Original Article

Influence of Radius of Curvature at Gingival Embrasure in Connector Area on Stress Distribution of Three-Unit Posterior Full-Contour Monolithic Zirconia Fixed Partial Denture on Various Amounts of Load Application: A Finite Element Study

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Abstract

Objectives: To test the hypothesis that radius of curvature at gingival embrasure in connector area significantly affects the fracture resistance of full-contour monolithic zirconia three-unit posterior Fixed partial denture (FPD) on various amounts of load application. **Materials and Methods:** In this study, two types of three-dimensional finite element models of a three-unit posterior full-contour monolithic zirconia FPD with two gingival embrasure radii (r_{GE} I, 0.45 mm and r_{GE} II, 0.25 mm) were constructed. The components modeled through finite element modeling were subjected to 400, 600, and 800 N vertical loads at the central fossa of the pontic, and further analysis was carried out. **Results:** All the results were displayed by post-processor finite element analysis software (ANSYS). The study revealed that with increase in the amount of load application as well as decrease in the gingival embrasure radii, stress concentration values were increasing gradually for both the full-contour monolithic zirconia FPD. **Conclusion:** The fracture resistance of the zirconia posterior FPD was significantly affected by the gingival embrasure radii and the mode of load application. When there is a clinical situation of heavier occlusal forces, the fracture resistance can be increased by designing greater gingival embrasure radii in the connector region.

Keywords: *Finite element analysis, full-contour monolithic zirconia FPD, gingival embrasure radii*

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Introduction

Modern restorative spectrum for fixed ceramic prosthesis has been revolutionized by the introduction of full-contour monolithic zirconia systems such as Lava, Cercon, Zenostar, and Nobel Biocare.[1-4] The absence of veneering materials makes these new zirconia blocks extremely strong and durable.[5,6] Clinical data, however, suggest a relatively high failure rate of Fixed partial denture (FPD) most often around connector areas between retainers and pontics. The stress distribution in a fixed prosthesis can be affected by a change in contour of the prosthesis

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components; this effect may be more significant at locations with an abrupt change of contour. Through modification of the connector design in regions where maximum stress occurs, the stress pattern may be altered to improve the fracture resistance of three-unit FPDs. Previous studies showed better stress distribution in broadly curved connectors than in more sharply curved connector geometries (Kelly *et al.*,[7] and Sorensen *et al.*,[8]) under load application. Studies conducted by Oh and Anusavice[9] and Oh *et al.*[10] stated that in connector area, a gingival embrasure with a broader radius of curvature reduced stress concentration under loading and improved fracture resistance at the gingival area of the connector. The relationship between fracture resistance and curvature at the gingival embrasure radii in connector area has not been explored in sufficient detail, however, to allow specific recommendations to be made it has been studied for full-contour monolithic zirconia FPD. Evaluating the stress distribution and stress concentration by finite element analysis is one such method to know the fracture resistance of three-unit FPD.[11,12] The objective of this *in vitro* study was to analyze the magnitude and sensitivity of gingival embrasure radii, as a function of the radius of curvature at the embrasure areas on the fracture resistance of three-unit full-contour monolithic zirconia posterior FPD on various amounts of load application.

Materials and Methods

Two different types of full-contour monolithic zirconia (Lava; 3M ESPE, Seefeld, Germany and Cercon; Degudent, Hanau, Germany) three-unit posterior FPD with two different gingival embrasure radii (rGE) in connector area were prepared using a computer-aided design/computer-aided manufacturing system [Figures 1 and 2]. All the remaining dimensions in connector region included in the study have constant occlusal embrasure (rOE) radius of 0.45 mm with identical sharp notch design and constant connector height width ratio (4:3).[9,10,13,14] Accordingly, two radii of curvature at the gingival embrasure were designed as: Design I, 0.45 mm and Design II, 0.25 mm, respectively.

The geometries of prepared three-dimensional (3D) three-unit zirconia posterior FPD models were transferred to finite element software and were tetra meshed due to the complexity of the prepared tooth and supra-structure shapes by using solid185 element program. Meshing was carried out by giving a meshing command to the software and the number of elements and nodes for Design I = 386336 and 62684 and Design II = 424121 and 62891 were calculated, respectively [Figures 3–7]. The mechanical properties of different

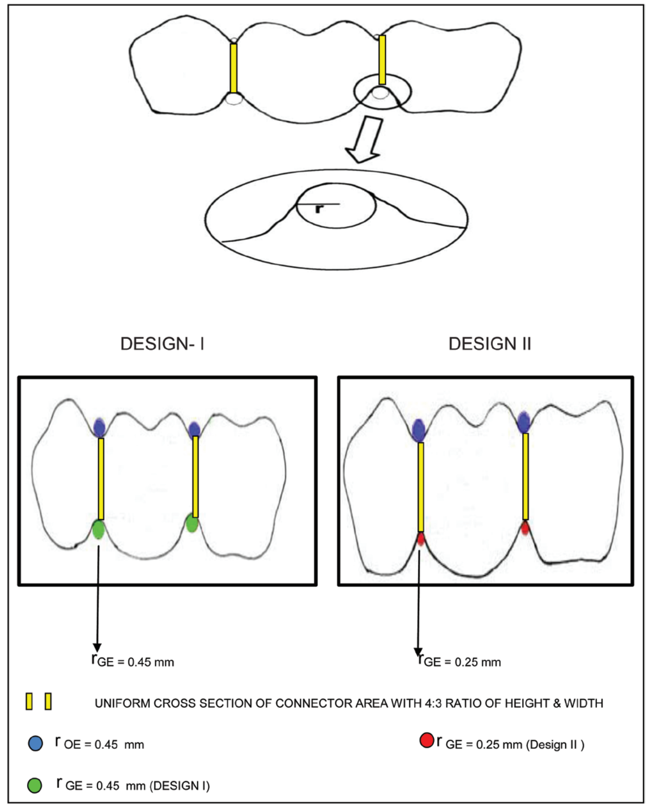


Figure 1: Schematic illustration of connector design

components modeled in the study are shown in table in section "Mechanical properties."

Mechanical properties

Material	Elastic modulus	Poisson's ratio
Dentin	18,600 MPa	0.31
Periodontal ligament	50 MPa	0.49
Cortical bone	345 MPa	0.3
Cancellous bone	13,800 MPa	0.26
Lava	205 GPa	0.19
Cercon	210 GPa	0.3

The components modeled were subjected to magnitudes of static loads of 400, 600, and 800 N, which were applied on the center of the pontic at the central fossa in the vertical direction to simulate the clinical situation.[15,16] All the displacements at the nodes were displayed by post processor (ANSYS) in the form of color-coded maps using von Mises stress values [Figures 8 and 9].

Results

The results of the finite element study (stress distribution and stress concentration) are presented through figures [Figure 10, Tables 1–4] and graphs. The maximum and minimum stress values are denoted by red and blue colors and in between values are represented by bluish green, green, greenish yellow, and yellowish red in their ascending

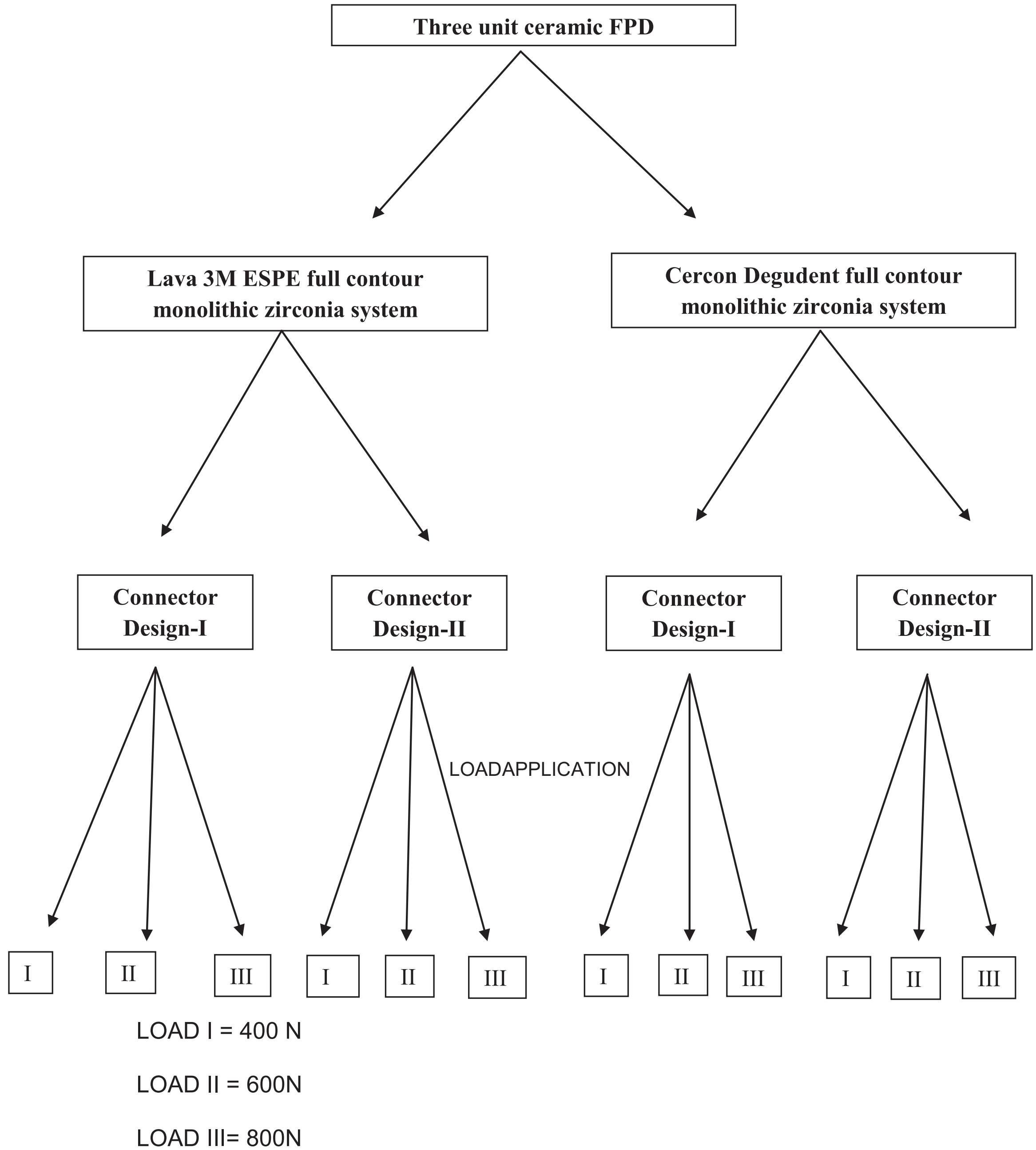


Figure 2: Grouping different parameters involved in the study

order. Each value denotes the beginning of deformation for ductile materials, and fracture propagation occurs when these values exceed the yield strength of a particular ductile material. When a load of 400, 600, and 800 N was applied at the central fossa of the pontic, von Misses stress values were obtained from the results of principal stress at the nodes inside the selected regions of both gingival embrasure region of connector areas (premolar-con and molar-con) for both the zirconia systems. In this study, special attention was dedicated to the maximum tensile stress values as they have a higher potential to cause damage to brittle FPD materials and dental tissues.

Demographic representation of stress concentration

Stress concentration graphs

- Graph I showing when the radius of curvature at gingival embrasure along premolar–molar connector area decreases from design I to design II, von Mises stress concentration increases irrespective of the zirconia system and the amounts of load applied
- Graph II showing when the radius of curvature at gingival embrasure along molar–molar connector

area decreases from design I to design II, von Mises stress concentration increases irrespective of the zirconia system and the amounts of load applied.

- The statistical analysis of the values in graph 3 reveals a significance in stress concentration when compared to stress distribution in both modalities of gingival embrasure radii designs irrespective of the zirconia system as well as the amount of load application.

Discussion

Zirconia-based three-unit FPD showed an excellent clinical survival of restorations as they have the potential to withstand physiological occlusal forces applied in the posterior region. Despite having all of these advantages, these restorations indicate high-failure rates of veneering–porcelain fractures because of its opaqueness. To overcome this, full-contour monolithic zirconia restorations were recently introduced showing high-fatigue resistance and mean molar masticatory forces without superficial fracturing of the layering ceramics. Fabricating monoblock restorations from pure zirconia could increase the mechanical stability and expand the range of clinical suitability.[1-6]

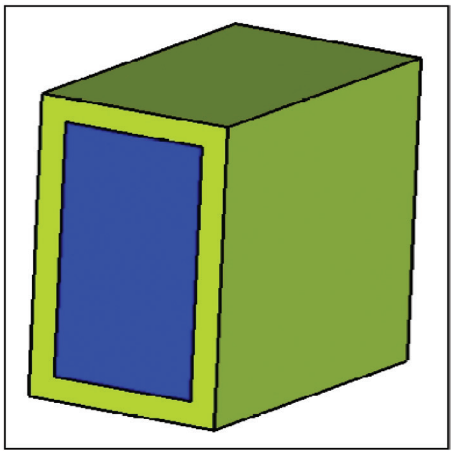
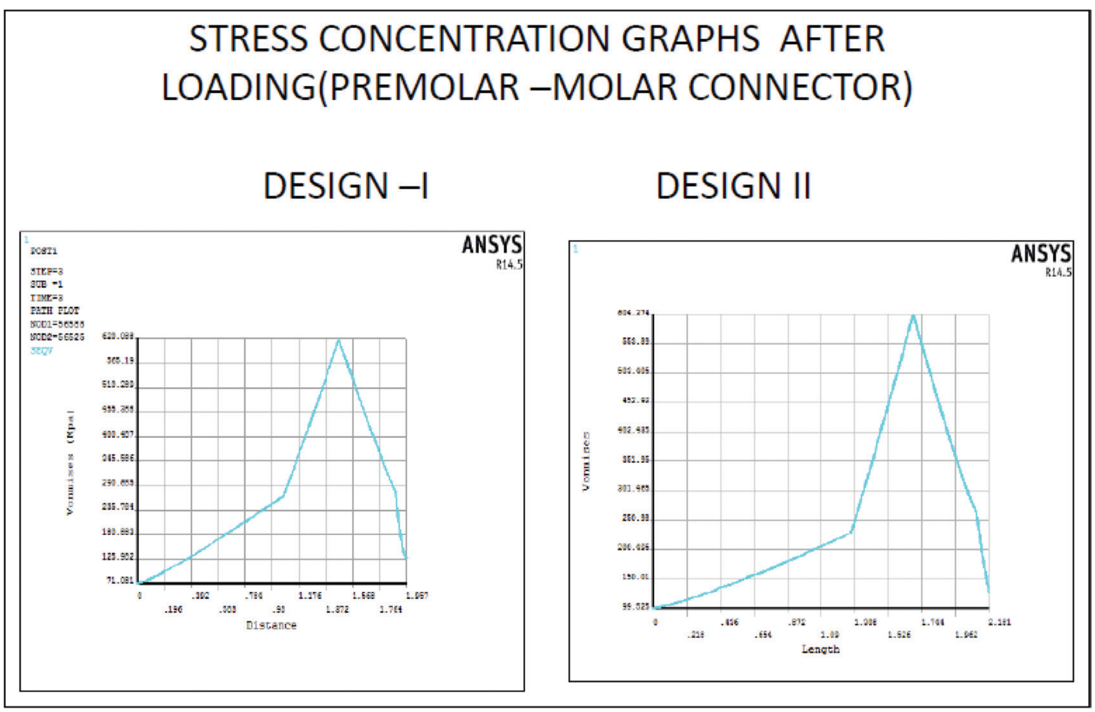
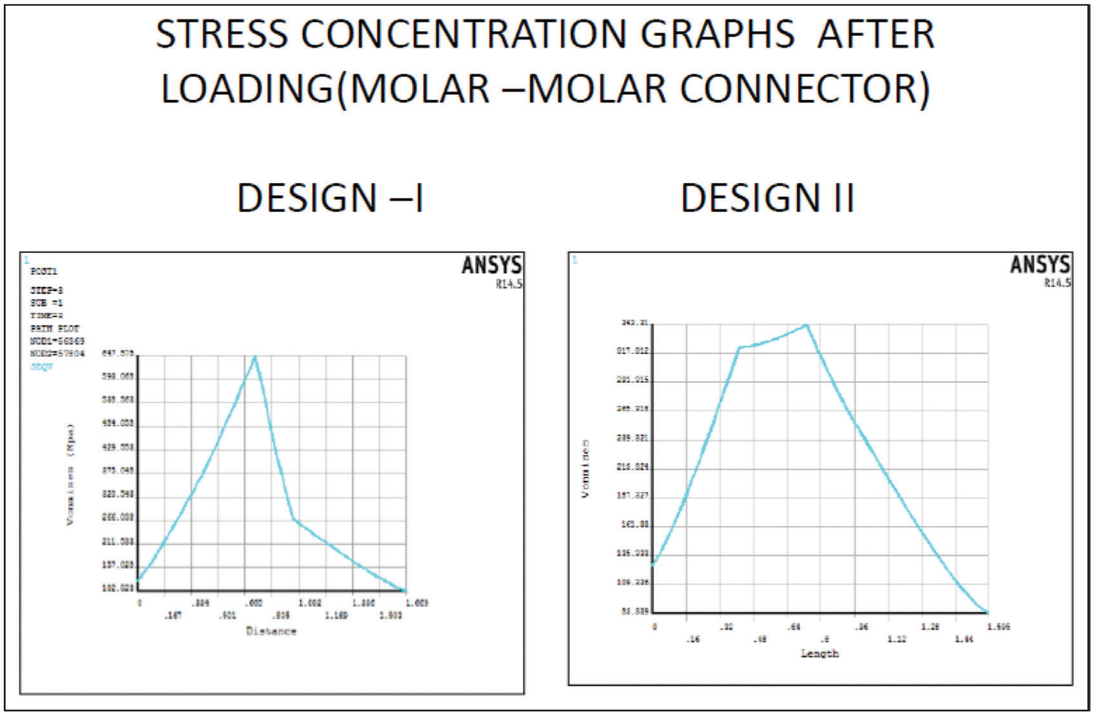


Figure 3: Modeling of D2 density bone

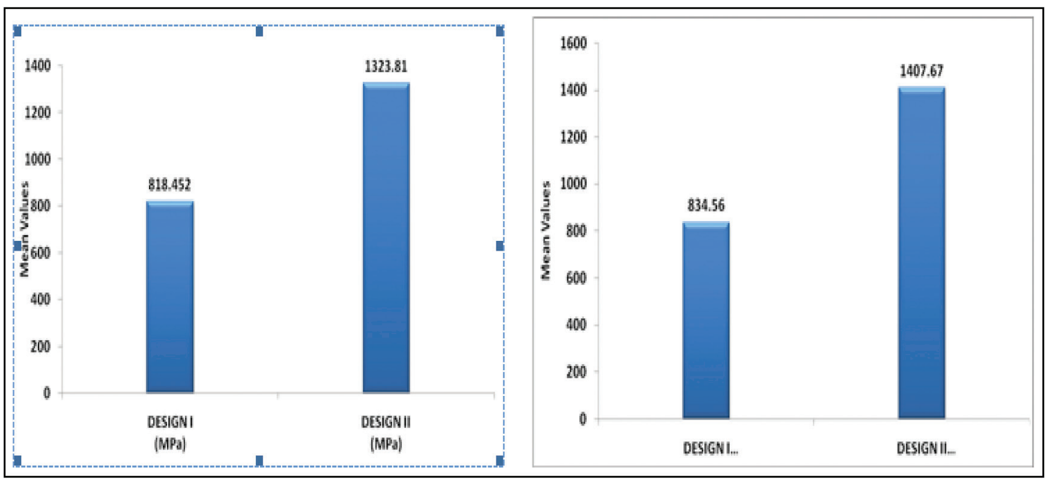
As fixed ceramic prostheses are highly susceptible to tensile forces, in three-unit posterior FPD, the connector area can be considered as a fracture-risk factor, which can increase the tensile stress concentration under flexural compressive loading; hence, it is important to reduce such stress concentrations, especially in the connector area. So, the connector design in the three-unit ceramic FPD is altered in the regions where maximum stress occurs so that the characteristic stress pattern may be altered to improve the fracture resistance of three-unit FPDs. Studies conducted by Kelly *et al.*[7] (1995), Sorensen *et al.*[8] (1999), Komposiora *et al.*[17] (1996), and Kohorst *et al.*[18] (2007) suggest that stresses are better distributed with broadly curved connectors than through the use of more sharply curved connector geometries. Oh *et al.*[10,11] reported that the radius of gingival embrasure of connector area has major effect, whereas the radius of curvature at the occlusal embrasure had only a minor



Graph 1: Stress concentration graph after loading in premolar-molar connector



Graph 2: Stress concentration graph after loading in molar-molar connector



Graph 3: Statistical analysis

effect on the fracture susceptibility of three-unit FPDs. The study conducted by Plengsombut *et al.*[19] (2009) showed that the fracture of all ceramic FPDs was initiated from gingival surface of the connector and propagated toward pontic. Similarly, the study conducted by Bahat *et al.*[20] (2009) showed that radii of curvature in the gingival embrasure of the connector area could affect the fracture resistance of anterior three-unit Yittrium partially stabilized tetragonal zirconia FPDs.

A suitable engineering tool to predict the fracture resistance in dental restorations is finite element analysis. In these analyses, the material properties were assigned to the individual components and after applying the loads and constraints, meshing and approximations were carried out, and the resultant stresses were analyzed through the von Mises stress criteria in the form of color-coded bands. As the von Mises stresses could not be distinguished into specific contributions of tensile, compressive, and shear stresses, the mechanical behavior of the three-unit FPD was investigated as functions of magnitude of the load applied. However, the reported results of any Finite element analysis (FEA) are the normal and shearing stress values of the structure on loading. The primary normal stresses act

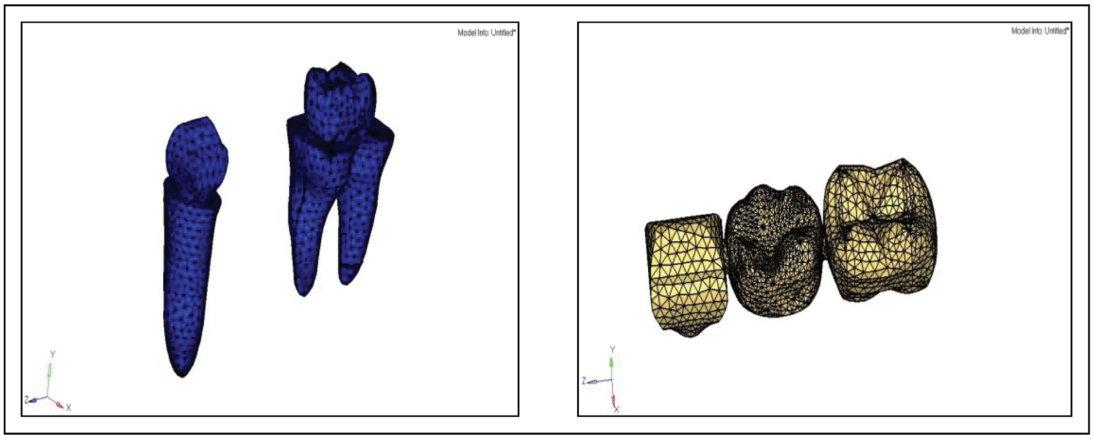


Figure 4: Modeling of abutment teeth and supra-structure

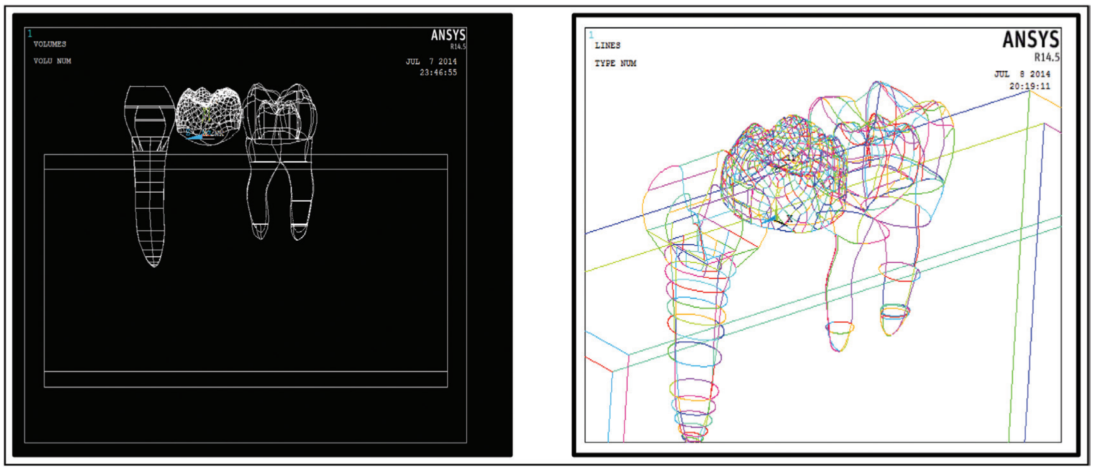


Figure 5: Modeling of three-unit FPD-line diagram

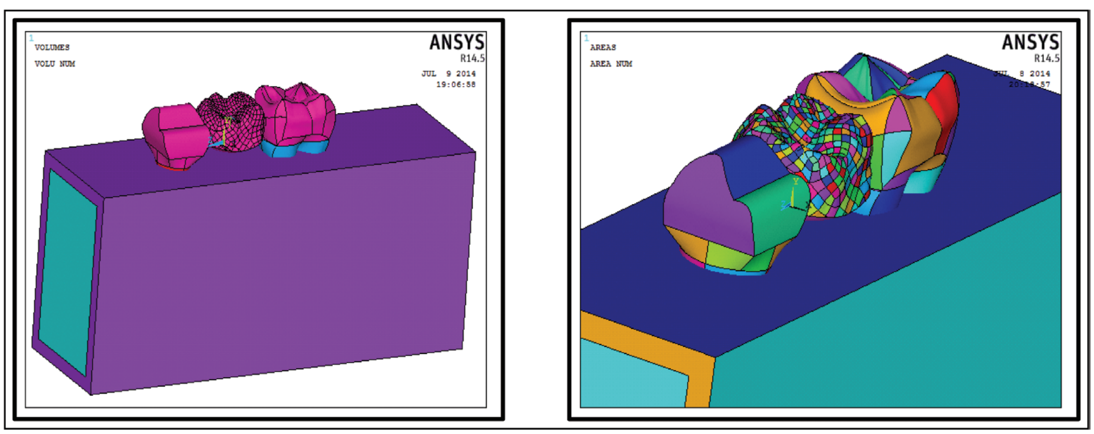


Figure 6: Modeling of FPD in D2 density bone

on the principal planes, on which the shearing stresses are zero. The structural configuration and loading may result in all positive, all negative, or a combination of positive and negative principal stresses, whereas the von Mises stress is always positive. Von Mises criteria, which result in a tensile-type normal stress, were chosen because the brittle ceramic material primarily fails because of tensile-type normal stresses. A negative-stress value would represent compression and a positive value would represent tensile stress.[11,12,14,21,22]

Almost all laboratory studies pertaining to the mechanical behavior of an FPD use a load applied on the pontic

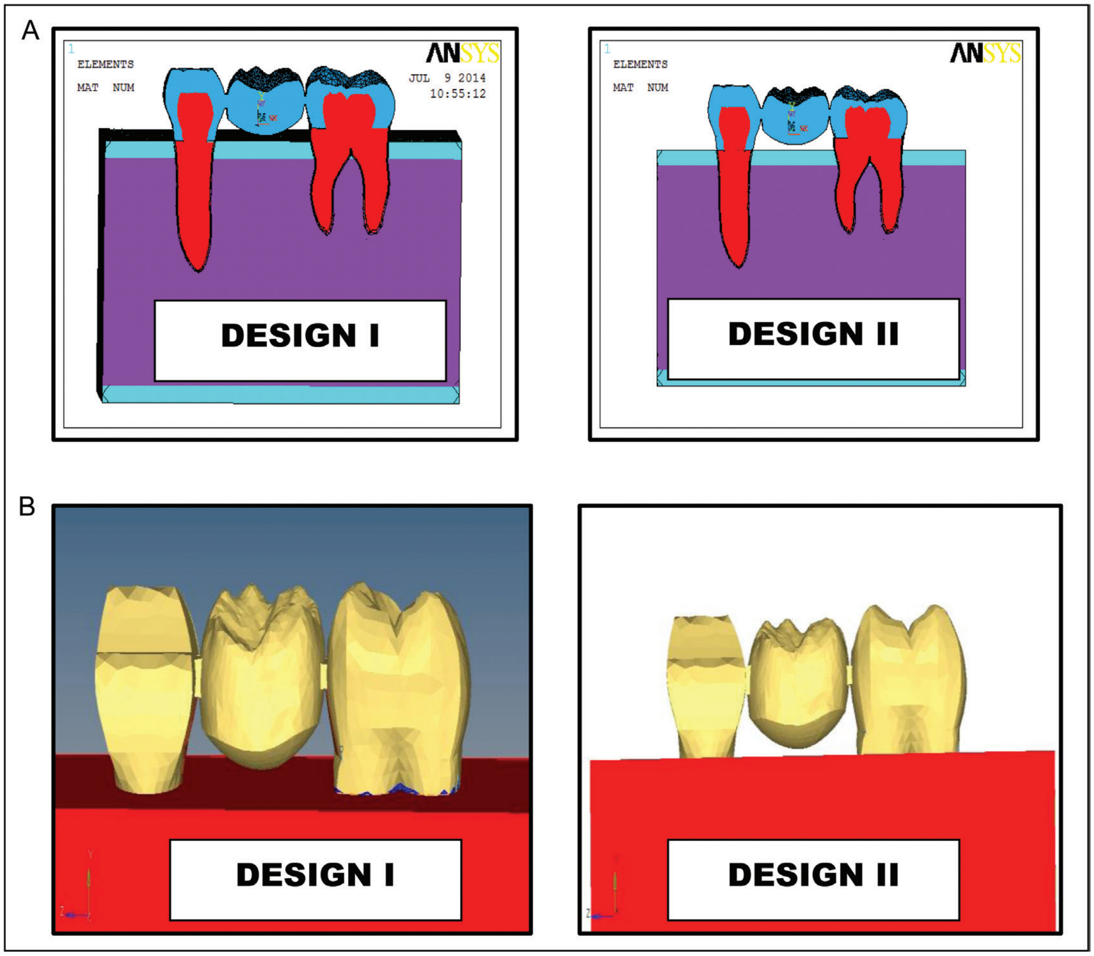


Figure 7: Modeling of three-unit FPD connector designs

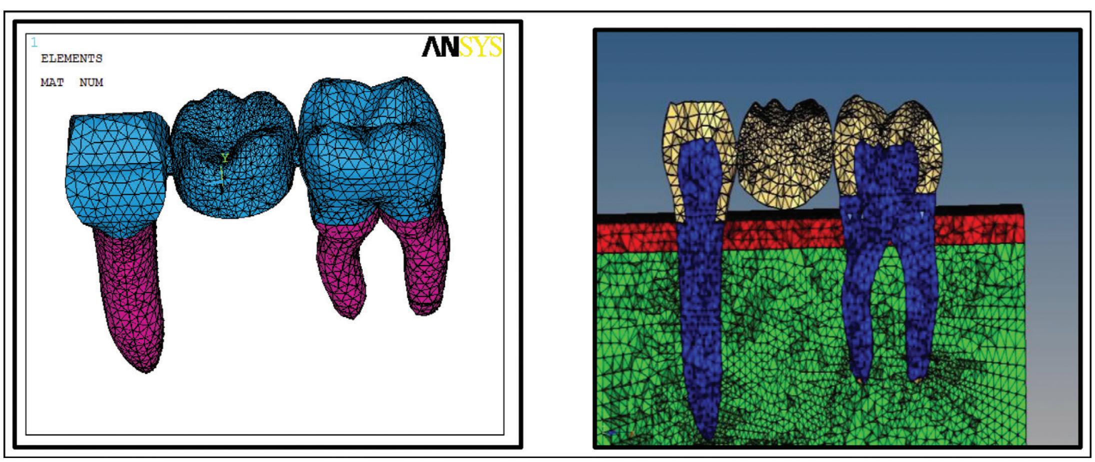


Figure 8: Mesh structure of three-unit FPD

central region.[11] When an FPD is subjected to functional loading, the transferred forces give rise to stress, and in turn, to strain within the structure. Several factors can influence the resulting stress distribution, for example, the magnitude and direction of applied forces.[23]

So in this study, the bridges were loaded by 400, 600, and 800 N on the occlusal surface at 90° in the middle of the pontic component. The load was distributed onto three nodes of adjoining finite elements. Even though the amount of load taken was very large for a single occlusal contact, it was considered so that we could analyze the stress distribution under extreme conditions.

When a load of 400, 600, and 800 N was applied at the center of pontic, the von Mises stresses recorded for the

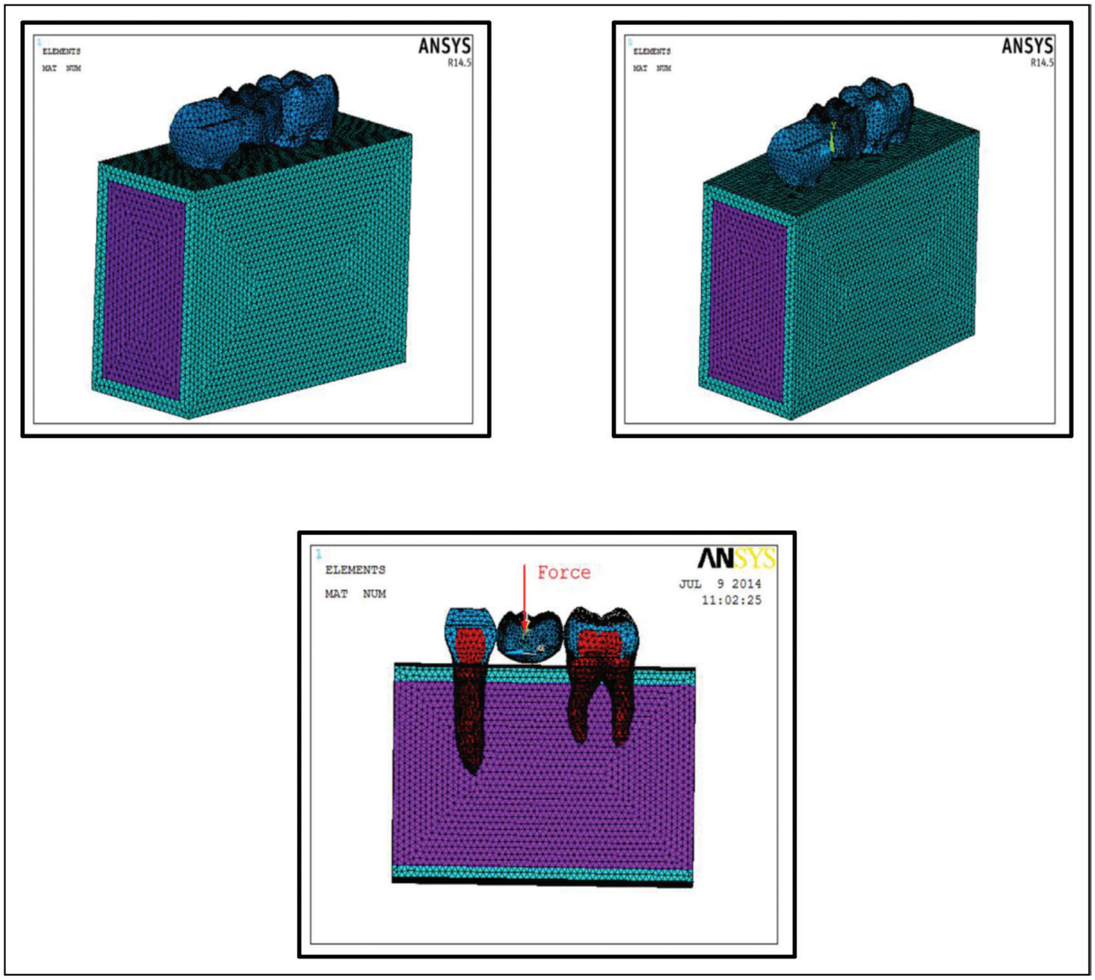


Figure 9: Application of load

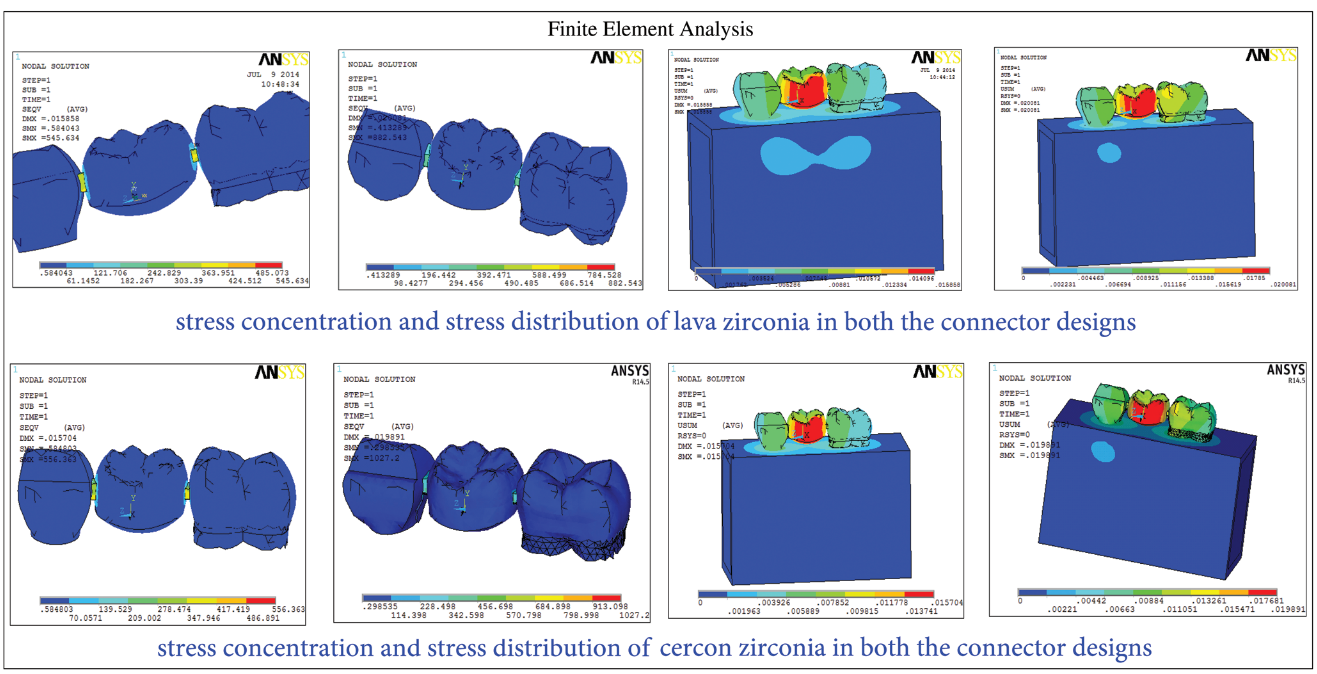


Figure 10: Finite element analysis

Table 1: Stress concentration for lava zirconia in both connector designs- Stastical analysis

Load (N)	Design I (MPa)	Design II (MPa)
400	545.634	882.543
600	818.451	1323.81
800	1091.27	1765.09
Mean ± SD	818.452 ± 272.82	1323.81 ± 441.27
P value	0.035 (Sig.)	

connector designs - Stastical analysis

Table 2: Stress concentration for cercon zirconia in both connector designs- Stastical analysis

Load (N)	Design I (MPa)	Design II (MPa)
400	556.363	1027.2
600	834.545	1369.6
800	1112.78	1826.2
Mean ± SD	834.56 ± 278.21	1407.67 ± 400.86
P value	0.016 (Sig.)	

connector designs- Stastical analysis

Lava three-unit zirconia FPD along connector design I were 545.634, 818.451, and 1091.27 MPa, respectively, and for connector design II, they were recorded as 882.543, 1323.81, and 1765.09 MPa, respectively. Similarly, stress distribution/displacement for connector design I was recorded as 0.01, 0.02, and 0.03 MPa and for design II as 0.02, 0.02, and 0.03 MPa, respectively.

For Cercon three-unit zirconia FPD, when a load of 400, 600, and 800 N was applied at the center of pontic, the von Mises stresses recorded along connector design I were 556.363, 834.545, and 1112.78 MPa and for connector design II, they were recorded as 1027.2, 1369.6, and 1826.2 MPa, respectively. Similarly stress distribution/displacement for connector design I was recorded as 0.01, 0.02, and 0.03 MPa and for design II as 0.19, 0.02, and 0.03 MPa, respectively.

The analysis of the stress concentration in the loaded monolithic zirconia three-unit bridge showed that the connector area between the bridge abutment and the pontic component was the critical part of the fixed dental restoration. When a vertical load was applied at the center of the pontic in a three-unit FPD, high stresses were observed in the gingival embrasure region of connector area. This propagates the cracks from the gingival embrasure toward the occlusal loading on the pontic. So, the fracture of origin was most commonly at the center of the gingival embrasure in the buccolingual dimension and was shifted slightly toward the abutment crown in the mesiodistal direction. Moreover, it showed that connector design I, having a larger radius of curvature at the gingival embrasure, exhibited a lower tensile stress concentration value compared with design II having smaller radii of curvature irrespective of the zirconia system and the amount of load applied. As the radii of curvature at gingival embrasure were decreasing from design I to design II, the tensile von Mises stress concentration was increasing, whereas no significant variation was observed in stress distribution pattern. This result was in agreement with the previous findings of finite element analysis and photoelastic studies conducted by Oh *et al.*[10] and Komposiora *et al.*[11,17]

Table 3: Stress distribution for lava zirconia in both connector designs- Stastical analysis

Load (N)	Design I (MPa)	Design II (MPa)
400	0.01	0.019
600	0.02	0.02
800	0.03	0.03
Mean ± SD	0.02 ± 0.01	0.023 ± 0.006
P value	0.423 (Not significant)	

Table 4: Stress distribution for cercon zirconia in both connector designs- Stastical analysis

Load (N)	Design I (MPa)	Design II (MPa)
400	0.01	0.019
600	0.02	0.02
800	0.03	0.03
Mean ± SD	0.02 ± 0.01	0.023 ± 0.006
P value	0.423 (Not significant)	

Hence, the results yielded from this mathematical (FEA) analyses study confirmed that the morphology of the connector design at the gingival embrasure is critical in reducing the fracture probability. Considering this, the fracture resistance can be improved by increasing the radius of curvature at the gingival embrasure of the FPD, without affecting aesthetics while keeping the occlusal embrasure as sharp as possible.

Elastic modulus plays an important role in determining the fracture resistance and stress distribution of FPD, particularly in the mathematical studies such as FEA and photoelastic analysis. Previous studies conducted by Scherrer and de Rijk[24] stated that the fracture resistance of an all-ceramic crown was dependent on the modulus of elasticity of the supporting material. The fracture load increased markedly with the increase in elastic modulus. The largest increase was seen when only the occlusal surface of the crown was covered. Studies conducted by Proos *et al.*[25] concluded that the peak tensile stress magnitude value increases with the increase in the value of elastic modulus of core materials. Similarly, study conducted by Möllers *et al.*[26] showed that with the decreasing Young's modulus of the veneering material, the load was transferred to the framework material, resulting in higher stress concentrations. But the studies conducted by Beuer *et al.*[27] and Preis *et al.*[28] showed milling zirconia to full-contour systems with glazed surface might be an alternative to traditionally veneered restorations.

Comparative data of von Mises stress concentration between two zirconia systems showed that with slight increase in the elastic modulus (210 > 205 GPa), Cercon zirconia showed slightly higher stress concentration than Lava zirconia, irrespective of the connector design. This result was in agreement with the aforementioned studies confirming that elastic modulus plays an important role in determining the stress distribution.[29,30] However, variation of elastic modulus of both the full-contour systems was very less and as the same zirconia material was used, the fracture resistance or stress distribution might depend on their fabrication technique, surface treatment, and aging of the material.[31,32] Further research has to be carried out to check whether these effects limit the clinical reliability of those restorations.

Similarly, as the amount of load on the pontic increases, the amount of maximum principal stress generated in three-unit zirconia FPD also increases, irrespective of the connector design and zirconia system. As the amount of the load increases from 400 to 600 N and 600 to 800 N, the stress concentration also increases by 30% approximately. However, higher or lower load values would only change the magnitude of stress but not the distribution pattern in finite element analysis. But if the load configuration, that is, the site of application load is changed then the stress distribution also changes.

With the available data for both the zirconia systems, the load-bearing capacity of posterior three-unit zirconia FPDs was approximately 1600–1800 N when they were under static load immediately.[2,3] In general, they confirm the findings of this study, subject to the usual difficulties encountered when comparing results from different load applications.[31-34]

Technical Limitations

This study has certain limitations:

1. The vital anisotropic tissues were assumed as isotropic and homogeneous, and the loads applied were static loads that were different from dynamic loading seen during function.
2. No adhesive or cement layer and pulp were included in the FEA model. These simplifications may have an effect on the stress distribution.
3. Fabrication technique, surface treatment, and aging of the material might affect the clinical suitability.
4. Finite element analysis is based on mathematical calculations, which are based on simulation of structure in its environment. But biological tissues are beyond the confines of set parameters, and values are more than mere objects, as biology is not a compatible entity.

Summary and Conclusion

Within the limitation of this 3D FEA study, the results of this *in vitro* study were concluded as follows:

1. The gingival embrasure at the connector region of the three-unit zirconia FPD represented the tensile stress concentration zones when the load was applied at the center of the pontic.
2. With the decrease in the gingival embrasure radii, the stress concentration values were found to be increasing for both the full-contour zirconia systems (Lava and Cercon) on various amounts of load application.
3. Among both the full-contour zirconia systems, Cercon zirconia showed higher stress concentration values compared to Lava zirconia in both connector designs on various amounts of load application.
4. With the increase in the load application at the center of the pontic, the tensile stress concentration values were also increasing, irrespective of the change in the connector design and zirconia systems.
5. There is no significant change in stress distribution pattern or displacement stress for both the zirconia systems and connector designs.

The computational method (the finite element method) used is a suitable tool for predicting different life expectancies for different ceramic bridge materials and different connector designs. However, it should be stressed that the calculated life expectancies will not be identical with the respective clinical life times because of the mentioned limitations used for the numerical simulations.

In general, the radii of curvature of gingival embrasure at connector area could affect the fracture resistance of three-unit FPD on load applications, irrespective of the ceramic system. Depending on the clinical situation, when there are heavier occlusal forces, the fracture resistance of full-contour monolithic posterior three-unit zirconia FPD can be increased by designing greater radii of curvature at gingival embrasure in the connector region.

In a further study, the predicted life expectancies should be correlated with experimental clinical data. Such correlations could reveal *in vivo* influences that additionally affect the long-term failure probability of zirconia-based ceramic bridges.

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Nil.

Conflicts of interest

There are no conflicts of interest.

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