https://doi.org/10.48047/AFJBS.6.12.2024.4036-4050

African Journal of Biological Sciences

Journal homepage: http://www.afjbs.com

Research Paper Open Access

Marginal Adaptability and Fracture Resistance of Bio HPP Three-unit Fixed Dental Prosthesis Framework Following Thermomechanical Aging: In-vitro Study

Fatma A. Hasaneen*, Abeer Atef Younes*, EmanM.Sobhi El Bahrawy, Radwa Eldessouky***

* Lecturer of Fixed Prosthodontics, Faculty of Dentistry, Tanta University, Tanta, Egypt.

** Assistant Professor of Dental Biomaterials, Faculty of Dentistry, Tanta University, Tanta, Egypt.

Corresponding author:

Dr Fatma A. Hasaneen

Department of Fixed Prosthodontics, Faculty of Dentistry, University of Tanta

Tel: +20 106 112 7028

Postal code:31111

E-mail: fatma.morssy@dent.tanta.edu.eg

This research did not receive any specific grant from funding agencies in the public, commercial.

No conflict of interest.

Article History
Volume 6, Issue 12, 2024
Received: 02 Jun 2024
Accepted: 30 Jun 2024
doi:
10.48047/AFJBS.6.12.2024.4036-4050

ABSTRACT

Purpose: To evaluate the Yttrium-oxide partially stabilized zirconia (Y-TZP) to Bio HPP 3-unit fixed dental prosthesis (FDP) frameworks in terms of marginal adaption and fractureresistance.

Methods: Using a standardized 3-unit FDPs framework, two groups (n = 10/group) were created, Group 1: Zirconia (Z) and Group 2: Bio HPP(B). Using a USB digital microscope with built-in camera connected to an IBM compatible personal computer (50 X) and each specimen was photographed before cementation and after thermodynamic aging to establish marginal adaptability.After thermodynamic aging, fracture resistance was measuredusing Universal Testing machine. Compressive static force was applied at thecenter of pontic'socclusal surface, with a cross-head speed of 1 mm/min. The load required for fracture was recorded in Newton.

Results: 3-way ANOVA followed by pair-wise Tukey post-hoc test indicates statistically non-significant difference in marginal gap mean value (P=0.0825>0.05) between Zirconia(24.31±3.1µm) and Bio HPP (23.27 ±2.13µm).

Regardless the type of material and abutment, with a p-value of <0.0001>0.05, the mean gap size after aging(30.88 ±3.3 µm)was found to be statistically significantly higher than before cementation. Student t-test (t=2.3, p=0.003<0.05) revealed that Bio HPP had a statistically significant (p<0.05) greater fracture resistance mean value (1828.41±181.41N) than Zirconia (1441.23±165.3N).

Conclusions: In contrary to zirconia, Bio HPP FDP showed non-significantly greater marginal adaptation; still, Bio HPP outperformed zirconia in terms of fracture resistance. Marginal adaptability of

thermo-mechanical aging was significantly lower in all frameworks examined, although it was still within a clinically acceptable range.
Keywords: Marginal adaptation, Zirconia, Bio HPP,fracture resistance.

Introduction

Different indirect dental restorations generated has dramatically increased in recent years due to advancements in dental materials and technology[1, 2]. Compared to traditional manufacturing methods, computer-aided design, and computer-aided manufacturing (CAD/CAM) technology allows for better results, higher time-effectiveness, greater predictabilityand precision. [3]. Marginal fit, fracture resistance, and aesthetics are variables that influence the way a dental restoration performs. [4].

Owing to its chemical stability, good biocompatibility, superior compressive strength, and satisfactory aesthetics, zirconia is the most utilized core material in ceramic prosthesis. [5, 6].

However, one of the most frequent problems with zirconium restorations is that overlaying the zirconium core with porcelain veneer leads the restoration to chip or laminate the veneer.[7-11].

However, the disadvantage of a fully-coverage Zirconia crown is that it abrades the opposing Natural tooth upon formation of the Occlusal surface with zirconia. In addition, clinical long term studies have shown that many Zirconia Restorations have poor marginal adaptation. The defective margins were found to be a major contributor to the high rate of unsuccessful restorations[12].

With similar wear characteristics in the ceramic range, Bio HPP may be an appropriate choice in dentistry for handling those problems. Bio HPP is resistant to nearly every organic and non-organic solvents and is biocompatible. It has excellent mechanical properties, can withstand high temperatures and good dimensional stability[11].

In addition, Bio HPP's elasticity is adaptable to the bone while zirconia's rigidness is 20 times that of bone. Bio HPP's elasticity makes it more natural material, as it compensates for bone torsion when occlusal force is applied, especially in larger implant works and long frameworks. It also does not cause abrasive damage to the remaining teeth.[13]Bio HPP's superior physical and biological characteristics make it an ideal superstructure in dentistry, such as dental implants, temporary abutments, and framework for FDPs.[14-16]

Polyether-ether ketone (PEEK) is a partially crystalline high-performance thermoplastic polymer that has a low melting point (343°C) and may be handled in a variety of techniques, PEEK is the basis for Bio HPP. In the dental technical lab, one approach is to press the material using a vacuum pressing machine. Bio HPP can be industrially pre-pressed into granules or pellets. The preheated muffle is placed inside a vacuum pressing machine and pushed as part of the pressing procedure. Another option is to mill the material using CAD/CAM technology, pressurizing the Bio HPP blank industrially utilizing established parameters like as temperature, pressure, and time. The same basic materials used to make Bio HPP FDPs can be used for all these fabrication techniques. [17]

One of the most crucial elements in evaluating a restoration's long-term success is marginal adaptability. Nonetheless, its value was considered acceptable for ceramic restorations up to 120 μm in clinical situations[18].Inadequate marginal adaptation can cause periodontal disease, secondary caries, dental plaque accumulation, and eventually tooth loss[19].The marginal adaption for ceramics repair has been evaluated earlier using both destructive and non-destructive approaches. Three non-destructive techniques were used: optical coherence tomography (OCT), resin replica with scanning electron microscopy (SEM), and silicone replica with stereomicroscopy. Using a stereomicroscope to measure samples and slice them were examples of destructive approaches [20-24].

The most prevalent cause of dental prosthesis replacements is fractures. Thus, it's critical to assess ceramic's fracture resistance in a variety of clinical situations prior to employing it as a permanent dental replacement[25]. Artificial aging was used to better assess fracture resistance in clinically simulated conditions. The tested groups were subjected to thermomechanical aging with dynamic loading combined with thermocycling[26, 27].

Zirconia & milled Bio HPP FDPs Framework have been proven to be effective in fixed prosthodontic. However, there is a lack of information in the literature on the performance of the bridges in terms of essential criteria for restoration success (margin adaptation, fracture resistance, etc.). Scare studies have reported on the effects of thermo-mechanical aging on Bio HPP Framework marginal adaptability and fracture resistance. Therefore, the objective of this study was to assess and contrast the fracture resistance and marginal adaptability of the 3-Unit Bio HPP and Zirconia FDPs frameworks after thermomechanical aging. The null hypothesis of this study was assuredness that the marginal adaptation and fracture resistance of Bio HPP FDPs frameworks don't differfrom those of Zirconiaframeworks.

Materials and Methods

Ethical approval:

This study was approved by Research Ethics Committee of Faculty of Dentistry, Tanta University(R-BIO-9-23-3059). The design and procedures of this study followed guidelines published by the Research Ethics Committee, Faculty of Dentistry, Tanta University.

Materials

Materials used in this study are listed in (Table-1). A total of 20 framework samples were selected for this study. The samples were randomly assigned to 2 groups (n= 10 each) based on the materials used. Group 1:manufactured from Zirconia blank (control group), and Group 2: manufactured from Bio HPP blank. Both materials milledby (CAD / CAM) milling machine.

Methods

Construction of the model:

Using a standardized computer numerical control (CNC) machine, a model that represented an FDP between 2ndpremolar and 2ndmolar was created with two steel abutments [28]. The abutments had a 1-mm circular shoulder and were machined cylindrical with a 6º taper (premolar: 7 mm diameter, molar: 8 mm diameter) and a height of 4 mm for premolar and 5 mm for molar[29,30]. The abutments were anchored to an acrylic block (Fig.1).Parallelism was ensured by a surveyor (Bredent GmbH &Co.KGt, Senden, Germany) [31].

20 impressions of steel abutment model were made using Replisil Silicone impression material (Replisil 22N, DENT-e-Con, Germany) which was thenflooded in epoxy resin material (Epoxy 150 Chemical Industries of Constraction CIC – Egypt) constructing a model from epoxy resin. [32] (Fig.2)

Construction of FDPs framework:

a) Milling procedures of Bio HPP frameworks

The Bio HPP blank was milled using a (CAD / CAM) milling machine (Imes-Imere Core GmbH, Eiterfeld, Germany) to form ten "3 unit" FDPs framework with a flat occlusal surface, 0.7 mm wall thickness, rectangular cross-section connectors (7.36 mm2), occluso-gingival height (3.2 mm), and bucco-lingual width (2.3 mm) after the steel model was scanned using a CAD / CAM optical scanner (SHERA Werkstoff-Technologie GmbH & Co. KG EspohlstrasseLemförde, Germany). [24,33]

The Bio HPP blank was connected to the dry milling device, and the breCAM.cutter (a milling tool specifically designed to match the material's properties) was used to complete

the milling process in compliance with the technical parameters of the dry milling system.Then, a diamond bur was used to remove the frameworks from the disk, and gentle air steaming was used to remove any cutting waste that was connected to the restorations. Following their extraction from the disk, the bio HPP frameworks were examined on the steel and epoxy resin models. (Fig.3).

b) Milling procedures of Zirconia frameworks

The milling process followed the same process as Bio HPP material. Restorations were then placed into the Refractory Saggar tray, and then into the Sintering Furnace. Restorations are then sintered in accordance with the Sintering schedule below (Table-2)[34].

After sintering the frameworks were removed from the furnace and checked on the steel and epoxy resin model. After sintering, the frameworks were subjected to abrasion by airborne particulates Al_2O_3 (average particle size 110 µm) for 15 seconds at 2.5 Bar air pressure and working distance 7 mm[35].

Evaluation of marginal adaptation before cementation:

The specimens were secured over their respective epoxy dies with a special holding device that held the specimen firmly in place during the margin photography. Each sample was photographed using a USB Digital Microscope with an integrated camera connected to an IBM compatible personal computer using a 50 X fixed magnification. Using a Digital Image Analysis System (Image J1.43U, National Institute of Health, USA), the vertical gap length was measured and evaluated.The Image J software expressed all boundaries, dimensions, frames, and measurable indicators in pixels.Thus, in order to transform the pixels to exact real-world units, system calibration was done. The technique of calibrating involves contrasting a scale produced by the image j program with an object of known size, in this case a ruler.For every item, the borders were captured on camera. Then, using three equally spaced markers at the middle of the cervical circumference for each surface on each abutment, morphometric measurements were carried out for each shot.For the premolar (Mesial, Buccal, Palatal) and for the molar (Distal, Buccal, Palatal), with measurement at each point was repeated 5 times.[36]

FDP framework cementation on epoxy resin models:

Glass ionomer cement (GC Gold Label Luting & Lining Cement, GC Corporation, Tokyo, Japan) is mixed (powder to liquid ratio is 1.8g/1.0g =1 level scoop of powder to 2 drops of liquid) and applied in the fitting surface of FDPs the according to manufacturer`s instructions, after that, they were bonded to the equivalent models made of epoxy resin.A specially made cementing equipment was machined to help apply a force of three kilograms using an Instron testing machine throughout the cementation process to ensure a uniform cement flow.[37]

Thermo-mechanical aging

Aging treatment was based on ISO 13356 specifications whereall specimens were exposed to 5000 thermocycles (55˚C-5˚C)in an automated thermocycling machine (Robota automated thermal cycle; BILGE, Turkey) with dwell times of 25 seconds in each water bath and a lag time of 10 seconds[27].Next, a chewing simulator (Robota, ACH- 09075DC-T, AD Tech Technology Co. Ltd, Germany) was used formechanical aging. The teeth with cemented bridge frameworkswere fixed in the Teflon housing of the sample holder and subjected to 60000 loading cycles at a frequency of 1.6 Hz under a weight of 10 kg (98 N) utilizing a metallic rod with a 3.8 mm diameter round tip parallel to the long axis while immersed under distilled water at 37˚C [26].

Finally, vertical marginal gap was evaluated after thermomechanical aging as described before.

Evaluation of fracture resistance:

With load cell 5 KN, each sample was individually connected to the Universal testing machine (Model 3345; Instron Instruments Ltd., USA), and data was recorded using computer software (Lloyd Instruments, Nexygen-MT).Samples were firmly secured by tightening screws to the testing machine's lower fixed compartment(Fig. 4).The fracture test was carried out by introducing a compressive static load occlusally at the pontic center while the testing machine's top moving compartment moved at a crosshead speed (1mm/min). The load required for fracture was recorded in Newton. [33, 38]

Statistical analysis

The analysis of the data was done in stages. Descriptive statistics are first presented for each group. For marginal gap test results, three-way analysis of variance ANOVA test of significance followed by pair-wise Tukey's post-hoc tests were done for comparing variables (material group, abutment and cement) affecting mean values. Pair-wise student t-test was performed to detect interaction between variable of significant effect. Student t-tests were used to determine the significance of the fracture resistance test findings between material groups. Asistat 7.6 statistics software for Windows (Campina Grande, Paraiba state, Brazil) was used to conduct the statistical study. For every test, P values < 0.05 were deemed statistically significant.

RESULTS

Marginal gap

Before cementation and after thermomechanical aging, the mean values and standard deviation of marginal gap measurements (μm) were recorded as a function of material group type. (Table-3)

Overall, 3-way ANOVA followed by pair-wise Tukey's post-hoc tests revealed that Zr group had a statistically non-significant superior marginal gap mean value (24.31±3.1μm) compared to Bio HPP group (23.27 ±2.13μm) ($P=0.09325 >0.05$). (Table-4)

A 3-way ANOVA followed by pairwise Tukey's post-hoc tests ($P=<0.0001<0.05$) revealed that the marginal gap after thermomechanical aging recorded a statistically significant greater mean value (30.88±3.3μm) than before cementation (22.41 ±2.1μm), regardless of the type of abutment or material. (Table-5).

Fracture resistance

The student t-test ($t=2.3$, $p=0.003<0.05$) revealed that the Bio HPP group had fracture resistance mean values that were statistically significant ($p<0.05$) greater than those of the Zr group.(Table-6)

There was non-significant an inverse correlation between fracture resistance and vertical marginal gap as demonstrated by Pearsonlinear correlation statistics ($R=-0.236$, $R2=0.0557$ and $p=0.5116>0.05$) (Fig.5).

DISCUSSION

This in vitro study assessed andcompared the vertical marginal deficiencyas well as fracture resistancebetween CAD / CAM 3-unit bio HPP FDP frameworks (test material) and Zirconia (control material)following thermomechanical aging.

Studies conducted in vitro offer controlled and ideal conditions for testing outcomes that would not be possible in vivo [38, 39]. Furthermore, several factors, including dental preparation, the impression technique, and the cementation technique, could make the testing process more difficult and deviate from the ideal clinical scenario [40, 41].

A fixed design with the same proportions was built after the original steel abutment model was scanned to assure standardization. [33]

For the marginal adaptation assessment, the epoxy resin models were used as they are very clear and allow good visibility in in measurementusing digital microscope.[42,43]
The marginal adaptation of dental restorations has been evaluated using a variety of techniques, includingimpression replication methods,cross-sectional views, direct viewing, and clinical examinations. Since the direct viewing approach is the most widely used technique for measuring marginal difference, it was selected for this study due to its non-destructive nature, speed, simplicity, and ease of use.[44]Nawafleh et al., 2013[22], support this method of marginal gap measurement.
Further, marginal discrepancy was carried out both before cementation and after thermodynamic aging since numerous researchers observed differences in the marginal gap between the two periods. [22, 45-47]This could be the result of improper cementation procedure, such as applying excessive finger pressure to the crown while using cement, which can lead to an uneven cement layer film thickness.[22] In this study, cementation was performed with the aid of specially designed cementing device underconstant load to ensure uniformity of cement flow across the axial wall of the specimens. Glass ionomer cement (GIC) was chosen, due to its good visibility in measurements as opposed to the resin cements. [42]
In a study by Groten et al., [48] the minimum number of in vitro measurements per specimen was [20-25]. The study employed thirty specimens, with each specimen having three equidistant markers along the cervical circumference for each premolar retainer (Mesial, Buccal,Palatal) and each molar (Distal, Buccal,Palatal). Measurements were made five times in every point. As the framework has a major impact on the overall adaptation of the final restoration[21,27,28], in this study the framework marginal gap was assessed without veneeringand recorded a mean value of (24.31±3.1μm) for Zr group and (23.27±2.13μm) for Bio HPP group. Marginal gaps measurements recorded in the current study were within the clinical acceptability.According to McLean and Fraunhofer [49], a 120 μm marginal discrepancy was clinically acceptable, which support our result.They also reported that the average marginal discrepancy for CAD / CAM ceramic crowns was between 23 and 110.1 μm. [39, 44, 49-52].
To simulate the clinical situation of complex oral environment, all tested groups were subjected to thermo-mechanical aging by application of 60000 loading cycles equivalent to 6 months clinically[26] and 5000 thermocycles which correspond to 6 months [27] of physiological aging in the oral cavity.
Numerous studies corroborated the current study's finding that the marginal gap recorded a much greater mean value (30.88±3.3 μm) following thermo-mechanical aging than it did before cementation (22.41±2.1 μm) [22-45-47,53].This can be attributed to the that the hot water exposure causes the accelerated hydrolysis of exposed collagen fibers and the extraction of inadequately polymerized resin tags. In addition to generated stresses at the tooth/restoration interface due tomismatch in the coefficient of thermal expansion of tooth structure and restorative material which have been suggested as a crucial factor for deterioration of the marginal adaptability [54]. Alterations in temperatures also exaggerate the problem due to a mismatch in the coefficient of thermal expansion of the resin matrix and the filler particles [55].
In addition, the results of this study agreed with those of Krejci et al,1994[56], who also identified a significant negative impact of thermo-mechanical aging on marginal fit of crowns.But this is against the findings of Beschnidt and Strub[57] who found that the aging procedure had no significant effect on the marginal adaptation.
Although the overall marginal gap mean values of all tested groups were increased after thermo-cycling aging but still below the clinically accepted limit of 120 μm[58, 59]. It may be attributed to the good mechanical reliability of the restoration and durable

adhesion obtained from adequate tooth structure, proper restoration preparation, and the usage of GIC cement [60].

In the current study the marginal gap recorded a mean value for Zr group(24.31±3.1μm) which wasnon-significantly higher than Bio HPP group (23.27±2.13μm). Previous results are line up with the study made by Young Park J et al., [60]who reportedthat marginal deficiency of Zirconia crowns(77.06 ± 32.14 μm)wasnon-significantlyhigher than Bio HPP (66.83±22.31μm) using Silicone replica technique. These values were higher than those found in the current study, however there could be a variation in the values since different evaluation techniques were used. [61]

These previous results could be becauseCAD/CAM zirconia have volume shrinkage rates of "22 - 25%" after sintering, this could have a negative impact on the fit of dental prosthesis.Bio HPP on the other hand does not show any shrinkage.It should exhibit greater fitness as a result of the lack of contraction and the sintering process.[61] Despite that Bio HPP group showed better marginal adaptation than Zr group it was nonsignificant, and both were within the limit of clinicalacceptability.

The gap measured following thermo-mechanical aging, regardless of the type of material, had a statistically significant greater mean value (30.88±3.3 μm) than the gap measured before cementation (22.41±2.1 μm). This was reported by many researchers [22-45-47]

Despite that the gap width after thermo-mechanical aging recorded statistically significant higher mean value (30.88±3.3 μm) than before cementation (22.41±2.1μm) it was within the limit of clinical acceptability.

In the present study, ***the fracture resistance*** of three-unit Bio HPP and Y-TZP FPDs framework was evaluated and compared in vitro. Compared with in vivo studies, in vitro experiments are less expensive and easier. [62]

According to this study, Bio HPP is created within of a framework. The material cannot be handled in an overall shape since it lacks aesthetic value [33,61].In order to ensure standardization, the original model with the steel abutments was scanned. A fixed design with the same dimensions (the wall thickness was 0.7 mm, and the connectors had an almost rectangular cross-section of 7.36 mm2, an occluso-gingival height of 3.2 mm, and a buccolingual width of 2.3 mm) was then used to produce the Bio HPP and Y-TZP FDPs framework using a CAD/CAM system.[33]

The epoxy resin models, whose elastic modulus is more akin to dentin than metal, were used to evaluate the fracture resistance of the Bio HPP and zirconia frameworks. [62, 63]In this study, Zirconia 3-unit FDPs framework showed a mean fracture resistance of (1441.32±156.3 N). This previous result is in agreement with othersin which fracture resistance of zirconia was 900 to 1400N. [63-65]

In the present study, Bio HPP 3-unit FDPs framework showed a mean fracture resistance of (1828.41±181.41 N).The previous results supported by Taufall S et al [66]who examined the fracture loads of several veneered PEEK 3-unit FDPs and found that the mean fracture resistance was 1882±152 N. B. Stawarczyk et al.[65]support the results of the current study but with higher values (2,354±422N), which may be due to increased connector thickness.

On the other hand, the fracture resistance of PEEK three-unit FDP frameworks showed mean value (1383 ± 149 N) in the study made bySarfaraz H et al[11]. These values are comparable but lower to those in the present study (1828.41±181.41 N).

The characteristics of Bio HPP may be the cause of the aforementioned outcomes. Bio HPP is a semi-crystalline polymer that exhibits great ductility and can tolerate a wide variety of plastic deformations under uniaxial tension and compression. The hardness of PEEK is enhanced by its crystalline concentration, which varies based on its thermal processing. The yield strength and tensile modulus increase with crystallization. [67]

Furthermore, Bio HPP behaves like a semi-crystalline polymer and undergoes hardening during deformation due to SIC (strain induced crystallization), whereby the material's density and hardness improve because of the polymer chains' alignment increasing the material's total crystalline content[68]. PEEK seems to be able to solidify at high stresses when exposed to static compressive loads, while the precise mechanism underlying this behavior is yet unclear.[69]

Furthermore, under ideal circumstances, the industrial production of CAD/CAM PEEK blanks demonstrates a reduced probability of porosity inside repairs, leading to higher mechanical characteristics. [63]

Additionally, because there is no sintering process or contraction [61], Bio HPP exhibited a superior marginal adaptability than zirconium. This improves fracture resistance under functional stress. [70]

The null hypothesis of this study was partially rejected because Bio HPP group recorded statistically significant higher fracture resistance mean values than Zr group.

Thermo-mechanical aging was performed for a limited number of 5000 thermocycles and 60000 loading cycles equivalent to only 6 months of clinical surface [27] which is considered a limitation of the present study, so more research is needed to simulate long-term oral performance for better evaluation of the durability of restorations. Also, the incorporation of artificial saliva is needed.

CONCLUSION

In the light of the limitations of the study, the conclusions were as follows:

1- When compared to zirconia, the Bio HPP FDPs framework showed non-significantly superior marginal adaption.
2- Thermo-mechanical aging had resulted in a significant reduction in the marginal adaptability of all tested bridge framework, but all recorded marginal gap mean values were within the clinically acceptable range. (120 mm).
3- Fracture resistance of Bio HPP FDPs framework was found to be significantly higherthan Zirconia.
4- 4-Bio HPP may be utilized even in the posterior region as a material for crowns and bridges.
5- It was found that there was non-significant an inverse correlation between fracture resistance and vertical marginal gap.

REFERENCES

1. Lopez-Suarez C, Rodriguez V, Pelaez J, Agustin-Panadero R, Suarez MJ.Comparative fracture behavior of monolithic and veneered zirconia posterior fixed dental prostheses. Dent Mater J. 2017;36:816–21.

2.Park JK, Lee WS, Kim HY, Kim WC, Kim JH. Accuracy evaluation of metal copings fabricated by computer-aided milling and direct metal laser sintering systems. J Adv Prosthodont. 2015;7:122-8.

3. Kim MJ, Choi YJ, Kim SK, Heo SJ, Koak JY. Marginal Accuracy and Internal Fit of 3-D Printing Laser-Sintered Co-Cr Alloy Copings. Materials. 2017;10:93-105.

4. Gonzalo E, Suarez MJ, Serrano B, Lozano JFL. A comparison of the marginal vertical discrepancies of zirconium and metal ceramic posterior fixed dental prostheses before and after cementation. J Prosthet Dent.2009;102:378–84.

5. Agustin-Panadero R, Roman-Rodriguez JL, Ferreiroa A, Sola-Ruez MF, Fons- Font A. Zirconia in fixed prosthesis. A literature review. J Clin Exp Dent 2014;6:e66-73.

6. Gupta AK, Gupta R, Gill S. Evaluation of the failure modes and load-bearing capacity of different surface-treated polyether ether ketone copings veneered with lithium di-

silicate compared to polyether ether ketone copings veneered with composite: an in vitro study. J Indian Prosthodont Soc. 2021;21:295-303.
7. Shahin A, Al-Wahadni A, Masri R. Zirconia-based restorations: literature review restoration. 2017; 3(2); 253-60.
8. He M, Zhang Z, Zheng D, Ding N, Liu Y. Effect of sandblasting on surface roughness of zirconia-based ceramics and shear bond strength of veneering porcelain. Dent Mater J. 2014; 33:778–85.
9. Alp G, Subaşı MG, Seghi RR, Johnston WM, Yilmaz B. Effect of shading technique and thickness on color stability and translucency of new generation translucent zirconia. J Dent. 2018; 73:19–23.
10. Al-Qaysi SH, Saleh ARM, El-Damanhoury HM. Fracture resistance and failure mode of layered translucent zirconia with different crown designs. Dent Mater J. 2022; 41:54–9.
11. Sarfaraz H, Rasheed MN, Shetty SK, Prabhu UM, Fernandes K, Mohandas S. Comparison of the bond strength of composite resin to zirconia and composite resin to polyether ether ketone: an in vitro study. J Pharm Bioallied Sci. 2020;12(Suppl 1):S504.
12. Zandparsa R, El Huni RM, Hirayama H, Johnson MI. Effect of different dental ceramic systems on the wear of human enamel: an in vitro study. Journal of Prosthetic Dentistry. 2016;115(2):230-7.
13. Cigu AT, Ciobanu C, Covalciuc E, Popovici M, Cârligeanu L, Ardeshir S. Research Of Biohpp System Behavior In The Oral Cavity. International Journal of Medical Dentistry. 2015;5(1):44-9.
14. Park J-Y, Bae S-Y, Lee J-J, Kim J-H, Kim H-Y, Kim W-C. Evaluation of the marginal and internal gaps of three different dental prostheses: comparison of the silicone replica technique and three-dimensional superimposition analysis. The journal of advanced prosthodontics. 2017;9(3):159-69.
15. Taufall S, Eichberger M, Schmidlin PR, StawarczykB.Fracture load and failure types of different veneered polyetheretherketone fixed dental prostheses. Clinical oral investigations. 2016;20(9):2493-500.
16- Stawarczyk B, Eichberger M, Uhrenbacher J, Wimmer T, Edelhoff D, Schmidlin PR. Three-unit reinforced polyetheretherketone composite FDPs: Influence of fabrication method on load-bearing capacity and failure types. Dental materials journal. 2015;34(1):7-12.
17. Mostafa H. Hafez , Tarek S. Morsy , Jihan F. Younis. Fracture Resistance of Three-Unit Implant Supported Fixed Partial Dentures Fabricated from Zirconia and PEEK (An In-Vitro Study). Journal of Fundamental and Clinical Research.2023; 3(1): 50-62
18. McLean JW, von Fraunhofer JA. The estimation of cement film thickness by an in vivo technique. Br Dent J. 1971;131:107–11.
19. Baran MC, Demirci F, Tuzlalı M. Comparison of marginal and internal adaptation of three-unit fixed dental prostheses made using CAD/CAM metal-free materials. Eur J Oral Sci. 2022; 130: e12901.
20. AyseGozde T, Metin S, Mubin U. Evaluation of adaptation of ceramic inlays using optical coherence tomography and replica technique. Braz Oral Res. 2018; 32: e005.
21. Contrepois M, Soenen A, Bartala M, Laviole O. Marginal adaptation of ceramic crowns: a systematic review. J Prosthet Dent. 2013; 110: 447–54.
22. Nawafleh NA, Mack F, Evans J, Mackay J, Hatamleh MM. Accuracy and reliability of methods to measure marginal adaptation of crowns and FDPs: a literature review. J Prosthodont. 2013; 22: 419–28.
23. Naert I, Van Der Donck A, Beckers L. Precision of fit and clinical evaluation of all-ceramic full restorations followed between 0.5 and 5 years. J Oral Rehabil. 2005;32:51–7.

24. Boening KW, Wolf BH, Schmidt AE, Kastner K, Walter MH. Clinical fit of ProceraAllCeram crowns. J Prosthet Dent. 2000; 84:419–24.
25. Amesti-Garaizabal A, Agustin-Panadero R, Verdejo-Sola B, Fons-Font A, Fernandez-Estevan L, Montiel-Company J, et al. Fracture Resistance of Partial Indirect Restorations Made with CAD/CAM Technology. A Systematic Review and Meta-Analysis. J Clin Med. 2019;8: 1932-42.
26-Nawafleh N, Hatamleh M, Elshiyab S, Mack F. Lithium Disilicate Restorations Fatigue Testing Parameters: A Systematic Review. J Prosthodont. 2016; 25(2):116-126.
27.Gale MS, Darvell BW. Thermal cycling procedures for laboratory testing of dental restorations. J Dent 1999; 27:89-99
28. Azari A, Nikzad S. The evolution of rapid prototyping in dentistry: a review. Rapid Prototyping Journal. 2009; 15(3): 216-25.
29. Kim W-Y, Chung I-S, Jeon B-W. Comparison of mechanical properties of all ceramic crown on zirconia blocks. Journal of Korean Acedemy of Dental Technology. 2015; 37(3): 107-13.
30.Sailer I, Fehér A, Filser F, Gauckler LJ, Luthy H, Hammerle CHF. Five-year clinical results of zirconia frameworks for posterior fixed partial dentures. International Journal of Prosthodontics. 2007; 20(4): 151-6.
31. Kheiralla LS, Younis JF. Peri-Implant Biomechanical Responses to Standard, Short-Wide, and Mini Implants Supporting Single Crowns Under Axial and Off-Axial Loading (an In Vitro Study). Journal of Oral Implantology. 2014; 40(1): 42-52.
32. Comlekoglu M, Dundar M, Özcan M, Gungor M, Gokce B, Artunc C. Influence of cervical finish line type on the marginal adaptation of zirconia ceramic crowns. Operative dentistry. 2009; 34(5): 586-92.
33.Stawarczyk B, Beuer F, Wimmer T, Jahn D, Sener B, Roos M, et al. Polyetheretherketone—a suitable material for fixed dental prostheses? Journal of Biomedical Materials Research Part B: Applied Biomaterials. 2013; 101(7): 1209-16.
36. Zaghloul HH, Younis JF. Marginal fit of implant-supported all-ceramic zirconia frameworks. Journal of Oral Implantology. 2013; 39(4): 417-24.
37. Palacios RP, Johnson GH, Phillips KM, Raigrodski AJ. Retention of zirconium oxide ceramic crowns with three types of cement. The Journal of prosthetic dentistry. 2006; 96(2): 104-14.
38. El-Naga A, El-fallal A, Ibraheim S, Alaraby HA. Fracture strength of two zirconia all-ceramic crown systems: influence of intaglio surface conditioning. Mansoura J Dent. 2014; 1(3): 67-71.
39. Beuer F, Aggstaller H, Richter J, Edelhoff D, Gernet W. Influence of preparation angle on marginal and internal fit of CAD/CAM-fabricated zirconia crown copings. Quintessence international. 2009: 40(3): 243-50.
40. Kim J-H, Cho B-H, Lee J-H, Kwon S-J, Yi Y-A, Shin Y, et al. Influence of preparation design on fit and ceramic thickness of CEREC 3 partial ceramic crowns after cementation. Acta Odontologica Scandinavica. 2015; 73(2): 107-13.
41 Good M-L, Mitchell CA, Pintado MR, Douglas WH. Quantification of all-ceramic crown margin surface profile from try-in to 1-week post-cementation. journal of dentistry. 2009; 37(1): 65-75.
42. Comlekoglu M, Dundar M, Özcan M, Gungor M, Gokce B, Artunc C. Influence of cervical finish line type on the marginal adaptation of zirconia ceramic crowns. Operative dentistry. 2009; 34(5): 586-92.
43. Alghazzawi TF, Liu PR, Essig ME. The Effect of Different Fabrication Steps on the Marginal Adaptation of Two Types of Glass-Infiltrated Ceramic Crown Copings Fabricated by CAD/CAM Technology. Journal of Prosthodontics. 2012; 21(3): 167-72.

44. El-Dessouky R, Salama M, Shakal M, Korsel A. Marginal adaptation of CAD/CAM zirconia-based crown during fabrication steps. Tanta Dental Journal. 2015; 12(2): 81-8.
45. Preis V, Behr M, Hahnel S, Rosentritt M. Influence of cementation
on in vitro performance, marginal adaptationand fracture resistance of CAD/CAM-fabricated ZLS molar crowns. Dental Materials. 2015; 31(11):1363-9.
46. Stappert C, Dai M, Chitmongkolsuk S, Gerds T, Strub J. Marginal adaptation of three-unit fixed partial dentures constructed from pressed ceramic systems. British Dental Journal. 2004; 196(12):766-70.
47. Suárez MJ, Lozano JF, Salido MP, Martínez F. Marginal fit of titanium metal-ceramic crowns. International Journal of Prosthodontics. 2005; 18(5): 390-1
48.Groten M, Axmann D, Pröbster L, Weber H. Determination of the minimum number of marginal gap measurements required for practical in vitro testing. The Journal of prosthetic dentistry. 2000; 83(1): 40-9.
49. McLean J, Von Fraunhofer J. The estimation of cement film thickness by an in vivo technique. British dental journal. 1971;131(3):107-11
50-Att W, Komine F, Gerds T, Strub JR. Marginal adaptation of three different zirconium dioxide three-unit fixed dental prostheses. The Journal of prosthetic dentistry. 2009;101(4):239-47.
51. Dittmer MP, Borchers L, Stiesch M, Kohorst P. Stresses and distortions within zirconia-fixed dental prostheses due to the veneering process. Acta biomaterialia. 2009; 5(8):3231-9.
52. KARATASLI Ö, Kursoglu P, Nuray Ç, Kazazoglu E. Comparison
of the marginal fit of different coping materials and designs produced by computer aided manufacturing systems. Dental materials journal. 2011;30(1):97-102.
53. Taha D, Spintzyk S, Sabet A, Wahsh M, Salah T. Assessment of marginal adaptation and fracture resistance of endocrown restorations utilizing different machinable blocks subjected to thermomechanical aging. J EsthetRestor Dent. 2018; 30(4): 319-328.
54.Rosentritt M, Sikora M, Behr M, Handel G. In vitro fracture resistance and marginal adaptation of metallic and tooth-coloured post systems. J Oral Rehabil. 2004 Jul; 31(7):675-81.
55. Morresi AL, D'Amario M, Monaco A, Rengo C, Grassi FR, Capogreco M. Effects of critical thermal cycling on the flexural strength of resin composites. J Oral Sci. 2015; 57(2):137-143.
56. Krejci I, Mueller E, Lutz F. Effects of thermocycling and occlusal force on adhesive composite crowns. J Dent Res. 1994; 73(6): 1228-1232.
57. Beschnidt SM, Strub JR. Evaluation of the marginal accuracy of different all-ceramic crown systems after simulation in the artificial mouth. J Oral Rehabil. 1999; 26(7): 582-593.
58.Donmez MB, Okutan Y. Marginal gap and fracture resistance of implant-supported 3D-printed definitive composite crowns: An in vitro study. J Dent. 2022; 124:104216.
59.McLean JW, von Fraunhofer JA. The estimation of cement film thickness by an in vivo technique. Br Dent J. 1971; 131(3): 107-111.
60. Zimmermann M, Ender A, Egli G, Özcan M, Mehl A. Fracture load of CAD/CAM-fabricated and 3D-printed composite crowns as a function of material thickness. Clin Oral Investig. 2019; 23(6): 2777-2784.
61.Park J-Y, Bae S-Y, Lee J-J, Kim J-H, Kim H-Y, Kim W-C. Evaluation of the marginal and internal gaps of three different dental prostheses: comparison of the silicone replica technique and three-dimensional superimposition analysis. The journal of advanced prosthodontics. 2017; 9(3): 159-69.

62. Kohorst P, Herzog TJ, Borchers L, Stiesch-Scholz M. Load-bearing capacity of all-ceramic posterior four-unit fixed partial dentures with different zirconia frameworks. European journal of oral sciences. 2007; 115(2): 161-6.
63. Stawarczyk B, Eichberger M, Uhrenbacher J, Wimmer T, Edelhoff D, Schmidlin PR. Three-unit reinforced polyetheretherketone composite FDPs: Influence of fabrication method on load-bearing capacity and failure types. Dental materials journal. 2015; 34(1): 7-12.
63. Raigrodski AJ. Contemporary materials and technologies for all-ceramic fixed partial dentures: a review of the literature. The Journal of prosthetic dentistry. 2004; 92(6): 557-62.
64. Larsson C, Holm L, Lövgren N, Kokubo Y, Vult von Steyern P. Fracture strength of four-unit Y-TZP FPD cores designed with varying connector diameter. An in-vitro study. Journal of oral rehabilitation. 2007; 34(9): 702-9.
65. Pfeiffer P, Grube L. Effect of pontic height on the fracture strength of reinforced interim fixed partial dentures. Dental materials. 2006; 22(12): 1093-7.
66. Taufall S, Eichberger M, Schmidlin PR, Stawarczyk B. Fracture load and failure types of different veneered polyetheretherketone fixed dental prostheses. Clinical oral investigations. 2016; 20(9): 2493-500.
67. Chivers R, Moore D. The effect of molecular weight and crystallinity on the mechanical properties of injection moulded poly (aryl-ether-ether-ketone) resin. Polymer. 1994; 35(1): 110-6.
68. Swallowe G, Fernandez J, Hamdan S. Crystallinity increases in semi crystalline polymers during high rate testing. Le Journal de Physique IV. 1997; 7(C3): C3-453-C3-8.
69. Rae P, Brown E, Orler E. The mechanical properties of poly (ether-ether-ketone)(PEEK) with emphasis on thelarge compressive strain response. Polymer. 2007; 48(2): 598-615
70. Kohorst P, Brinkmann H, Li J, Borchers L, Stiesch M. Marginal accuracy of four-unit zirconia fixed dental prostheses fabricated using different computer-aided design/ computer-aided manufacturing systems. European journal of oral sciences. 2009; 117(3): 319-25.
71. Beuer F, Steff B, Naumann M, Sorensen JA. Load-bearing capacity of all-ceramic three-unit fixed partial dentures with different computer-aided design (CAD)/computeraided manufacturing (CAM) fabricated framework materials. European journal of oral sciences. 2008; 116(4): 381-6.
72. Stawarczyk B, Ender A, Trottmann A, Özcan M, Fischer J, Hämmerle CH. Load-bearing capacity of CAD/CAM milled polymeric three-unit fixed dental prostheses: effect of aging regimens. Clinical oral investigations. 2012;16(6):1669-77.

Tables

Table-1. Materials used in the study

Material	**Product name**	**Manufacturer**	**Composition**
PEEK	bre.CAM Bio HPP	Bredent GmbH &Co.KGt,Senden,Germany	Difluoro benzophenone (1.31 g of 4.4), hydroquinone (0.66 g), and K_2CO_3(1.24 g) has to be dissolved in a mixture containing 15ml of solvent and 35ml of toluene
Zirconia	Sagemax	Prettau,	Zirconium oxide ZrO2≥ 89%

	NEXXZR.T	Zirconzahn, Italy	Yttrium oxide + Y2O3 4–6% Hafnium oxide HfO2≤ 5% Aluminium oxide AI2O3 < 1%
Surface treatment	Particle abrasive	Renfert,GmbH,Germany	Aluminium oxide particles (110 µm)for airborne abrasion and 2.5 bar pressure at a working distance of 10 mm for 15 second.

Table-2. Sintering schedule of zirconia FDPs frameworks

Temperature	**Programming Rate**	**Holding Time**
Room Temp. to 1550ºC	10ºC/min	____
1550ºC	Constant	2 hr
1550ºC to room Temp.	-10ºC/min	____

Table-3. The mean and standard deviation of the marginal gap (µm) prior to cementation and after thermomechanical aging

	Variables	**Molar**		**premolar**	
		Before cementation	**After thermomechanical aging**	**Before cementation**	**After Thermomechanicalaging**
Material Groups	**Bio HPP**	**22.16±1.8**	**24.21±3.1**	**22.19±1.7**	**27.53±2.9**
	Zr	**20.61±2.1**	**28.25±2.7**	**22.25±2.5**	**35.19±3.1**

Table-4. Comparing the overall marginal gap (mean values +/- SDs) depending on the type of material group.

Variables		**Mean± SD**	**Tukey's rank**	**Statistics (P value)**
Material groups	**Bio HPP**	**23.27 ±2.13**	**A**	**0.09325 ns**
	Zr	**24.31±3.1**	**A**	

ns; non-significant (p>0.05)

Table-5. Comparison between marginal gap (Mean values± SDs) before cementation andafter Thermo-mechanical aging

Variables		**Mean± SD**	**Tukey's rank**	**Statistics (P value)**
Thermo-mechanical aging	**before cementation**	**22.41 ±2.1**	**A**	**<0.0001***
	AfterThermo-mechanical aging	**30.88±3.3**	**B**	

; significant (p < 0.05)

Table-6. Measurements of fracture resistance (N) results (Mean values ±SD) for both material groups.

Variables	**Mean ± SD**	**Range**		**Confidence intervals**		**t-test**
		Min.	**Max.**	**Lower**	**Upper**	**P value**

Material groups	**Bio HPP**	1828.41 ± 181.41	1637.2	2126.4	1689	1963.6	0.003*
	Zr	1441.23 ± 165.3	1326.3	1737.5	1317.8	1569.3	

; significant (p < 0.05)

Figures legend

Fig. 1. Steel abutments in an acrylic block

Fig.2. Epoxy resin model

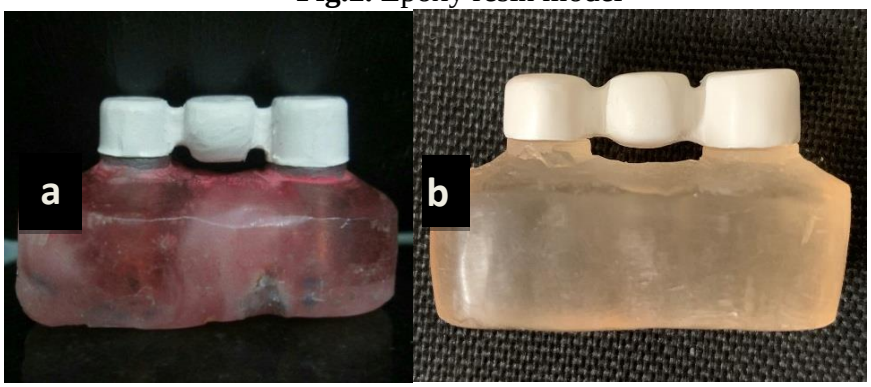

Fig.3. Framework checking on : (a) steel model and (b) epoxy resin model

Fig.4. Bio HPP framework under load of Universal testing machine

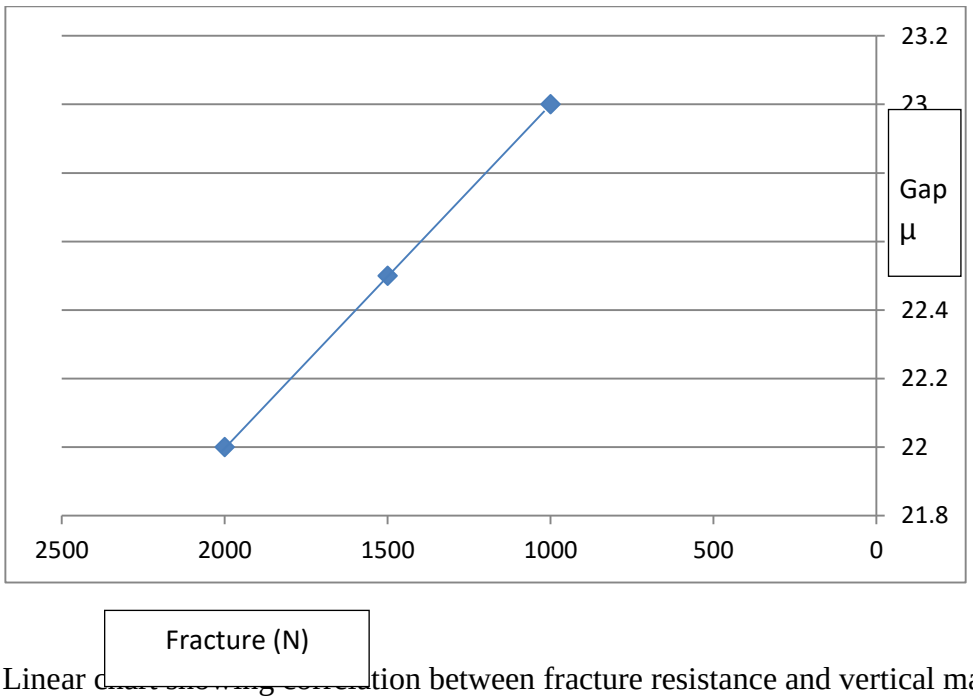

Fig.5. Linear chart showing correlation between fracture resistance and vertical marginal gap