extreme values of total deformation (Ut) and Von Mises stresses on each component appeared in the

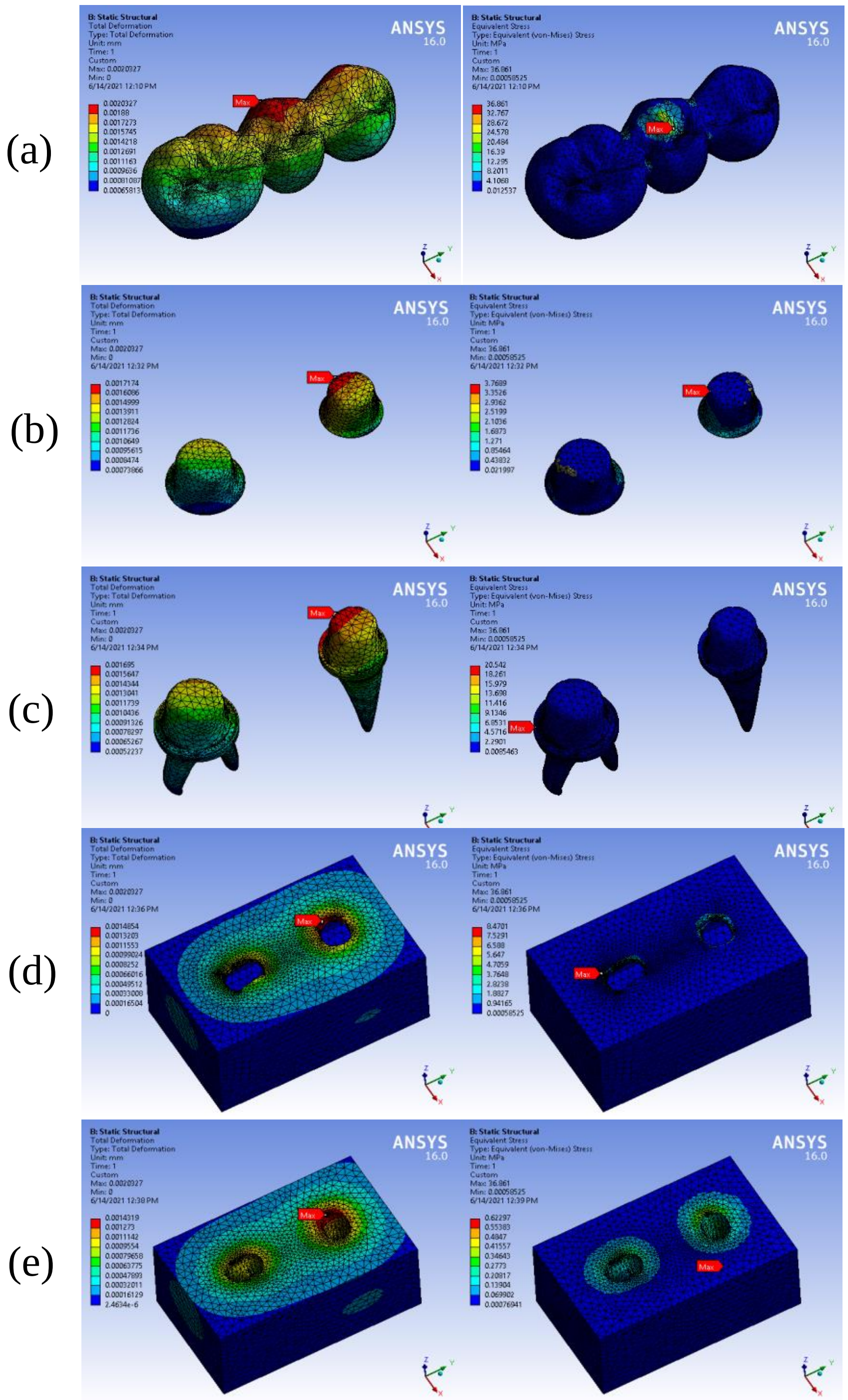

Figure (4): Shows total deformation and von mises stresses for **(a)** Zirconia restoration. **(b)** Cement layer. **(c)** Abutment teeth and roots. **(d)** Cortical bone. **(e)** Spongy bone.

eight cases. Odd numbered columns (Zr Test, PEEK Test, Zr Test-ob, PEEK Test-ob) show the cases where the applied load was 50N in vertical or oblique directions. Even numbered columns (Zr Failure, PEEK Failure, Zr Failure-ob, PEEK Failure-ob) show the cases of maximum applied load required for the material to fail in either vertical or oblique direction.

In **Figure 5** PEEK restoration showed 35MPa under 50N vertical loads. In accordance with linear extrapolation, this indicates restoration failure at about 110N vertical loading. Thus, a case of vertical load of 110N on PEEK restoration cause failure in restoration body at the connector.

The thick cement layer did not fail under loading in the eight cases (**Figure 6**). Cement layer reached critical values close to yield point under PEEK restoration cases and Zirconia restoration under oblique loading. Extreme values were altered between finish line and the occlusal surface under first premolar abutment.

Maximum Von Mises stress values on the dentin in (**Figure 7)** indicated failure under oblique loading of 50N or more regardless of the restoration material. The tooth structure fails under such low value of loading.

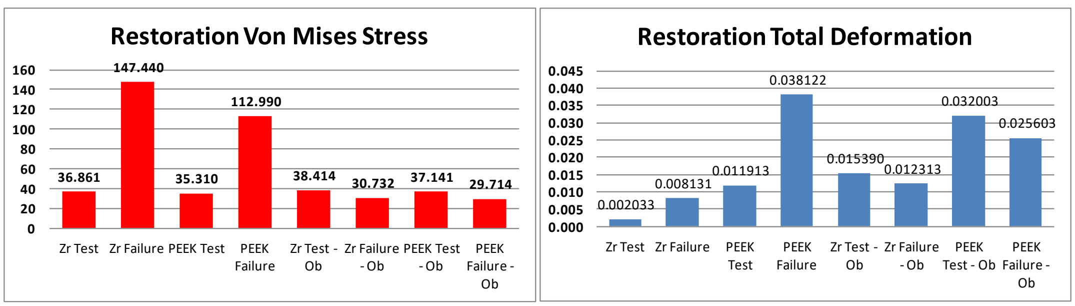

Figure (5): Restoration results comparison in the eighth cases.

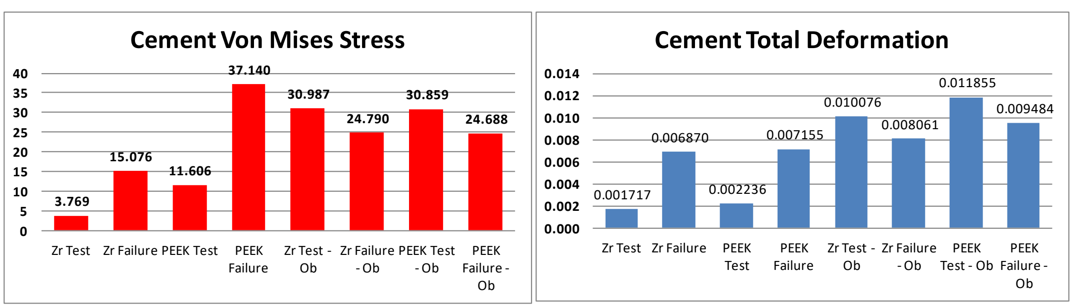

Figure (6): Cement layer results comparison in the eight cases.

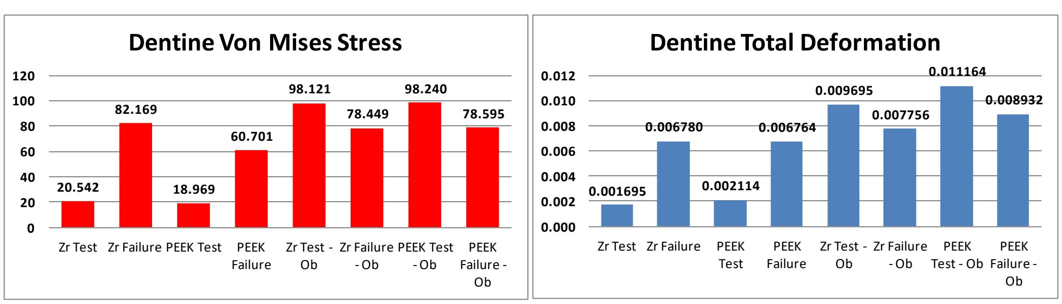

Figure (7): Dentin results comparison in the eight cases.

All values of stresses and deformation that appeared on cortical bone (**Figure 8**) were within physiological limits. Therefore, no failure in the cortical bone is expected to be seen which indicates the safety of the restorations. Maximum stresses did not exceed 41MPa that is less than half of yield strength. Total deformation did not reach 6 microns.

All values of stresses and deformation that appeared on spongy bone (**Figure 9**) were within physiological limits. Therefore, no failure in the spongy bone is expected.

it is the clinical standard.[15] The dies were attached to an acrylic base to simulate supporting structures. The premolar and molar had a diameter of 7 and 8 mm with a uniform height of 5 mm simulating an occlusal reduction of 2 mm. The distance between the centers of the holes to receive the dies is to provide sufficient space for the pontic and accommodate a restoration with adequate connector size.[16]

Digital impression was taken using Medit Identico hybrid scanner, which is in agreement with a study by **Seelbach et al.**[17]

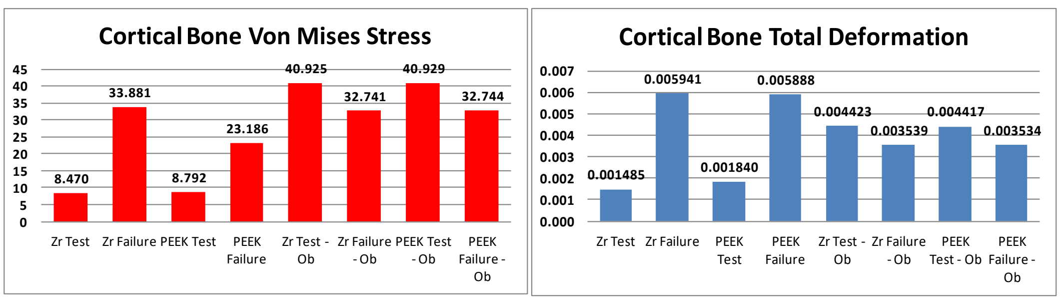

Figure (8): Cortical bone results comparison in the eight cases.

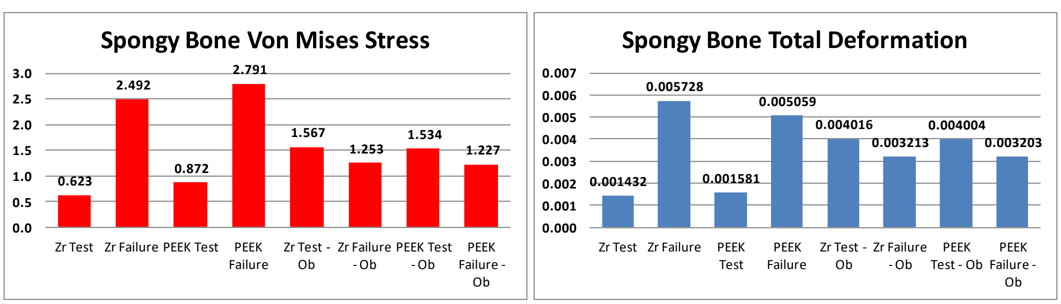

Figure (9): Spongy bone results comparison in the eight cases.

DISCUSSION

A model was fabricated to simulate abutment teeth and to ensure standardization. Stainless steel abutments were cylindrically shaped with 6° taper to resemble the preparation taper in premolars and molars as

that concludes the use of digital scanner as a reliable impression tool for the fabrication of crowns and short span fixed dental prosthesis.

CAD/CAM systems was used to produce dental restorations through standardized manufacturing processes. This allows for optimal replication of restorations.[18]

Katana STML is a multilayer 4 mol% yttria stabilized zirconia. It aims to balance mechanical properties and the optical properties of the ceramic to allow its suitability as a monolithic restoration. It acquires flexural strength of approximately 600 MPa. This renders the ceramic suitable for three-unit prostheses involving molar restoration.

A study by **Pöppel et al.**[19] supports the use of Katana STML in posterior FDPs as it is categorized as class 4 ceramic according to the International Organization of Standardization ISO6872:2015. However, the study only discourages the use of Katana STML as a posterior FDP in bruxer patients where the chewing forces reaches values above 1000 N.

BioHPP PEEK crowns offer a great alternative for patients with parafunctional habits. This is because they have the ability to withstand the high chewing forces without cracking. BioHPP is high performance polymer derived from PEEK with 20% volume ceramic fillers. It acquires a modulus of elasticity of about 4GPa which resembles dentin and human bone. This similarity in young’s modulus allows the material to evenly distribute stresses and chewing forces along the dental structures.[20]

The cement space was set to 100 µm in accordance with the findings of **Ha et al.**[21] who studied the effect of different cement thicknesses on the stress distribution in monolithic zirconia crowns. The study recommended the use of thinner cement layers as stresses significantly increased in the cervical areas with cement thickness above 100µm.

It was hypothesized that the restoration material will influence the stress distribution of three-unit fixed dental prosthesis. This hypothesis was accepted as the stress distribution of both materials varied significantly.

The FEA in the current study aimed to find critical loads that may cause failure in the restoration. The anatomy of the abutment teeth (first premolar & first molar) is not the same in geometry which resulted in tendency to have more stresses and deformations towards the smaller resting side (first premolar).

Results showed that Zirconia restoration can withstand the compressive loads up to 200N. On the other hand, PEEK restoration failed (start cracking) at about 110N vertical loading. Both restoration materials did not fail under 50N oblique loading; however, the stresses exceeded the yield strength of the cervical margin of abutment teeth and the cement layer. This could indicate failure to the underlying cement layer caused by stress peaks at the margin in comparison to the more rigid zirconia restoration.

A 2018 finite element study by **Dal Piva et al.**[22] compared the stress distribution between different monolithic crowns acquiring different modulus of elasticity. The study reports that low elastic modulus restorations such as PEEK allow the passage of stresses to the cementing layer. This was deduced by recording higher values of shear stress in the cementing layer of PEEK restorations. The study also concluded that zirconia restorations distributed minimal stresses in the cement layer. This is due to their high elastic modulus which allows the rigid material to retain the stresses within its own confinement. During the current study, the cement layer (100µm) showed relatively high stress values that are in proximity to its yield stress under static loading.

The difference in geometry of the supporting teeth affects the behavior of deformation. The smaller one (first premolar) received the highest stress at finish line where the restoration tends to deform/micro-move towards the smaller resting tooth. This finding conforms to a finite study by **Reimann et al.**[23] who evaluated the stress distribution for a three-unit anterior bridge of different connector sizes. They concluded that the smaller abutment receives higher stress due to presence of smaller connector size. The study reports that an increase of 1 mm^2 in the connector size leads to 16% reduction in the bridge deflection. They also reported that an increase in the connector dimensions could lead to a 30% reduction in stress levels.

A 2018 study by **Mazen A. Attia.**[24] evaluated the effect of material type on the stress distribution for three-unit fixed dental prosthesis. The study compared between Zirconia, PEEK and nickel chromium materials for different elastic modulus. He concluded that the highest recorded von misses stresses were located at the preparation line of the abutment teeth towards the pontic regardless of the type of material used. This finding proposes the optimal attention towards the interface between the preparation line and the FDP.

This goes in agreement with the findings of the current study as both restoration materials showed high stress peaks at the preparation line of abutment teeth under oblique loading with about 50N. Under vertical loading on Zirconia restoration, again the remaining tooth finish line region towards the pontic failed under approximately 200N. This finding matches mechanisms of load transfer. The results indicate that PEEK restoration might slightly be advantageous in terms of stress distribution for dentine due to its resilience which absorbed higher amount of applied load energy in comparison to Zirconia one.

Cortical bone crest received the extreme stress values, while all these values were within physiological limits for bone. Spongy bone work such as shock absorber for the teeth will never fail under both restoration materials and all types and values of tested loading. Both restoration materials show comparable von misses stresses distributed on the spongy bone; however, PEEK restoration shows slightly reduced bone deflection.

Controversially, **Sirandoni et al.**[25] reports that polymer-based restorations show increased total deformation values on bone in comparison to restorations with higher modulus of elasticity.

The current finite element study has some limitations in that it does not fully represent the oral environment conditions. Fracture resistance, dynamic loading, salivary functions, PH and thermal fluctuations are important factors that reduce the mechanical properties of the materials. Furthermore, periodontal ligaments were not simulated in the study which could have a significant influence on the supporting structures and should be inspected in future research.

According to the current study from a biomechanical point of view, both restorations perform favorably under static loading rendering both materials suitable to withstand chewing forces in the posterior region. However, PEEK restorations showed enhanced stress transfer which could lead to a shorter lifetime for the underlying cement.

CONCLUSION

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

1) Zirconia restoration material does not fracture under the applied vertical or oblique loads, while it may damage underlying structures.
2) PEEK restoration material will fail under vertical loading, while under oblique loading cause failure to the underlying structures
3) For PEEK restoration, even without reaching critical loads that may crack the cement layer, the cement layer will fail shortly under cyclic loading in comparison to Zirconia one.
4) Finish line/ abutment interface needs extreme level of care not to cause failure as in this study where three out of four cases failed at supporting tooth finish line.
5) Bone is generally safe under both restoration materials and tested loading conditions.

CONFLICT OF INTEREST: The authors

declare no conflict of interest.

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