A Simulation of Fracture Resistance and Stress Distribution of Endocrown in Different Depth of Pulp Chamber and Modification: Finite Element Analysis

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Abstract

The use of endocrown in conserving the tooth becomes promising strategy due to its ability in covering widely area of coronal damages and residual fragile dentine structure. This study is a computer simulation that was conducted to investigate the fracture resistance and stress distribution from 3D models of endocrown, based on depth pulp chamber (1.5 mm and 2.5 mm). The models were designed in different shapes that were modified into roots canal for different depths for 2 mm and 3 mm. Six 3D models that were designed by AutoCAD software in accordance of external scanning of first molar tooth of mandibula were prepared. The simulation was performed via Ansys 17.2 conducted by loading 225 Newton (mastication force) and 600 N (biting force) within axial degree (0o) and non-axial degree (45o and 90o). The results from the simulation were analysed by Finite Element Analysis (FEA) program with colour contours of von Mises as well as the distribution of the contours. The findings showed that an acceptable fracture resistance in 225 N without the involvement of roots canal were obtained, however, not in higher loads. It can be concluded that the proposed endocrown designs without involvement of roots canal were considered to be acceptable both in mastication and biting force.

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Introduction

The restoration of posterior tooth after endodonthics generally becomes challenging issues among dentists, in which a number of residual tooth structures must be conserved and protected to establish the esthetical features via the shapes and functions. Subsequently, this restoration must also be able to satisfy accordingly to the principal of minimal invasive preparation that is performed to protect the structure of hard tissue of tooth in maximum conditions[1,2,3]. In conventional methods, the restoration that is performed after the treatments on roots canal is conducted by establishing the pegs and cores within the roots. However, this restorative method is considered to have drawbacks, such as the removal of healthy tissue in certain amounts due to the needs of pegs into the roots canal. On the other hand, this procedure is quietly affected the entire biomechanical features of restored tooth[4,5]. Consequently, a possibility of failure that is classified as favourable and catastrophic fractures could have been occurred, which are started by the presence of microcracks as an initiator; given its continuous incidents to be developed into chronic conditions, it must be restored by removing the tooth and requires prosthetic replacement[2]. The fracture resistance during a restoration depends on several factors, including the modulus elasticity of supporting tissues, the materials for restoration, the luting material characteristics, the thickness of restoration, and the designs of prepared tooth[3].

Recently, the endocrown becomes one of the alternatives of modern tooth treatments. This is due to its combined strategies between intra coronal restoration (inlay/onlay) and full coverage extra coronal, which specifically have been used as selective clinical restoration on posterior non-vital with small amounts of tooth structure residue due to its acceptable biointegration, esthetics,

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and functions[6]. The endocrown that was introduced by Bindl and Mormann in 1999, is a non-vital restoration method by taking advantages of pulp chamber as a retention[7]. However, this method has been reported to obtain different results based on the findings that the combination of macro retention into roots canal for 3 mm was found to be acceptable with minimum probability of failures[8]. Furthermore, regarding to this finding, no investigations have been conducted about the ability of endocrown prepared with retention of modification and its preparation into roots canal.

The aim of this study is to investigate the fracture resistance and shear distribution from six 3D models of endocrown prepared with different depths of pulp chambers as well as varied retention designs, that were simulated via Ansys software.

Materials and methods

This study was carried out via CAD modelling systems from the scanning results of first molar of mandibula, which was in form of external models. Then, the shape models were reconstructed in term of the solid materials by using AutoCAD software (AutoCAD 2016). The obtained external shapes were then simulated via Ansys program (Ansys 17.2 Inc., Houston, TX, USA) which was used to create the 3D models. The volumes and sizes of the models were prepared accordingly from the references of molar mandibula, which were previously conducted by Wheeler (2015)[9].

The solid model of samples was prepared by connecting to the occlusal surface area cross-sectionally. The cavity of mesial-occlusal-distal-palatal (MODP) was reconstructed on the models with the depths of 2 mm on the cement enamel junction which is demonstrated in the following Figure 1. The cavity was designed to be 1.5 mm from the margin with the depth of pulp chambers for 1.5 mm and 2 mm. The angle of sloping wall in the pulp chambers was 6°. The periodontal ligaments were reconstructed with the thickness of 0.2 mm, while the preparation of cervical margin which is shaped into circumferential butt joint surrounded by tooth cervices with 1.5 mm of width and 2 mm of height from the cervices. The structures were assumed to be linearly elastic, isotropic, and homogenous to simplify and overcome the computation parameters. The six models were fabricated in exact dimensions, excepts for the depth of pulp chambers which were designed variously with the depths of 1.5 mm and 3 models for 2.5 mm of depths. Every model with proposed depths of pulp chambers were prepared into: 1 conventional model without modification into the roots canal, 1 model for 2 mm of modification into the roots canal, 1 model for 3 mm of modification into the roots canal.

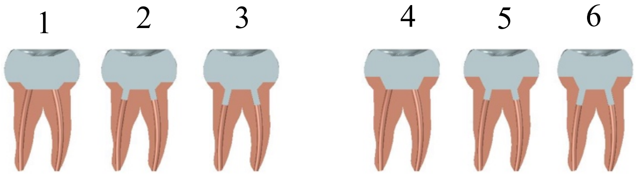

Figure 1. The six solid 3D models with varied depths of pulp chamber and their modifications into the root canals.
Model 1, 2, and 3: Endocrown for the 1.5 mm of pulp chamber depth without and its modification 2 mm and 3 mm of roots canal.
Model 4, 5, and 6: Endocrown for the 2.5 mm of pulp chamber depth without and its modification into 2 and 3 mm of roots canal.

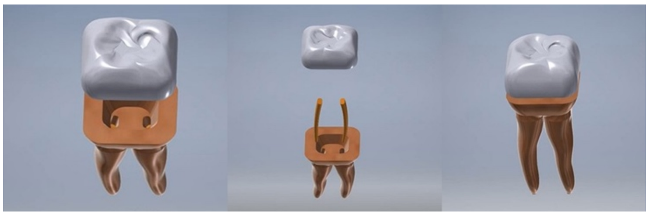

Figure 2. The 3D models prepared by three separated parts (endocrown, gutta percha, and root).

The coronal models of endocrown were prepared in accordance to the shape of crown model of tooth, which were designed to be limited to the cervical areas. Then, six models of endocrown were set up into in 2-dimension with varied depths of pulp chambers, i.e. three models for dimension of 3.7 mm x 4 mm x 1.5 mm, whereas 3 models for 3.7 mm x 4 mm x 2.5 mm. Two models from each depths of pulp chambers were modified with the enlargement of retention into the roots canal for 2 mm and 3 mm of depths, with diameter of 1.3 mm in oval shapes. The solid model of molar was prepared by combining the restorative solid model, which was attached via cement layers with 1 mm of thickness between the surface on internal restoration and tooth models. In between these two models, certain amounts of gutta-percha within both of the first molar of root canals were prepared for each canal. The following Figure 2 displays the attachment mechanisms.

The 3D models that have been designed via AutoCAD software were imported into the simulation software of Ansys to be analysed via static structural in 6 steps. Firstly, the Engineering Data that was used to input the mechanical properties of every model of material component, and then the Geometery that was carried out to input the model images, which was continuously analysed. Next, the Model which was conducted to obtain meshing model, and then the Setup, that was used to input the load parameters as well as the support. Afterward, the Solution which was set up to input the parameter of proposed results, and lastly, the results that was aimed to obtain the results of analysis. The material properties of tissues, bones, and restoration treatments were selected in accordance of standard uses, which are displayed by the following Table 1.

Material	Youngs' Modulus (MPa)	Poisson Ratio	Tensile Strength (MPa)	References
Dentin	18. 600	0,31	105,5	Binwen, 2015
Cementum	18.600	0,31	105,5	Desai, 2012
PDL	50	0,49	-	Desai, 2012
Cortical Bone	13.700	0,30	-	Prina, 2016
Gutta-percha	140	0,28	155,8	da Fonseca, 2018
Resin luting cement	8.300	0,35	45,1	Dejak, 2013
Lithium disilicate reinforced glass ceramic	95.000	0,35	360	da Fonseca, 2018

Table 1. The material properties prepared in the six 3D models of endocrown in FEA simulation.

The division of 3D complex geometry via the mesh to the all models in determining the numbers of elements and notes were carried out automatically in the Ansys (the number of nodes = 930837 and the numbers of the elements = 624659), so that the complete simulation of every model was obtained. The maximum and minimum stress values were evaluated via colorimetry graph for each element and solid tetrahedral elements with certain numbers of elements and nodes.

The setup stage in which every model was assumed to be isotropic, homogenous and elastic was loaded by external loads which were compressive strengths on the entire surface of endocrown for 225 N (representing the force of normal mastication) and 600 N (representing the maximum biting, which was prepared in axial and non-axial angels: 0° to the vertical area, 45° to the oblique non-axial area, and 90° to the horizontal non-axial. Consequently, the simulation was set up to obtain real conditions within the oral. The support step was assumed to be elastic in the periodontal ligaments.

The obtained results of simulations were observed to find the values and colour contours that were plotted in von Mises graphs to confirm the total deformations, equivalent plastic strain, and equivalent of von Mises stress as the parameters in displaying the fracture resistance and shear distribution characteristics. The colour contours plotting that were displayed confirmed the value numbers of shear/ stress, in which the red colours were the maximum peaks and the blue colours were the minimum peaks.

Results

The 225 N of loads that were performed in the simulation showed the highest maximum deformation in the endocrown design of 1.5 mm of a pulp chamber depth, no modification to the root canal, and a load of 45° angle. Meanwhile, the 600 N of loads, the highest maximum deformation was found at a pulp chamber depth of 1.5 mm with a modification of 2 mm to the root canal with a load of 45° angle (Table 2). The lowest maximum deformation at a load of 225 N was observed at a pulp chamber depth of 2.5 mm with a modification of 3 mm to the root canal with a load of 0° angle, while in the load of 600 N, the lowest maximum deformation is at a depth of 1.5 mm with a modification of 3 mm to the root canal with a load of 0° angle. Additionally, the highest and lowest maximum equivalent elastic strains was given a load of 225 N and 600 N (Figure 3 and 4).

Endocrown 3D Model	TOTAL DEFORMATION (mm)					
	Load 225 N			Load 600 N		
	0°	45°	90°	0°	45°	90°
Model 1	0,24818	0,38652*	0,32037	0,66181	1,03070	0,85432
Model 2	0,24811	0,38635	0,32016	0,66163	1,16740*	0,85375
Model 3	0,24797	0,38611	0,31979**	0,66126**	1,02960	0,85278**
Model 4	0,24826*	0,37902	0,33410*	0,66203*	1,01070	0,89094*
Model 5	0,24819	0,37884	0,33385	0,66183	1,01030	0,89027
Model 6	0,24084**	0,37861**	0,33343	0,66145	1,00960**	0,88913

Table 2. The comparison of total deformation of 6 endocrown designs based on all loads with axial (0°) and non-axial angles (45° and 90°).

(highest value*, lowest value **)

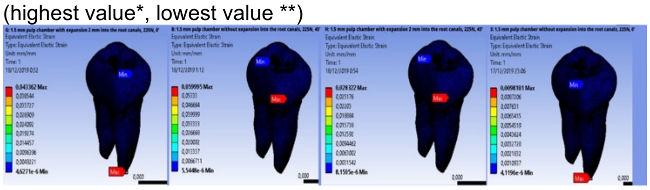

Figure 3. The highest and lowest maximum equivalent elastic strains given a load of 225 N and 600 N.

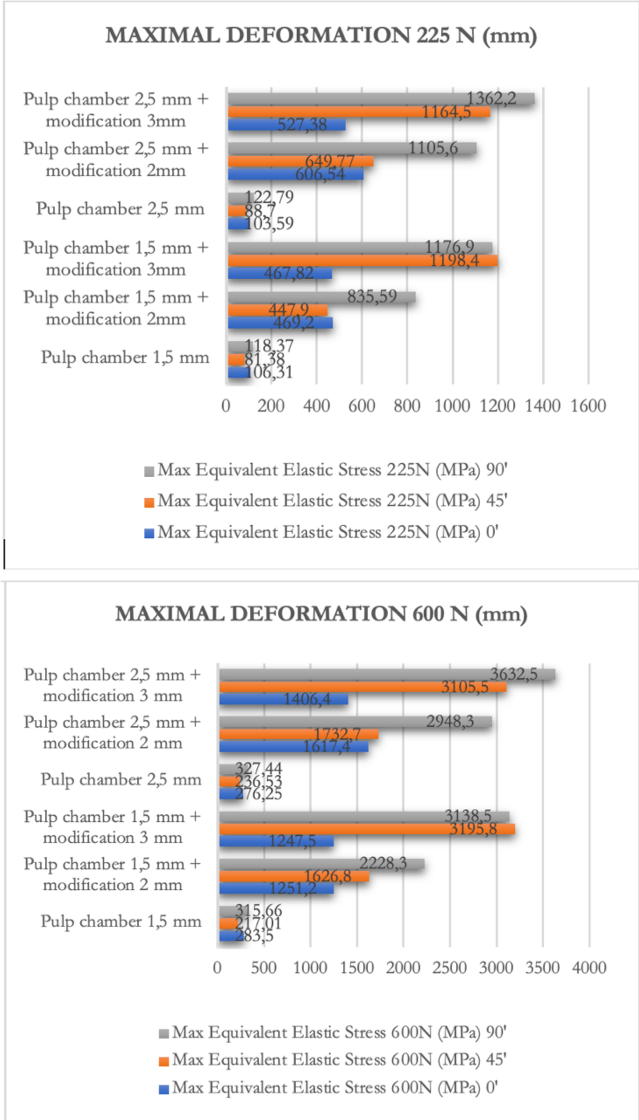

Figure 4. The maximum equivalent elastic strain of 6 endocrown designs with different pulp chamber depths given a load of 225 N and 600 N.

In regard to the lowest maximum equivalent elastic stress, the pulp chamber depth of 1.5 mm without the modification to the root canal with a load 45° angle was found to be the lowest, and at the load of 600 N, the maximum equivalent elastic stress value was the lowest at the pulp chamber depth of 1.5 mm without modification to the root canal with a load 45° angle (Fig 5 and Fig 6).

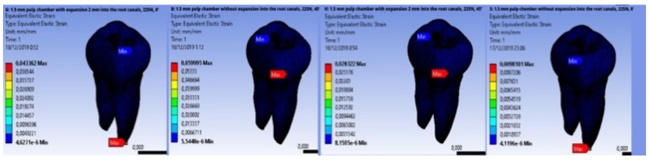

Figure 5. Equivalent maximum and lowest maximum elastic strain given a load of 225 N and 600 N.

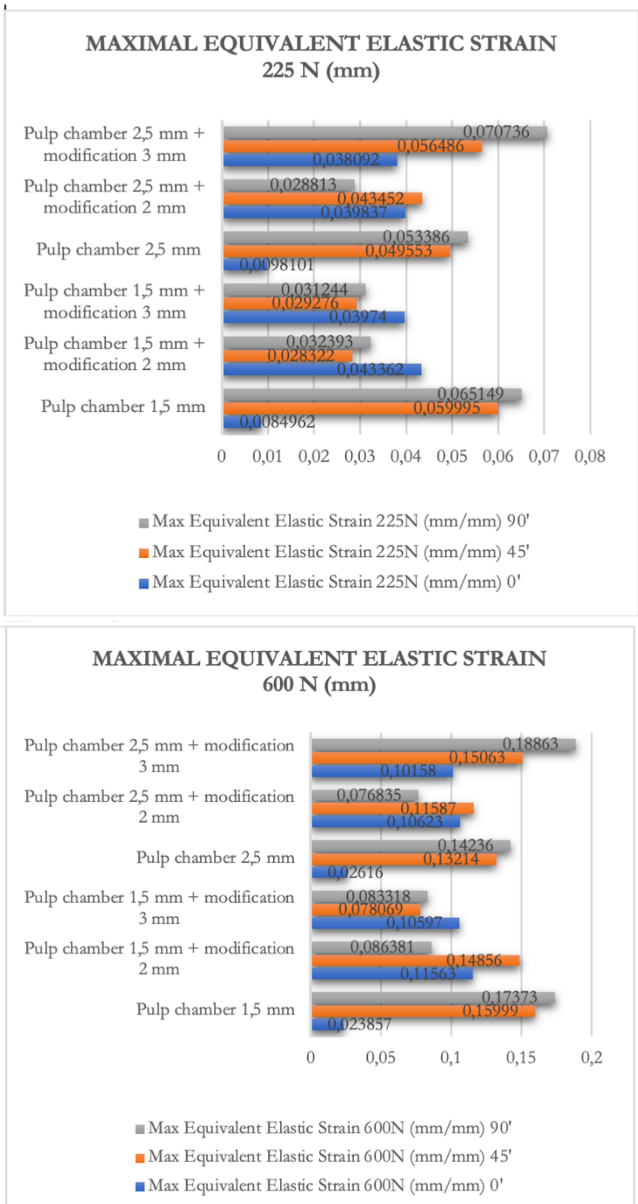

Figure 6. Graph of the maximum equivalent elastic stress of 6 endocrown designs with different pulp chamber depths given a load of 225 N and 600 N.

Discussion

Teeth that have been treated endodontically have a high risk of fracture. There are two factors that can cause the fractures, and they are (1) the stress concentration that potentially cause a crack; (2) the stress that is large enough to cause plastic deformation at the end of the concentration in macroscopic part. It has been reported that the results of an in vitro fracture resistance test from a tooth that has been extracted, and it was found that there was a relationship between the concentrated area of stress and the location of the fracture on the tooth. The deformation which is a response to mechanical strength in form shape changes could be elastic, in which the material has the ability to return to its original size and shape, or inelastic (permanent changes)[10,11]. In elastic deformation, the strain experienced by the material is linearly related to the applied stress, and this phenomenon is explained Hooke's Law. Deformation, stress, and strain that occur in a material after being addressed with a biomechanical load could generate fracture that is observed via FEA simulations. Thus, fracture resistance and stress distribution within the design can be observed as well as predicted in the area, in which the starting point of fracture in a structure or material is initiated in the concentrated area[12].

The simulation results have shown that the designs without involving the root canals or the designs without modification to root canal have a higher maximum deformation value. By contrast, the designs which is modified to the canals with up to 3 mm have the lowest maximum deformation values. Similar results have been reported that the endocrown designs with involving the root canals up to 3 mm had higher probability of failure compared to that without engaging the root canals. The modification of macromechanical retention that its form extends into the root canal resembles into a short post, and this modification is constructed in a cylindrical shape, connected to 1 mm thick-cement-and-gutta-percha as a filling from the root canal up to the apical[8]. The mechanical stress could experience internal and external pressures, exerted on the cylindrical structures. Consequently, the amount of load would be distributed on the structure wall in a circular (perpendicular to the tooth axis), which will continue axially to the end of the cylinder[13]. This theory can be adopted to explain that the chewing and biting force, received by the endocrown will be transmitted apically through gutta-percha, which is cylindrical and elastic. The stress distribution through the gutta-percha can reduce the stress that is concentrated in the coronal area.

The simulation results also have suggested that deeper pulp chamber could reduce the fracture resistance. As the cross-sectional area of the restorative material fills the pulp chamber deeply, smaller amounts of stress are received by the restoration, in which this phenomenon follows the equation of $\sigma = N/A$. The stress concentration is inversely proportional to the height of the restoration, in which given that the restoration is higher, then the less compressive stress is applied to the remaining dentin[14]. This also explains the mechanisms of higher maximum equivalent elastic strain value for the endocrown sample, designed with the involvement of root canals. The increase in the volume of material used in the modified endocrown design is due to the expansion of the design to the root canal, so that the pressure or force that can be absorbed by the material is relatively higher. As the volume of the material used for a restoration increases, the material would receive more stress which also can reduce the transmitted stress from the tooth tissue. Thus, the fractures risk of the remaining tooth structure would be reduced.

Endocrown that was modified to the root canals was simulated successfully at normal functional loads (chewing force of 225 N), but it was insufficiently simulated to withstand higher loads, such as at 900 N of occlusal load. The pulp chamber which is 2 mm to 2.7 mm failed to withstand at a load of about 675 N and 90% of the specimens showed fracture[15]. However, the load applied to this study was much lower than that a literature which reported the failure during a load of 2606 N in the endocrown with 1.5 mm of pulp chamber depth[16]. Likewise, the endocrowns with 2 mm of pulp chamber depth had failure resistance at a load of 1368 N[17].

The angle during the administration of loads has a significant effect on the value of the deformation that occurs. As this study only performed the mastication and biting force that resemble common condition and situation within the mouth, various angles were selected, which were 0°, 45°, 90° angle. The 0°, also known as the vertical force, have been reported to have a smaller deformation value, compared to the 45° (oblique force) and 90° (horizontal force) angles[18]. The deformation and stress distribution of implants may have occurred due to symmetrical distribution of deformation to the tooth axis, and the stress distribution would be decreasing at the centre of the tooth axis[19]. In addition, the vertical forces to the occlusal would be apically transmitted via photoelastic media[20]. It has been reported that the mastication and biting force within the endocrown would be transmitted apically through gutta-percha due to its cylindrical and elastic shape, so the stress concentration in the coronal area (pressure point), especially at the edge of the pulp chamber or in the cervical region can be considered as the initial point of crack which is mostly beings at the location of the stress concentration[21].

The simulation results from the load of 600 N were almost the same as a load of 225 N, in which at the angle of 0°, the maximum elastic strain value was located in the apical region with gutta-percha. In the applying load at 45° angle, the maximum elastic strain value was located in the proximal area, except for the designs with 2 mm of modification to the root canal. Whilst, the applying loads at 90° angle, all designs showed that the maximum elastic strain value occurred in the proximal area. These results have indicated the direction of force or angle based on every load, and the positions are related to the equivalent elastic strain/ stress distribution. There was a symmetrical distribution of deformations to the tooth axis, and the stress distribution would decrease at the centre of the tooth axis. The 0° angle is considered to be symmetrical with the tooth axis, so the deformation and stress that were received by the tooth were transmitted to the abutment or the root of the tooth the apical[19]. Given that the gutta-percha in the root canal distributes to the deformation and stress to the apical area, then no stress concentration would be occurred which lead to the increase of fracture resistance. The pressure received at the apical part is related to the elastic modulus of the type of restorative materials used. Higher modulus elasticity of the restorative material than the modulus of elasticity of dentin, will result in the stress concentration being distributed into the root of canal system[22]. Similar results was also reported that the highest stress concentrations were located in the apical region in all design models, and an increase stress concentration was found in the region around the cervix[23]. The design of the cervical margin also affects the fracture of restoration. Endocrown with a characteristic of cervical margin design within the butt joint is designed to achieve a wide and stable surface, bonding optimisation, and to provide a resistant ability to masticate stresses on molars. Hence, the surface is prepared parallel to the occlusal plane to ensure stress resistance along the main axis of the tooth[24,25].

The modulus of elasticity of the material, the magnitude of the load force applying, and the angle of loads have influences on the simulation results. They are the value of total deformation, equivalent elastic strain, equivalent von Mises stress as an indicator of fracture resistance, and stress distribution, which is experienced by a design after loading. Although FEA is a good and easy research method, it is necessary to carry out further tests on the variation of endocrown designs that include experimental tests to obtain more accurate results.

Conclusions

The simulation results have shown that an increase in the depth of pulp chamber in the macro retention of the design can increase the fracture resistance of the tooth after endodontic treatments due to its ability in reducing the deformation value of the model. Deeper pulp chamber and wider cross-sectional area of the restorations have been predicted via the simulation to increase the fracture resistance of a restoration. The results of this study have indicated that at the load of 225 N, the retention modification of the endocrown restoration design to the root canal had no significant difference in the term of resistance compared to the conventional endocrown design which had involvement to the root canal. However, at a higher load (600 N), the endocrown design that has no modification to the root canal showed the lowest deformation, indicating a better ability to withstand the fracture resistance compared to that from the modified design to the root canal of endocrown. The load that was given axially with 0° is the angle of the load with the lowest risk of fracture, whereas the 90° non-axial angle is the angle with the highest risk of fracture. However, it is necessary to do in vitro clinical trials in the form of experimental tests on the real models, so the designs can be compared to the results of FEA, considering that FEA should be used in multiple tests.

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Declaration of Interest

The authors report no conflict of interest.

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