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# Influence of connector dimensions on the stress distribution of monolithic zirconia and lithium-di-silicate inlay retained fixed dental prostheses – A 3D finite element analysis

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## Abstract

*Purpose:* The objective of the study was to analyze the stress distribution between monolithic Lithium-disilicate and monolithic Zirconia inlay retained Fixed Dental Prostheses by varying the connector dimensions using the 3D- Finite Element Analysis.
*Methods:* Two models of three unit inlay retained Fixed Dental Prosthesis replacing the lower right first molar was fabricated, each with the connector dimensions of 3 mm × 3 mm and 4 mm × 4 mm. Using three dimensional Finite Element Analysis, the Poisson's ratio and Young's modulus for monolithic Zirconia and monolithic Lithium-di-silicate were added to each of these groups. These were then subjected to a vertical load of 500 N directed occlusally over a surface area of 5 $mm^2$; and the results were analyzed.
*Results:* By increasing the connector dimensions up to 4 mm × 4 mm, both the materials are capable of withstanding a force of up to 500 N, simulating the maximum posterior bite force. According to this study, monolithic Zirconia and Lithium-di-silicate can be used a posterior restorative material in all ceramic inlay retained Fixed Dental Prosthesis.
*Conclusion:* Increasing the connector dimensions up to 4 mm × 4 mm, has a significant improvement in the stress distribution among both materials, making it suitable as a posterior restorative material in all ceramic inlay retained Fixed Dental Prosthesis. Further long term clinical studies are required.


*Keywords:* Zirconia; Lithium-di-silicate; Inlay retained fixed dental prostheses; Connector; Finite element analysis

## 1. Introduction

There are various treatment modalities available to replace a missing posterior tooth. The conventional removable prosthesis, crown retained – Fixed Dental Prostheses (FDPs) and the recently booming implant therapy [1]. The risk of involving pulp and also of increase in the coronal tooth structure removal is high in during conventional fixed dental prostheses [2].

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If the clinical situation does not seem convincing for implants, where sound tooth structure is available or in cases of tilted abutments, an inlay retained Fixed Dental Prostheses (IRFDPs) can be used as the treatment of choice for restoration [3].

But these IRFDPs have a higher risk of dislodgement and fracture due to the minimal preparation [4]. Initially, cast resin bonded FDPs were manufactured exclusively using noble metals like high-gold alloy [5,6]. A wide range of new materials are available nowadays: hybrid microfilled or fiber-reinforced composites (FRC), ceramics with a high content of glass particles (i.e., lithium disilicate, glass-infiltrated zirconia. or alumina), or high strength ceramics (densely sintered zirconia/alumina polycrystal) to be used as frameworks for subsequent veneering or to fabricate monolithic restorations [7,8].

Changes in material strength can also be due to the difference in ceramics used. Monolithic ceramics when compared to bilayered ceramics seem to show superior quality in terms of fracture resistance [9,10]. There are various extensive studies comparing the compatibility of IRFDPs as opposed to crown retained FDPs, to which it states that IRFDPs can be safely used as an alternative [11–14]. There are not many studies comparing the difference in fracture resistance in two different all ceramic systems.

Therefore, the purpose of this study was to evaluate the stress distribution between monolithic Lithium-di-silicate and monolithic Zirconia inlay retained Fixed Dental Prostheses by varying the connector dimensions using 3D- Finite Element Analysis.

## 2. Materials and methods

### *2.1. Preparation of model*

Cylindrical die of about 25 mm high and 16 mm wide was fabricated. Into the die, autopolymerising resin is mixed and typhodont teeth (45, 47) were embedded till the cemento enamel junction, leaving an average gap of about 11 mm bucco-lingually and 10 mm mesiodistally for the first molar.

Using a flat end diamond bur, a distal inlay in 45 and a mesial inlay in 47 was prepared. Cavity depth of between 1.5 and 2 mm; isthmus cavity width of more than 1/3 the intercuspal width (around 2 mm); total occlusal convergence of 20°, and rounding of all internal line angles was done.

The preparation was completed according to the principles involved for all ceramic inlay retained Fixed Dental Prosthesis preparation [15]. The preparation was cross checked by using a putty index which was taken prior to the preparation.

Impressions were made using double stage putty[1] and light body[2] and sent to laboratory for the prosthesis fabrication of both monolithic Zirconium and Lithium-di-silicate inlay [16].

The finite element analysis is a computer aided mathematic technique for obtaining accurate numerical solutions used to predict the response of physical systems that are subjected to external stress. Basically, any problem can be split up into a number of smaller problems with finite element method. It uses subdivision of a whole problem domain into simpler parts, called finite elements, and variational methods from the calculus of variations to solve the problem by minimizing an associated error function. This is done by considering that a complex geometrical shape is made up of a number of simpler shapes and each simple shape being known as an “element” and the whole collection of elements being known as “mesh”.

The relevant property of each material is found within each element. Material properties such as young's modulus and Poisson's ratio can be utilized by computer generated analysis to describe the mechanical behavior, induced stresses, or the relationship between forces and displacements for a structural element.

Type of stresses in finite element studies are generally described by means of direction (shear, tension, and compression) or by an effective absolute magnitude of principal stresses (equivalent stress of Von Mises).

### *2.2. Steps in FEA*

1. Use of Optical Comparator.
2. CATIA V5® - for designing the prosthesis and modeling.
3. HYPERMESH® - Meshing the models.
4. RADIOSS® - Finite Element Analysis to find out Principal stresses and von Mises stresses.

The scanned model was observed for dimensions and structural formation through the optical comparator (DV 114®, Deltronic Optical Comparator, USA). The magnification was set to 10× for better observation. Here, the magnified silhouette of a part was projected upon the screen, and the dimensions and geometry of the part were measured against prescribed limits.

For data preprocessing and to convert two-dimensional scan reformatted views to a three-

[1] Aquasil Soft Putty/Regular set, DENTSPLY DETREY GmbH, Germany.

[2] Aquasil LV, DENTSPLY International Inc., USA.

dimensional (3-D) model CATIA V5® and Reverse Engineering (RE) was used and HyperMesh10® (HM 10) was used for meshing the models. After which the model was used for analysis with FEM software RADIOSS® (Altair Engineering).

A mandibular D-2 (misch classification) bone model was simulated. The shape of the bone was simplified to a cuboidal block. A bone block 25 mm in height, 16 mm wide was designed. The interface between the tooth and bone was to be an immovable junction since FEA models assume a state of optimal bone, meaning that bone are assumed to be perfectly bonded to the tooth without any bone loss. The model of single tooth simulated the second premolar and second molar. For simplicity, cement thickness was not included in the models. All materials used in the models were considered to be isotropic, homogenous and linearly elastic.

The resulting three-dimensional Finite Element Models were used as the basis for the analysis. These models were meshed using 10-node quadratic tetrahedral elements.

### 2.3. Post-processing

In this study, the pontic and abutments were assumed to be subjected to a vertical load of 500 N, directed lingually, which corresponds to a steel ball of approximately 5 $mm^2$.

In FEA, the stress distribution analysis can be recorded by von Mises criteria or maximum principal stress. The stress analysis of von Mises does not have appropriate failure criterion for brittle materials. Therefore, maximum principal stress can be adopted to analyze the results.

Resultant geometry was brought into the FEA programme RADIOSS® (Altair Engineering) for post-processing, with the analysis displaying principal and von Mises stresses (Tables 1 and 2).

## 3. Results

Figs. 1 and 5 display the Principal stresses of Lithium-di-silcate when load of about 500 N is given. It can be clearly visualized, the differences in stress between 9 $mm^2$ connector and 16 $mm^2$.

Figs. 2 and 6 reveal the Principal stresses of Zirconia, which also decreases on increasing the connector dimensions.

Figs. 3, 4, 7 and 8 display the resultant von Mises (also known as the Distortion Energy Theory) stress contours. These clearly show that the highest stress concentrations exist in the vicinity of the embrasure areas between the inlay and pontic and at the loading contact site. It can be concluded that von Mises stress and displacement peaks concentrate around the connector areas.

Table 2
Material property values [26].

| S. no | Name of the material | Young's modulus (GPa) | Poisson's ratio |
|---|---|---|---|
| 1. | Teeth | 18.6 | 0.31 |
| 2. | Bone (average of bone properties) | 3.7 | 0.3 |
| 3 | Y-TZP | 205 | 0.30 |
| 4. | Lithium-di-silicate | 95 | 0.23 |

## 4. Discussion

The increase in patient demand as well as to deliver high strength restorative option without compromising on the esthetic front is the challenge faced by most clinicians. The metal free restorative options such as Zirconia and Lithium-di-silicate face a major disadvantage [17].

The first one being that the core material had a high value and opacity which was disadvantageous when conservative tooth preparation was done as it might not be esthetically pleasing. The next one was that, the layering ceramic veneered had a much lower flexural strength and fracture toughness when compared to the core material [18].

The next challenge while encountering layering ceramic was, they have a much lower flexural strength and fracture toughness as compared to the core material. The Zirconia core (flexural strength: 900–1000 MPa) consists of less than half the cross-sectional thickness of the restoration. The remaining

Table 1
Groups.

| Groups | Connector dimensions | Sub group | Name |
|---|---|---|---|
| I | 3 mm × 3 mm (or) 9 $mm^2$ | Li-I | Lithium-di-silicate inlay of connector dimension of 9 $mm^2$ |
| | | Zi-I | Zirconia Inlay of connector dimension of 9 $mm^2$ |
| II | 4 mm × 4 mm (or) 16 $mm^2$ | Li-II | Lithium-di-silicate inlay of connector dimension of 16 $mm^2$ |
| | | Zi-II | Zirconia Inlay of connector dimension of 16 $mm^2$ |

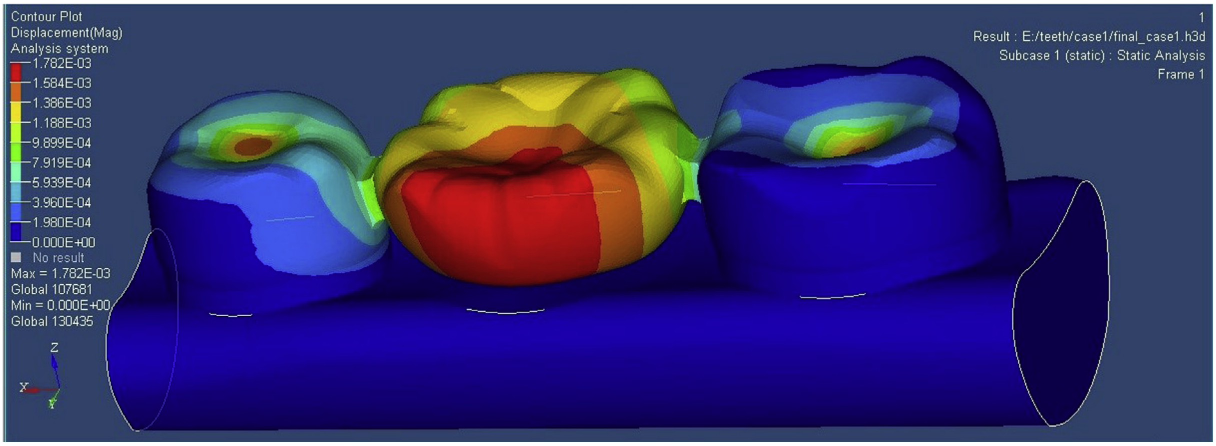


Fig. 1. Group I: Li-I:Maximum principal stress for Lithium-di-silicate – 9 mm$^2$.

thickness must be completed using a veneering material which has a flexural strength of approximately 80–110 MP, which again depends on whether it is delivered through a powder build-up or by pressing. This makes the layering ceramic, a weak link in restorations, as it has to resist chipping or fracturing during function [19].

Monolithic glass-ceramics offers some distinct advantages-they provide exceptional esthetics without requiring a veneering ceramic. By excluding the veneered ceramic and its requisite bond interface, greater structural integrity and strength can be achieved.

Uniform stress distribution in the larger area and higher stress toleration were observed specially at the connectors in this study.

Also, the principal stress distribution was not equal throughout the structure, and the stress gradient decreased from Group I (9 mm$^2$) to Group II (16 mm$^2$).

In Group I (9 mm$^2$), both Li–I and Zi-I groups show an almost uniform distribution of stress, mostly being concentrated on the central fossa of the inlay in the premolar and second molar. Also, in Li–I, stress is concentrated on the lingual aspect, meaning the stress cannot be dissipated by the material as done by Zi-I.

Group II (16 mm$^2$), reads a subsequently less stress, on the pontic as well as the inlay in both Li-II and Zi-II. As the connector dimensions were increased by 1 mm in both dimensions, much variation were not seen.

Oh and Anusavice have suggested that fracture probability may be significantly minimized by using a connector with a curvature radius of 0.9 mm approximately. To reduce the stress concentration and to maintain a constant connector height of 4.0 mm without refining the curvature at the gingival embrasure, they have proposed that gingival embrasure

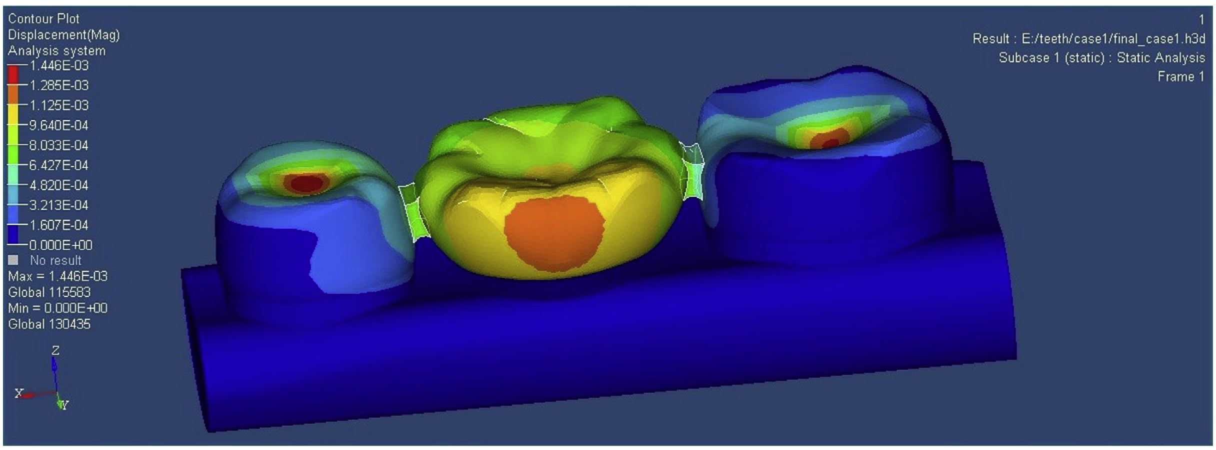


Fig. 2. Group I: Zi-I: Maximum principal stress for Zirconia – 9 mm$^2$.

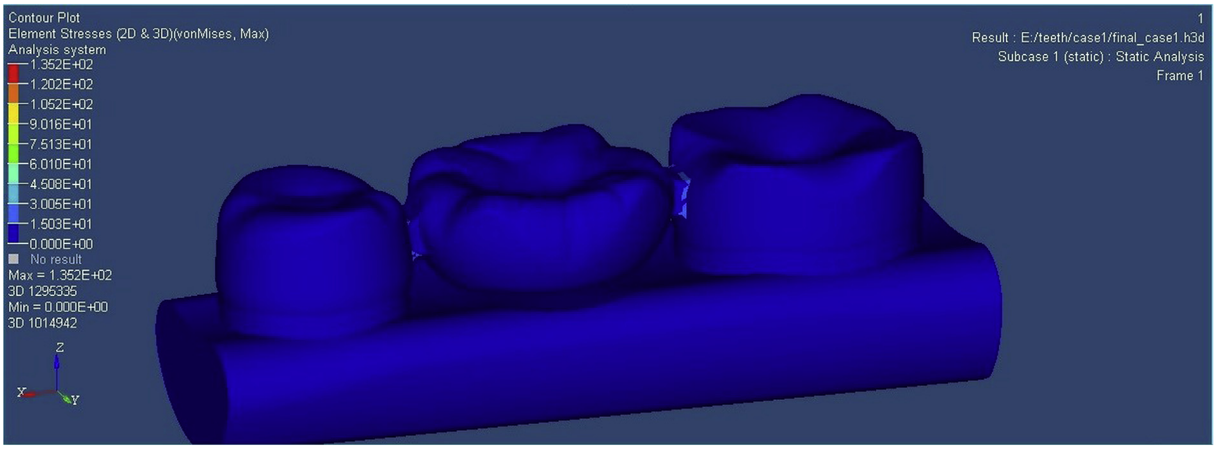


Fig. 3. Group I: Li–I: von Mises stress contours for Lithium-di-silicate – 9 $mm^2$.

curvature radius be of 0.45 mm. This propagates the cracks from the gingival embrasure toward the occlusal loading on the pontic. The fracture origin was most commonly at the connector area, especially toward the inlay region [20].

The current Finite Element Analysis shows that a critical area of von Mises stress peaks in the inlay retained Fixed Dental Prosthesis is located around the connector. This result is in agreement with other finite element research [14].

Comparable results were obtained compared to previous clinical and experimental studies [21–23]. Here, the inlay retained Fixed Dental Prostheses with Zirconia framework demonstrated greater fracture resistance compared to that of lithium-di-silicate based inlay retained Fixed Dental Prostheses framework.

Although, the von Mises stress shows a tension majorly in the connector, according to the study, more dissipated force, and less tension is seen in Group II.

Maximum von Mises stress under a 500 N load was found in Group Li–I, Group Li-II, followed by Group Zi-I and Zi-II in ascending order. The 3D model enabled to investigate the stress distribution within and along any section of the bridge.

In the present study, a Finite Element Model was utilized where the underlying structure was dentin, and influences due to the pulp, periodontal ligament, or the adhesive or cement layer were not considered [24].

The materials were assumed to be homogeneous, isotropic and linearly elastic. The bridge configuration was modeled based on the geometry of the natural teeth. Models were broken down into nodes, elements and meshed. Changes in prosthesis component contours, particularly abrupt ones, may affect the stress distribution in a ceramic prosthesis.

Local stresses can significantly increase the overall stress at highly curved regions, such as at surface notches or other abrupt shape changes. This effect may

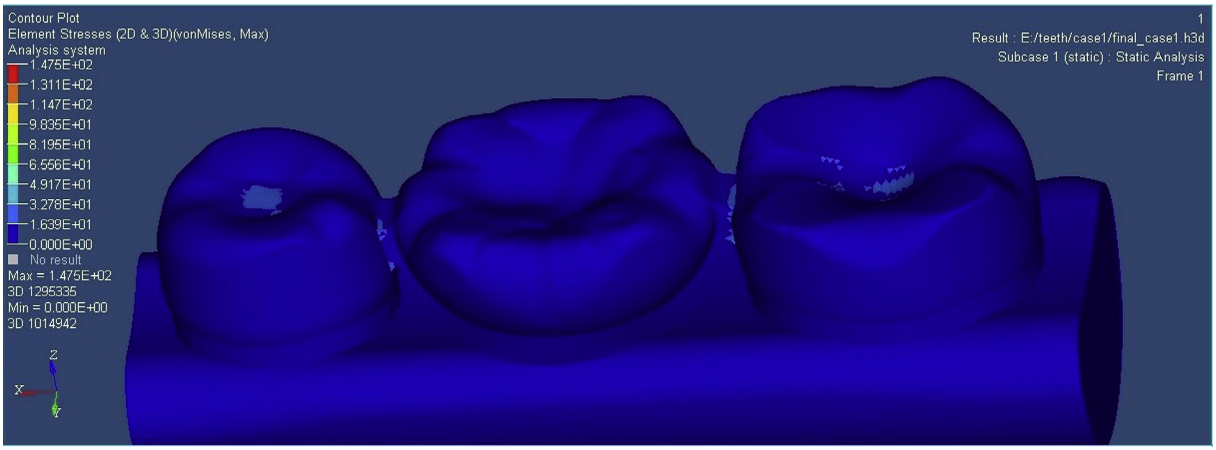


Fig. 4. Group I: Zi-I: von Mises stress contours for Zirconia – 9 $mm^2$.

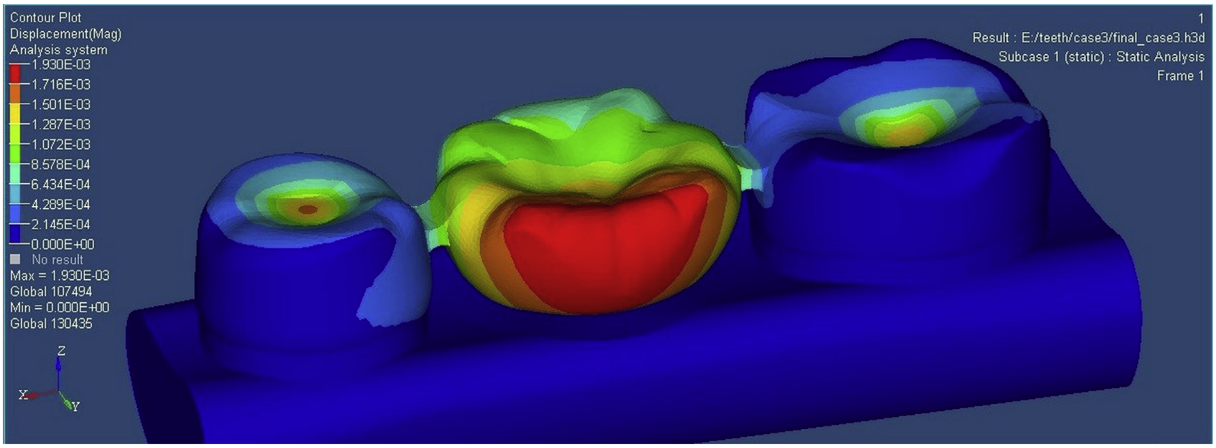


Fig. 5. Group II: Li-II: Maximum principal stress for Lithium-di-silicate − 16 $mm^2$.

be much more significant in a brittle material, such as ceramic which contains many small flaws or cracks of various sizes and orientations [25].

These factors may also be more critical in posterior inlay retained Fixed Dental Prostheses. Posterior areas experience higher loads, and the connector height may be limited by the short clinical molar crowns. Cracks are initiated adjacent to load points and propagate along the plane of maximum tensile stress to the gingival side of connectors.

The failed surface displayed some pores and incomplete crystallized or densified areas.

The height of the mastication force significantly influenced our results. Numerous authors investigated the maximum bite forces during mastication. The average chewing force in literature varies between 11 and 150 N, whereas force peaks are 200 N in the anterior, 350 N in the posterior and 1000 N with bruxism [26].

The amount of fracture resistance that is needed to achieve a good long-term outcome of inlay retained Fixed Dental Prostheses in the molar region is not known. Mean values for the maximum bite force level varied from 216 to 847 N. The highest bite force was found in the first molar region. Reviewing the literature, Körber and Ludwig summarized that posterior Fixed Dental Prostheses should be strong enough to withstand a load of 500 N [27−33].

Among the structural factors, the connector areas are the most influential in failure. Failure rate is relatively high in three unit all-ceramic bridges around the sharp connector area. The Fixed Dental Prosthesis shape is not uniform clinically, but is a complex combination of multiple convexities and concavities that depend on the geometry and alignment of the teeth. In all ceramic resin-bonded Fixed Dental Prosthesis, the occluso-gingival height of the interdental connector must be as large as possible (minimum 4.0 mm) [21].

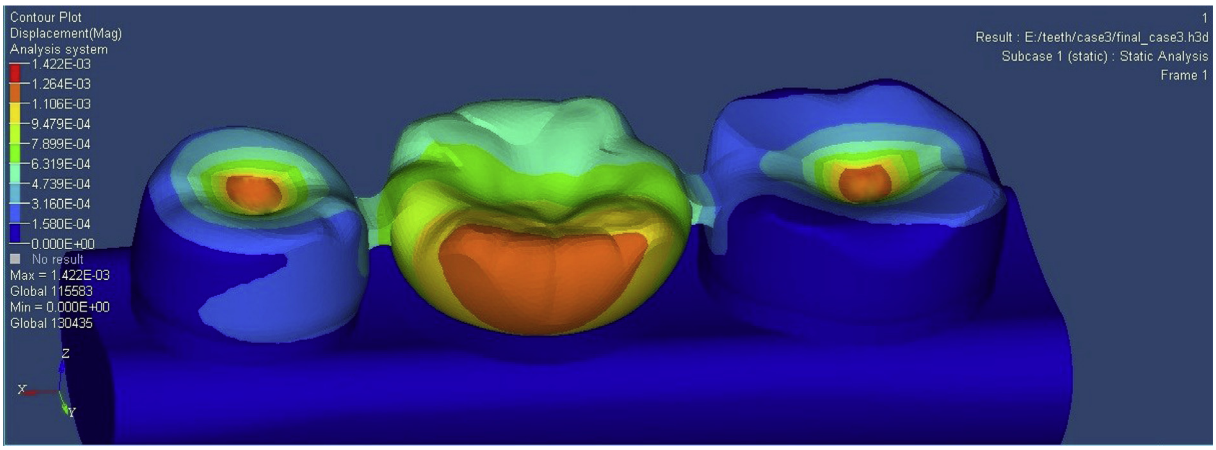


Fig. 6. Group II: Zi-II: Maximum principal stress for Zirconia − 16 $mm^2$.

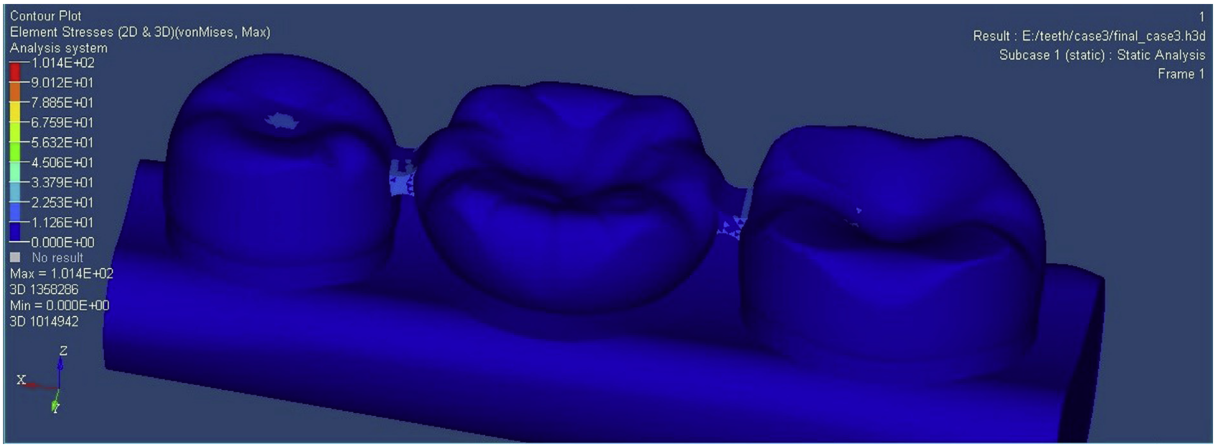


Fig. 7. Group II: Li-II: von Mises stress contours for Lithium-di-silicate − 16 $mm^2$.

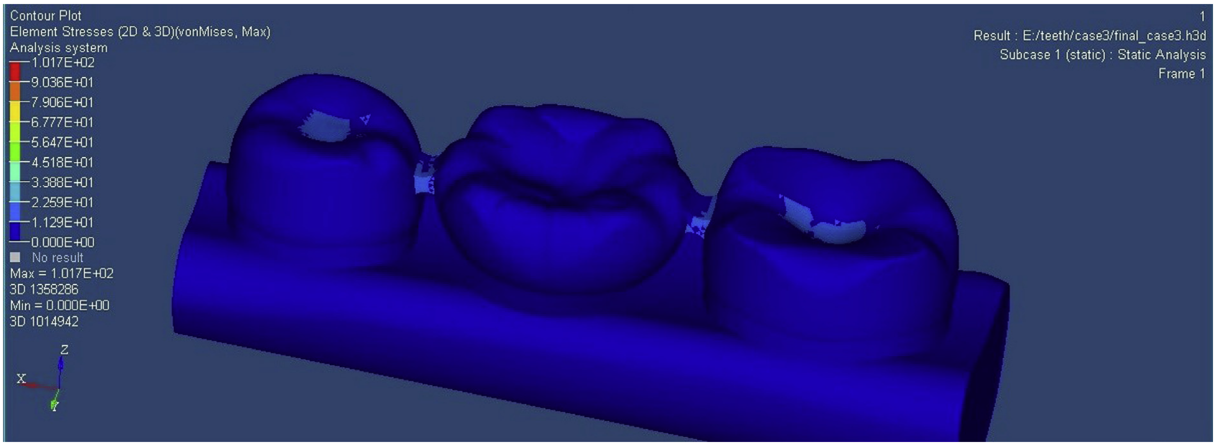


Fig. 8. Group II: Zi-II: von Mises stress contours for Zirconia − 16 $mm^2$.

Furthermore, the connector area is usually narrowly constricted for biological or esthetic reasons, which typically considers stresses relative to the average stress levels in other areas of the prosthesis. The minimal recommended connector cross section area is 12−16 $mm^2$. Previously, it had been hypothesized that fracture initiation sites in dental ceramics could be controlled by changing the ceramic thickness [34].

Several studies have analyzed the stress distributions in Fixed Dental Prostheses. One such study investigated distal cantilevered Fixed Dental Prostheses with differing cantilever morphologies made of two different restorative materials, where the width of the curved connector between the cantilever and primary abutment restoration was 2.25 mm. The average von Mises stress values revealed a higher stress at the occlusal embrasure of the connector between the pontic and second premolar abutment compared to the cervical embrasure [21].

In another study, the occlusal and gingival embrasures of connectors were reported to be the areas of highest stress [23]. In accordance with this, connector dimensions of 9 $mm^2$ and 16 $mm^2$ have been used in this study.

To conclude, the study showed that Zirconia and Lithium-di-silicate demonstrated sufficient stability for replacement of posterior teeth.

## 5. Conclusion

The following conclusions and recommendations may be drawn from the study-

1. Increasing the dimensions of the connector decreases fracture loads.

2. Both Lithium-di-silicate and Zirconia can be used as the material of choice for all ceramic inlay retained Fixed Dental Prostheses.
3. Tensile stresses are concentrated at the gingival aspect of the connector and the vast majority of ceramic failures are initiated at this site.
4. The flexibility of the framework may play an important role in the marginal adaptation of inlay retained Fixed Dental Prostheses and more rigid materials could transfer the stress to the margin to a smaller degree than flexible materials.
5. Long term in vivo studies are insisted to evaluate whether the results presented are transferable to the clinical situation.

## Conflict of interest

The authors hereby declare that have no conflict of interest. They warrant that the article has not received prior publication and is not under consideration for publication elsewhere. This research has not been submitted for publication nor has it been published in whole or in part elsewhere. We attest to the fact that all Authors listed on the title page have contributed significantly to the work, have read the manuscript, attest to the validity and legitimacy of the data and its interpretation, and agree to its submission.

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