

THE EFFECT OF THERMOCYCLING AND MECHANICAL LOADING ON THE FRACTURE RESISTANCE OF ALL-CERAMIC AND HIGH PERFORMANCE POLYMERS FIXED PARTIAL DENTURES

Mohamed Mohey Eldin Mohamed*

ABSTRACT

Objectives: The purpose of this study was to evaluate the fracture resistance of the heat pressed ceramic (IPS e-max press) and the high performance polymers BioHPP anterior FPDs.

Materials and Methods A total number of 96 sound freshly extracted maxillary central incisors and canines were used in this study. The teeth were used to prepare a total of 48 three units FPD tested specimens. Specimens were used to test the fracture resistance of the materials under investigation without and after thermocycling and mechanical loading.

Results : The fracture load values of the IPS e-max specimens without TCML were ranged from 271.7 N to 206.01 N. While, after TCML the values were ranged from 171.6 N to 191.2 N. The fracture load values of the BioHPP specimens without TCML were ranged from 789.7 N to 665.1 N. While, after TCML the values were between 789.7 N and 615.4 N.

Conclusion: IPS e-max press FPDs had a lower fracture resistance than the BioHPP FPDs before and after TCML. The TCML decreased the resistance to fracture of both materials.

KEYWORDS: IPS e-max, high performance polymers, fracture resistance

INTRODUCTION

The Improvement in dental technology and the increased esthetic demand made metal-free restorations more popular.⁽¹⁾

Fixed partial dentures (FPDs) made of metal frameworks with porcelain facing provide high fracture strength and long-term clinical experiences, but still have many drawbacks as metal display, and discoloration of the gingival margins, as a result of metal corrosion.⁽²⁾

In order to minimize these drawbacks all ceramic restorations are now the first choice for highly aesthetic results.⁽³⁾

Several all ceramic systems are developed to achieve the most challenging requirements in restorative dentistry, as the ease of fabrication, good esthetics with adequate strength and fracture toughness.

* Lecturer of Fixed Prosthodontics, Delta University For Science and Technology

IPS e.max Press and IPS e.max CAD are based on lithium disilicates and are available for the heat-pressed and CAD-CAM technique respectively. IPS e.max Press are in the form of ingots gaining their strength by means of fine dispersed crystals in glassy matrix that does not need an additional crystallization procedure.^(4,5)

The crystals incorporated in the material create a barrier against the formation of the microcracks, improve the flexural strength, and the fracture resistance.⁽⁵⁾

IPS e-max press exhibit the strength necessary to fabricate multiunits anterior FPDs up to second premolar. It offers good esthetics, high translucency, excellent marginal integrity and etchability that promote adhesion to the underlying tooth structure.⁽⁶⁾

Among factors related to the failure in all-ceramic restorations, are defects, mechanical residual stresses, thermal residual stresses and contact cracks, however defects as pores and small cracks may result in stress concentration and become the site of subcritical crack growth.^(7,8)

Crack growth and propagation within ceramic materials with cyclic loading leads to failure under cyclic stress more than under a static load. As a result defects inside or on the surface of ceramics are considered critical in initiating catastrophic fracture and chipping.⁽⁹⁾

Polyetherketone (PEEK), is the main thermoplastic polymer that has a tensile strength similar to that of bone, enamel and dentin, making it an optimal material for prosthetic restorations.

High Performance Polymer (BioHPP) is a partly crystalline poly ether ether ketone (PEEK) that is strengthened using 20% ceramic filler.

This homogeneity distributed ceramic of about 0.3-0.5 microns allow better polishing of the restorations and result in a lack of plaque retention and color stability.⁽¹⁰⁾

Due to the improvement in mechanical properties of PEEK material it is used not only as interim abutments, but also as implant-supported bars, dental implants and fixed partial dentures (FPDs). However, it is not used as a monolithic restoration but as a framework and additional veneering resin is required as the material has low translucency⁽¹⁰⁾

Previous studies showed lower fracture load as a results of the internal tensile stresses within the FPDs and thermal stress, as well, which affect the bond and flexural strength of the framework together with the veneering resin.

Adhesive properties also which are important for the stability of restorations, are influenced by the surface treatment and luting cement.⁽¹⁰⁾

Although several studies and clinical case reports are published, further investigations and clinical evaluations are required to prove the clinical reliability of these systems.^(11,12)

The purpose of this study is to evaluate the fracture resistance of the heat pressed ceramic (IPS e-max press) and the BioHPP anterior FPDs.

The hypothesis was that thermal cycling and mechanical loading does not affected the fracture resistance of the heat pressed ceramic and the High Performance Polymer FPDs.

MATERIALS AND METHODS

Materials:

One pressable ceramic material IPS e-max and one high performance polymer composite BioHPP blank together with a Luting resin cements, bonding resin and silan coupling agent were used in this study.

Materials used in the study

Materials & trade name	Materials composition	Manufactures
Heat pressed ceramic (IPS e max press ingot)	- Glass matrix Lithium disilicat needle-like crystals of 3 to 6 μm in length. (Approx. 70%)	Ivoclar vivadent AG Benderestrasse 2 FL-9494 Schaan Liechtenstein
IPS e.max Ceram (fluorapatite veneering ceramic)	- Glass ceramic Nanoscale fluorapatite crystals, less than 300 nm in length and approx. 100 nm in diameter.	Ivoclar vivadent AG Benderestrasse 2 FL-9494 Schaan Liechtenstein
High-performance polymers	Polyetheretherketone Ceramic filler (20 wt%) of 0.3 to 0.5 μm grain size.	Bredent group .GmbH & Co. KG Weissenhorner Str. 2 · 89250 Senden· Germany
Visio.lign Veneering composite	PMMA resin Nano Ceramic Filler (Approx. 50%) No ground glass filler	Bredent group .GmbH & Co. KG Weissenhorner Str. 2 · 89250 Senden· Germany

Methods

A total number of 96 sounds freshly extracted maxillary central incisors and canines were used in this study. The teeth were used to prepare a total of 48 three units FPD tested specimens. Specimens were used to test the fracture resistance of the materials under investigation.

The teeth were cleaned under running water to remove the soft tissues, blood, and debris, then were polished with pumice and stored in physiological saline until used.

Preparation of the Tested Specimens:

One central incisor and a canine were inserted in pairs in a prefabricated oblong copper cylinder (**20mm lengthx30mmx20mm inside diameter**) using autopolymerized acrylic resin. The surface of the cylinder and the resin were 3 mm apical to the finish line. The distance between the prepared teeth were 6.5mm that is equal to the mesiodistal diameter of the maxillary lateral incisor.⁽¹³⁾ After complete polymerization of the acrylic resin, the teeth and the acrylic blocks were removed from the copper mold.

The teeth in the acrylic block were prepared with the following standardized preparation criteria, 1.2 mm shoulder finish line labially and 1mm shoulder

finish line lingually placed 0.5 mm incisal to the CEJ, and 2mm incisal reduction.

An Isoparallelometers (Cruise 440)^(*) **Silfradent via G Divittorio 35/37-470185 Sofia (FORLI)-Italy**) milling machine, was used for tooth preparation using a taper diamond stone (846-012).

After teeth preparation, an elastomeric addition silicon impression was made for the prepared teeth to form the working model. IPS e-max press FPDs and BioHPP FPDs were fabricated and cemented according to the manufacture's instructions. Each acrylic block with the fabricated FPD was returned back to the copper cylinder.

The cylinder was fixed to a horizontally flat metal plate with an angle of 45⁽¹³⁾ corresponding to the average inter-incisal angle between the maxillary and mandibular incisors.⁽¹⁴⁾

Fabrication of the e-max press fixed partial denture

IPS e-max FPDs framework were waxed-up, invested and burned out and pressed from ingot IPS e-max (Ivoclar vivadent AG Benderestrasse 2 FL-9494 Schaan Liechtenstein). Each connector was 4 mm width and 4 mm height. The thickness of the framework was 0.8 mm. The framework was veneered with IPS e-max ceram nano-flouroapatite veneering ceramic, and glazed.⁽¹⁵⁾

Fabrication of the BioHPP fixed partial denture

Each prepared teeth FPD were scanned with a CEREC AC omnicam (Sirona Dental Systems, Bensheim, Germany), and FPD framework were designed (CEREC Software Sirona Dental Systems), and milled (CEREC inLab MC X5 milling unit) from an BioHPP blank (Bredent group GmbH & Co. KG Weissenhorner Str. 2 · 89250 Senden · Germany). Each connector was 4 mm width and 4 mm height with a minimal thickness of 0.8 mm.

The surface of BioHPP FPDs framework were sandblasted with 110 µm aluminium oxide at a pressure of 2 to 3 bars surface, conditioned with a thin film of visio.link using a microbrush and polymerized for 90s. Combo. Lign opaque was applied to the conditioned framework and polymerized for 180 s.⁽¹⁶⁾

Visio-lign veneer was milled from Visio-lign blank. The inner surface of the veneers were conditioned from inside with visio.link (Bredent) and polymerized for 90 s. Frameworks were put on the prepared teeth and the veneers were filled with combo.lign before pressing them on the prepared frameworks excess material was removed and polymerized for 360 s using a light-curing unit (bre. lux power unite).⁽¹⁶⁾

Cementation of the restorations:

Pre-treatment of the restoration:

The internal aspects of the IPS E-max press FPD was etched with IPS Ceramic Etching Gel (5% hydrofluoric acid) for 20 seconds, thoroughly rinsed with water and dried with air then silanated with Monobond-S for 60 seconds and dried with air.

The internal aspect of BioHPP FPDs were sandblasted with 110 µm aluminum oxide powders at 2 bar pressure.

Pre-treatment of the preparation and resin cementation:

Phosphoric acid gel (37%) was applied to the prepared tooth with a disposable syringe tip, for 20 seconds.

All etchant gel was removed with a vigorous water spray for at least 5 seconds. Excess water was removed with high-volume evacuation tip directly over the preparation surface for 1-2 seconds, subsequently, the adhesive, Excite DSC, was applied and gently agitated for at least 10 seconds.

Variolink II was mixed in a 1:1 ratio on a mixing pad for 10 s with a spatula, applied on the inner surface of the IPS e max FPD and placed in situ with slight pressure, cured for 3 seconds then excess cement was removed with scaler and light cured for 40 seconds.

The HPP FPDs were cemented by Rely X Unicem self-adhesive resin cement cured for 3 seconds then excess cement was removed with scaler and light cured for 40 seconds as previously mentioned.

Grouping of the Specimens:

The fabricated specimens were divided according to the material used into two main groups I and II of 24 three-unit FPD specimens each.

The group (I) was made of 24 pressable ceramic, IPS e-max press. FPD specimens.

The group (I) was divided into 2 subgroups Ia , and Ib of twelve specimens each

Subgroup Ia evaluated the fracture resistance without thermocycling and mechanical loading.

The Subgroup Ib evaluated the fracture resistance after thermocycling and mechanical loading.

The group (II) was consisted of 24 BioHPP FPD specimens.

The group (II) was divided into subgroup IIa, and IIb of twelve specimens each

The Subgroup IIa evaluated the fracture resistance without thermocycling and mechanical loading.

The Subgroup IIb evaluated the fracture resistance after thermocycling and mechanical loading

Fracture Resistance Test:

The FPD specimens of Ia and IIa were loaded using a Universal mechanical testing machine ((MAXI-TORQ) Comten-industries, inc, street Petersburg Florida USA) without thermocycling and mechanical loading (TCML).

Specimens were attached to the lower plate of the testing machine, and a round end ball eight mm in diameter was attached to the upper plate of the machine.

The ball was allowed to move vertically in a compressive mode at a crosshead speed of 0.5 mm /min at 135 degrees to the long axis of the teeth corresponding to the average inter-incisal angle between the maxillary and mandibular incisors until fracture occurs.⁽¹⁴⁾

The FPD specimens of subgroup Ib and IIB were subjected to 6000 thermal cycle at 5C° x55C in water bath with a dwell time of 2 minutes, then were subjected to load cycle of 1.2x10⁶x 50N, 1.2Hz the parameters represented 5 years of oral stress simulation⁽¹⁶⁾. After TCML the FPD specimens were loaded until fracture occurs.

The fracture loads were recorded in Newton (N)

The data from the specimens without and after TCML were tabulated, and statistically analyzed.

The fracture specimens were examined by the Stereomicroscope to evaluate the nature of the failure.

RESULTS

Statistical analysis

The data was collected and entered into the personal computer. Statistical analysis was done using Statistical Package for Social Sciences (SPSS/ version 20) software.

Mean and standard deviation was calculated for different measurements. Wilcoxon matched pairs signed ranks test (W tests) a nonparametric significant test used to compare paired, ordinal

data. Mann Whitney test was used for comparison between unpaired signed ranks test (U test). The 5% was chosen as the cut off level of significance.

Fracture resistance:

The fracture resistance value of e-max group without TCML and after TCML, are represented in Table (1).

The mean, standard deviation, minimum, and maximum of the fracture loads for all groups were calculated.

The fracture load values of the IPS e-max specimens without TCML were ranged from 271.7 N to 206.01N. While, after TCML the values were ranged from 171.6 N and 191.2 N.

It was observed from the collected data that the resistance to fracture of IPS e-max specimens was significantly decreased after TCML.

Statistical analysis using the U test showed significant difference between the mean fracture load of the IPS e-max specimens without TCML (236.094± 33.214) and after (178.215±11.328) TCML (P=0.05). Table (1)

Table (2) reveal that the fracture load values of the BioHPP specimens without TCML were ranged from 789.7 N to 665.1 N. While, after TCML the values were between 789.7 N and 615.4 N.

It was observed from the collected data that the fracture resistance was decreased significantly after TCML.

Statistical analysis using the U test showed significant difference between the mean fracture load of the BioHPP without (715.803± 65.458) and after (604.04±213.35) TCML (P=0.0315).

Comparison between the two groups regarding the difference in fracture resistance without and after TCML , showed that the mean of the fracture loads of the BioHPP prior to cyclic loading was (715.803 N), which was significantly higher than that of the IPS e-max specimens (236.094 N). Table (3)

Meanwhile, the mean fracture loads of BioHPP specimens were (604.04 N) after the TCML, which was significantly higher than that of the IPS e-max specimens (178.215 N). Table (3) Figure (1)

Stereomicroscope examination results

Examination of the fracture mode of IPS e-max press FPD specimens by Stereomicroscope revealed cohesive fracture of five FPDs specimens within the connector area and mixed fracture in one specimen.

The mixed fracture was cohesive fracture within the connector of one side of the pontic and splitting fracture of the retainer with adhesive failure of the other side. Table (4), Figure (2,3)

Examination of the fracture mode of BioHPP FPD specimens by Stereomicroscope revealed cohesive fracture of all specimens. Five FPDs specimens showed chipping of the veneered BioHPP and one specimen showed delamination. Table (5), Figure (4)

TABLE (1) Fracture resistance of e-max group without and after TCML. (U test)

IPS e-max press	Without TCML	After TCML
Min.	206.01	171.675
Max.	271.737	191.295
Mean	236.094	178.215
±S.D.	33.214	11.328
P	0.05*	

TABLE (2) Fracture resistance of BioHPP group without and after TCML. (U test)

BioHPP	Without TCML	After TCML
Min.	665.118	615.46
Max.	789.705	789.71
Mean	715.803	604.04
±S.D.	65.458	213.35
P	0.0315*	

TABLE (3) Comparison between the fracture resistance of IPS e-max group and BioHPP group without and after TCML.

IPS e- max press	Without TCML	After TCML
Min.	206.01	171.675
Max.	271.737	191.295
Mean	236.094	178.215
±S.D.	33.214	11.328
BioHPP	Without treatment	After treatment
Min.	665.118	615.46
Max.	789.705	789.71
Mean	715.803	604.04
±S.D.	65.458	213.35
P	0.0032*	0.0015*

TABLE (4) Mode of fracture of IPS e-max press FPDs.

Tested Material With and After TCML	IPS e-max	
	No.	%
Fracture connector	5	83.3%
Complete splitting	1	16.7%
Total	6	100%

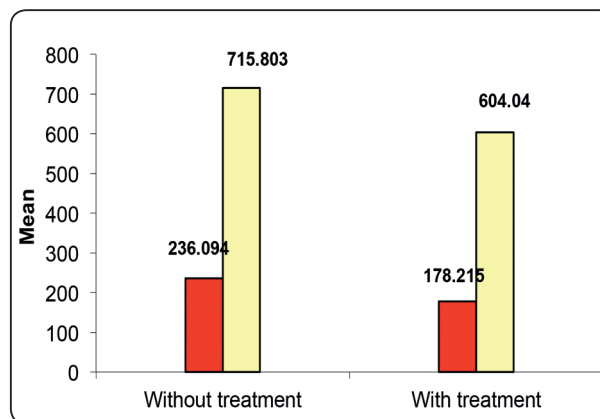


Fig. (1) Comparison between the fracture resistance of IPS e-max group and BioHPP group without and with TCML.

TABLE (5) Mode of fracture of BioHPP

Tested Material With and After TCML	BioHPP	
	No.	%
Chipping of veneered composite	5	83.3%
Delaminated fracture	1	16.7%
Total	6	100%

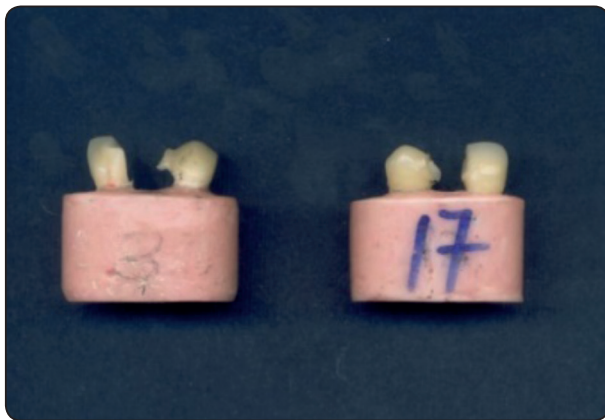


Fig. (2) Mixed fracture of IPS e-max specimen at the connector area and within the retainer.

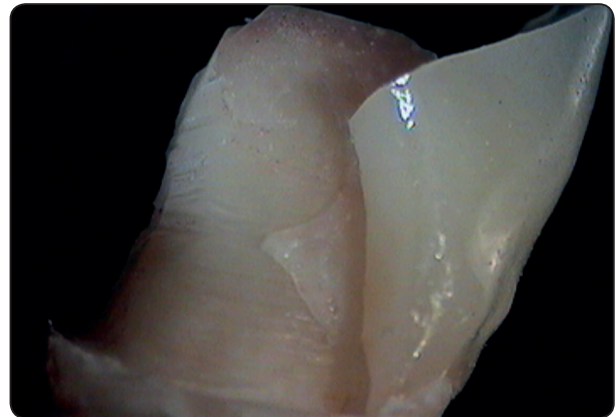


Fig. (3) Complete splitting fracture of IPS e-max specimen



Fig. (4) Delaminated Fracture of the BioHPP specimen.

DISCUSSION

All-ceramic dental restorations exhibit enhanced esthetics and biocompatibility as compared to traditional metal-based prosthesis.

However, long-term fatigue and subcritical crack growth in the presence of water and cyclic loading can decrease the strength of ceramic components over time.⁽¹⁷⁾

Fracture strength tests, within limits, provide some idea of the load-bearing capacity of restorations in simulated clinical situations.

It is important to determine the forces that may induce fracture of such restorations and then to suggest the material structure that will provide greatest fracture strength.^(17,18)

Fabrication of three-unit anterior FPDs using e- max press system are recommended because of the high fracture toughness of core ceramic however, in this situation when the core layer is coated with the low-strength glass veneer, the resulting ceramic composite has a significantly lower strength compared to the core ceramic. Distribution of stresses is affected by the elastic modulus differences between the glass veneer and ceramic core.^(19,20)

Other factors may explain the decrease in the strength of the veneering porcelain compared to the core ceramic as the complex stress state within the porcelain because of the thermal expansion mismatch with the underlying core material and also the possible tempering stresses that may develop as a consequence of the cooling rate when

the specimen is removed from the sintering furnace during the final cooling. Also defects present in the veneering porcelain are more complex and less homogeneously distributed than in the core ceramic.⁽²¹⁻²³⁾

In this study, natural teeth were prepared according to clinically established preparation criteria for metal free restoration, and the luting procedures also followed the clinical protocols to ensure a close simulation of clinically relevant conditions.

A large-diameter steel ball 8mm in diameter was used to develop as clinically relevant contacts as possible to allow more distribution of stresses through the pontic and the retainers.

This study evaluated the fracture resistance of the pressable ceramic e-max press and HPP FPD without and after TCML. The result of this study showed a significant decrease in the fracture resistance of IPS e-max press FPDs with TCML.

Evaluation of the results of this study indicated that the fracture resistance of both materials was decreased after TCML.

This finding was in agreement with the finding of Dueummond J et al who observed a decrease of 15% in the flexure strength of the lithuim disilicate material when tested in water. Their explanation was the presence of moisture at the crack tip aids in crack growth which in turn decrease the strength.⁽²⁴⁾

Another study by Studart A et al explained the decrease of the fracture strength of all ceramic materials as a result of cracks propagation, under wet cyclic loading conditions that lead to the fracture of the veneer layer before the rupture of the core framework material.⁽²⁵⁾

Several research results have shown that chipping mostly initiates from defects at the core–veneer interface, chipping and fracture are also observed to initiate from the defects in the connector areas which was in agreement with the result of this study.⁽²⁶⁻²⁸⁾

Regarding the fracture mode of the FPD, the current study found that the majority of IPS e-max press specimens were fractured within the connector area (66.66%) and other specimens within the retainer (33.33%) with a mean fracture load of 236.094 N without TCML and 178.215 N after TCML.

This finding is correlated to the results of Oh M et al⁽²⁹⁾ who reported that the radius of connector curvature at the gingival embrasure strongly affected the fracture resistance of all-ceramic FPDs.

Previous studies showed better stress distribution in broadly curved connectors than in more sharply curved connector geometries.

Another study performed by Plengsombut K et al⁽³⁰⁾ reported that, failure in IPS e-max FPDs initiated from the gingival surface of one connector and propagated toward the pontic. This fracture pattern was explained by the physical property of ceramic materials that enables them to withstand compressive forces better than tensile forces.

In this study, other mode of fracture was detected during examination of the fracture area of IPS e-max press specimens. A complete splitting within the retainer was observed.

This finding indicated that the failure passed completely through the core and the veneered ceramic.

Taskonak B et al.⁽³¹⁾ mentioned that the stress corrosion of glass-containing ceramics, acts as the dominant mechanism in reducing their fracture resistance and clinical lifetime, which are generally higher in glass-rich, feldspathic porcelains and silica-based glass-ceramics.

Pallis K et al.⁽³²⁾ reported that the core–veneer interface was the origin of failure of pressable ceramics, while in the current study a complete splitting of both layers was observed.

This observation might be due to the strong mechanical and chemical core–veneer bond strength

beside the similarity of coefficient of thermal expansion that allows cracks to cross this interface.

If a weak bond existed, cracks would travel in the interface separating the core from the veneered ceramic. Aboushelib M et al⁽³³⁾

Interfaces played an important role in the mechanical performance of bi-material ceramic composites such as core-veneered all-ceramic dental restorations. When interface toughness exceeded the flexural stresses in the tensile surface, a sharp crack propagates and penetrates through the core-veneer interface, behaving like a homogeneous material.

Alternatively, when flexural stresses at failure exceed the interface toughness the crack may deflect and extend along the interface between core and veneer with delamination Thompson G.⁽³⁴⁾

The exposure of polymeric restorative materials to water molecules push the polymeric chains apart causing expansion. These water molecules act as plasticizers leading to decreases in the resin strength.⁽³⁵⁾

Renan B et al.⁽³⁵⁾ concluded that resin composite materials are more fatigue resistant than glass-rich ceramics used in cyclic flexural loading.

The result of this study showed a significant decrease in fracture resistance of BioHPP FPDs after TCML. The mean fracture load was (715.803±65.458) without TCML and (604.04±213.35) after TCML (P=0.0315).

Concerning the mode of fracture of the BioHPP FPDs the results of the current study indicated that all FPDs specimens showed cohesive fracture either in the composite veneer or delamination between the veneer composite and the BioHPP core without complete fracture of the core.

This result was in agreement with Stawarczyk B et al⁽³⁶⁾ who reported that the FDPs fabricated from CAD/CAM blanks display a plastic deformation without complete fracture.

Skirbutis G. et al⁽³⁷⁾ found that PEEK properties are similar to dentin, enamel, and superior over metal alloys and ceramic restorations which is coinciding with the result of this study.

The mechanical properties of PEEK may represent a promising biomaterial, and are able not only to replace conventional polymers, but also metals, alloys and ceramics. Schwitalla A et al⁽³⁸⁾ evaluated the mechanical properties of PEEK compounds and concluded that the tested PEEK compounds exhibit very high flexural strength values. The result of this study also showed that there was a cohesive fracture in the composite veneer or delamination without complete fracture of the core which exhibit high strength.

The result of this study concluded that the IPS e-max press FPDs had a lower fracture resistance than the BioHPP FPDs. This was in accordance to the result of Stawarczyk B⁽³⁹⁾ who found that CAD-CAM milled PEEK fixed FPD had higher fracture resistance than lithium disilicate ceramic or zirconium FPD.

PEEK materials have an excellent mechanical properties and are suitable as FPDs at high masticatory forces in the posterior region however, a high frequency of cohesive failures occur because of uneven distribution of stresses at the bonding interface during the loading process.⁽⁴⁰⁾

CONCLUSION

Within the limitation of this study it was concluded that IPS e-max press FPDs had a lower fracture resistance than the BIOHPP FPDs before and after TCML. The TCML decreased the fracture resistance of both materials.

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