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ORIGINAL ARTICLE

Effect of Preparation Taper, Height and Marginal Design Under Varying Occlusal Loading Conditions on Cement Lute Stress: A Three Dimensional Finite Element Analysis

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Abstract To assess the effect of preparation taper, height and margin design under different loading conditions on cement lute stress. A 3-D FE model of an upper second premolar and molar was developed from CT scan of human skull using software programmes (MIMICS, Hypermesh and ANSYS). 10° and 30° taper, 3 and 5 mm preparation height and shoulder and chamfer finish lines were used. Type 1 Glass ionomer cement with 24 µm lute width was taken and the model was loaded under 100 N horizontal point load, vertical point load distributed axial load. The maximum shear stress and Von Mises stress within the cement lute were recorded. The maximum shear stresses ranged from 1.70 to 3.93 MPa (horizontal point loading), 0.66 to 3.04 MPa (vertical point loading), 0.38 to 0.87 MPa (distributed loading). The maximum Von Mises stresses ranged from 3.39 to 10.62 MPa (horizontal point loading), 1.93 to 8.58 MPa (vertical point loading) and 1.49 to 3.57 MPa (distributed loading). The combination of 10° taper and 5 mm height had the lowest stress field while the combination of 30° taper and 5 mm height had the highest stress field. Distributed axial loading shows least stress, better stress homogenization and gives a favorable prognosis for the fixed prostheses. Smaller preparation taper of 10° is biomechanically more acceptable than a 30° taper. It is desirable to decrease taper as height increases. The chamfer margin design is associated with greater local cement stresses toward the margins that could place the cement at greater risk for microfracture and failure.

Keywords Finite element analysis · Cement lute · Taper · Height · Finish line · Occlusal loading

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Introduction

The patients' need in the area of dental prostheses is a restoration that closely replicates the original natural tooth in both function and appearance [1–3]. Many studies have suggested that the average life of crown and bridge is between 10 and 15 years. However, these restorations do eventually fail and the common causes are due to lack of retention and/or caries [4–7]. Factors such as degree of convergence of the opposing walls of the preparation, the preparation height and width (and hence the height to width ratio), the preparation surface area and the surface roughness have an important role in achieving desired resistance and retention form for an extracoronal restorations [8–16].

It is anticipated that crowns within the mouth are subjected to functional and parafunctional forces of varying magnitudes and directions [17, 18]. These forces can generate stress in the luting agent. Therefore, the integrity of the cement lute and how this responds to occlusal loading during eating, swallowing and parafunctional activity is critical to the success of crown and bridgework [19].

A clinical evaluation of luting agents is not easy because much of the material is hidden by the seated restoration. Studies have shown that failure occurs within the cement layer instead of the interfaces between cement and tooth or restoration. Cement fractures lead to microleakage and changes in stress distribution to supporting tissues [4]. Although various researchers have determined failure stress for different cements available, little work has been carried out to evaluate actual stresses existing in the cement layers.

Various investigators have reported the use of photoelastic methods to analyse stress distributions in metal ceramic crowns [20–24]. Though general design concepts were focused upon but the effect of stress development in the cement layer was not considered in these studies. Apart from being tedious and time-consuming, the accuracy of the results is affected by the observer's ability to achieve proper identification of the fractional fringe orders. Though the results of photoelastic analyses are a useful adjunct to finite element stress analysis, the computerized stress analysis approach can handle a greater variability in the material properties and is an extremely accurate approach [25].

FEA has been used to investigate many aspects of crown preparation including cement microfracture, luting cement thickness and physical properties, the influence of margin design on cervical stresses and the influence of crown metal thickness on cervical stresses [19, 26–34]. However the interplay between preparation taper, preparation height, and marginal configuration on the stresses developed within the cement lute has not been investigated. Therefore a three dimensional finite element stress analysis study was planned from a comprehensive perspective to assess the influence of preparation taper, height and margin design under different loading conditions on luting cement.

Materials and Methods

The study was conducted at M R Ambedkar Dental College and Hospital, Bengaluru, Karnataka, India. A Computed Tomography scan of human skull was used as a reference to model the geometry of maxillary second premolar and first molar. The images were recorded in Dicom format. Using MIMICS (Version 8.11), the Dicom format of the images was converted to Initial Graphics Exchange Specification format (IGES) leading to the generation of the geometric model. The geometric model was then converted into finite element model by using software called Hypermesh (Version 9.0). ANSYS (Version 12.0) was the solver used to do the analysis of the present study.

The model comprised of a buccolingual cross-sectional view of maxillary second premolar and first molar. Each model for both premolar and molar had a crown height of 5 mm (long preparation) and 3 mm (short preparation). Both the 'long' and the 'short' preparations were further modified to give a total preparation taper (convergence angle) of 10° and 30° (Fig. 1a). The thickness of porcelain was maintained between 1.2 and 1.4 mm. The thickness of the alloy was kept at 0.3 mm. The final representative values for porcelain fused to metal crown are depicted in Fig. 1b. The integrity between ceramic veneer and metal coping were defined using a 'glued contact option' within the software. Both the models were prepared with a shoulder margin on buccal side and chamfer margin on the palatal side and restored with porcelain fused to metal crown. The cement chosen was type 1 glass ionomer cement and the film thickness of the cement layer was kept 24 µm wide.

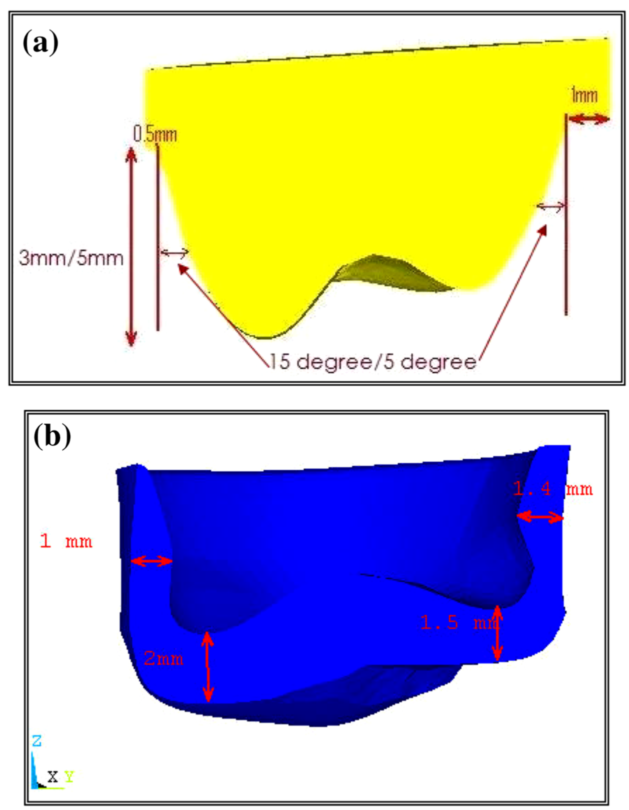

Fig. 1 Representation of properties specified to the finite element model

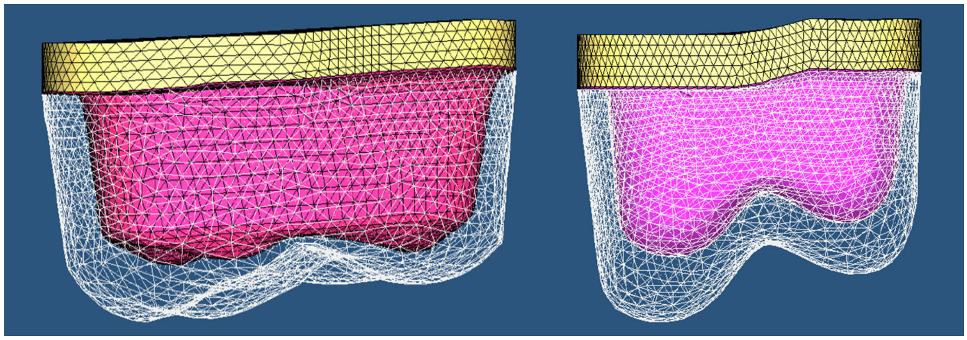

Fig. 2 Three dimensional meshed FE model of maxillary second premolar and maxillary first molar

Table 1 Number of elements and nodes for each model

Tooth type	Preparation length (mm)	Preparation taper (°)	Nodes	Elements
Premolar	3	30	12,428	46,042
Molar	3	30	12,328	45,111
Premolar	5	30	12,428	46,042
Molar	5	30	12,328	45,111
Premolar	3	10	12,885	47,675
Molar	3	10	13,096	47,790
Premolar	5	10	12,885	47,675
Molar	5	10	12,885	47,790

Using Hypermesh Pre-processor (Version 9.0), the 3-D finite element model corresponding to the geometric model was meshed and the model was divided into large number of elements and nodes (Fig. 2). An overview of the meshes used in study and the number of elements and nodes for each model is given in Table 1.

For the accurate analysis of the problem and interpretation of the results, Young's modulus of elasticity and Poisson's ratio were utilized. The corresponding values of Young's Modulus (E) and Poisson's ratio (ν) of dentine was chosen as 20 GPa and 0.31, glass ionomer cement as 7.5 GPa and 0.35, porcelain as 80 GPa and 0.30 and Ni–Cr alloy as 172 GPa and 0.32 respectively according to literature survey [26].

Before applying the boundary conditions, the system of equations is not completely defined. This is because, any model which is generated, has to be constrained depending upon the requirements of the study. Thus, boundary conditions are applied to have enough fixed nodal displacements to prevent the structure from moving in space as a rigid body when external loads are applied. The contact between the cement layer, the tooth and crown were described using the 'bonded contact' option. This created tying equations that described the contact, both normal and tangential to the contact segments. This contact option allowed specifications to be entered regarding the circumstances at which separation at the interfaces would take place. The separation force was set to a very high level (>500 N) in order to ensure that no separation could occur. The finite element model was rigidly constrained in the x, y and z directions at the base of the model. In clinical reality this set of boundary conditions is likely to represent a crowned tooth that has been loaded by a relatively high occlusal load causing the tooth to 'bottom out' in the alveolar housing.

As this study was concerned primarily with the cement lute stress and the influence of preparation geometry, modelling of the supporting tissues and alveolar bone was omitted. Similarly, the effect of the pulp was considered negligible due to its comparatively low volume and the comparatively high stiffness characteristics of the surrounding dentine and cement. Furthermore, Rubin et al. [32], Morin et al. [33] and Anusavice and Hojjatie [35] have shown that modelling the pulp as a void has no effect on the magnitude of the coronal stress field. The modelling of root was omitted as it was found that the root experienced minor stresses as compared to the coronal portion of the tooth. It was assumed that all materials were homogenous, linearly elastic, and isotropic and perfect bonding existed between all layers.

An attempt was made to simulate actual clinical situation. The magnitude of normal masticatory forces ranges from 9 to 180 N (2–40 lbs) with a duration of from 0.25 to 0.33 s [36]. The finite element models were loaded in three ways: (i) Point load of 100 N applied horizontally to the top edge of palatal cusp of maxillary 2nd premolar and mesio-palatal cusp of maxillary 1st molar (Fig. 3a). (ii) Point load of 100 N (Fig. 3b) applied vertically on the occlusal surface of palatal cusp of maxillary 2nd premolar and mesio-palatal cusp of maxillary 1st molar. (iii) Distributed load of 100 N (Fig. 3c) across the entire occlusal surface acting apically.

After applying load on each model, a record of the patterns and values of maximum shear stress and maximum Von Mises stress within the cement lute was displayed using color coded figures. The shear stress was chosen as this would be greater numerically than the local tensile or compressive stresses. The Von Mises stress was chosen as

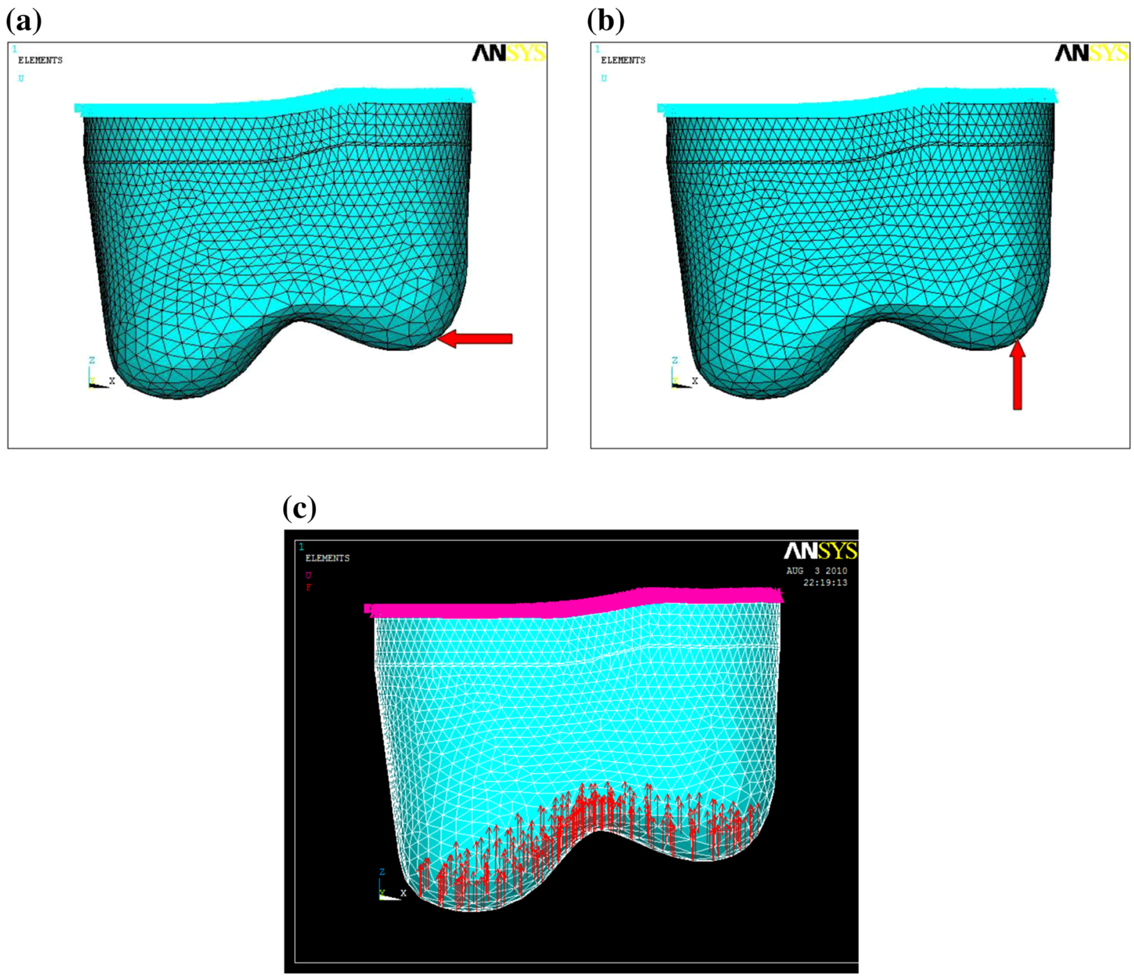

Fig. 3 Representation of loads specified to the finite element model. **a** Point load of 100 N applied horizontally, **b** point load of 100 N applied vertically, **c** distributed load of 100 N across the entire occlusal surface

it provides an overall picture of the stress field, since it contains components of both the local tensile and shear stresses. Both of these stress parameters are likely to predict failure.

Results

The maximum shear stress and maximum Von Mises stress within the cement lute were calculated and represented in color-coded figure. (Fig. 4; Tables 2, 3). A representative stress plot is depicted in Figs. 5, 6, 7, 8 showing the maximum Shear stress and Von Mises stress in the cement lute for each combination of tooth type, taper and preparation height. For loadcase-l, which depicts a lateral force applied near the occlusal surface of the crown, the maximum shear stresses ranged from 1.70 to 3.93 MPa, while

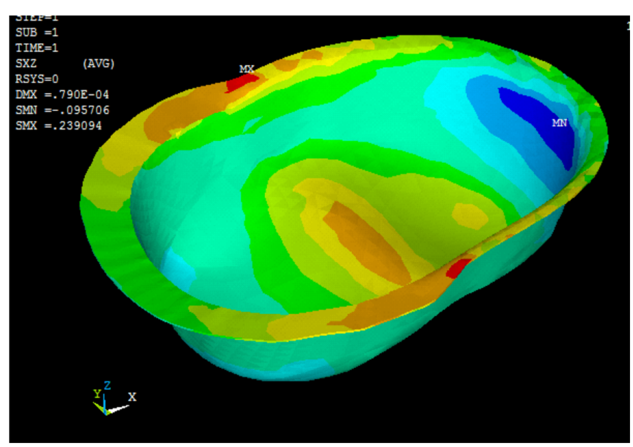

Fig. 4 Shear stress contours in premolar with 3 mm height and 10° taper under point load of 100 N applied horizontally to the top edge of the crown

Table 2 Maximum shear stress and maximum Von Mises stress in premolar under different loading conditions

Height (mm)	Taper (°)	Shear stress in horizontal point load (MPa)	Von Mises stress in horizontal point load (MPa)	Shear stress in vertical point load (MPa)	Von Mises stress in vertical point load (MPa)	Shear stress in distributed load (MPa)	Von Mises stress in distributed load (MPa)
3	10	2.39	6.39	1.69	4.95	0.48	1.62
3	30	2.52	7.53	1.94	6.31	0.52	1.63
5	10	1.70	3.39	0.66	1.93	0.44	1.55
5	30	3.40	8.75	3.04	6.64	0.87	3.57

Table 3 Maximum shear stress and maximum Von Mises stress in molar under different loading conditions

Height (mm)	Taper (°)	Shear stress in horizontal point load (MPa)	Von Mises stress in horizontal point load (MPa)	Shear stress in vertical point load (MPa)	Von Mises stress in vertical point load (MPa)	Shear stress in distributed load (MPa)	Von Mises stress in distributed load (MPa)
3	10	2.61	8.48	2.37	5.77	0.57	1.54
3	30	3.00	9.81	2.82	7.37	0.62	2.17
5	10	2.47	6.36	2.36	5.63	0.38	1.49
5	30	3.93	10.62	2.83	8.58	0.79	2.29

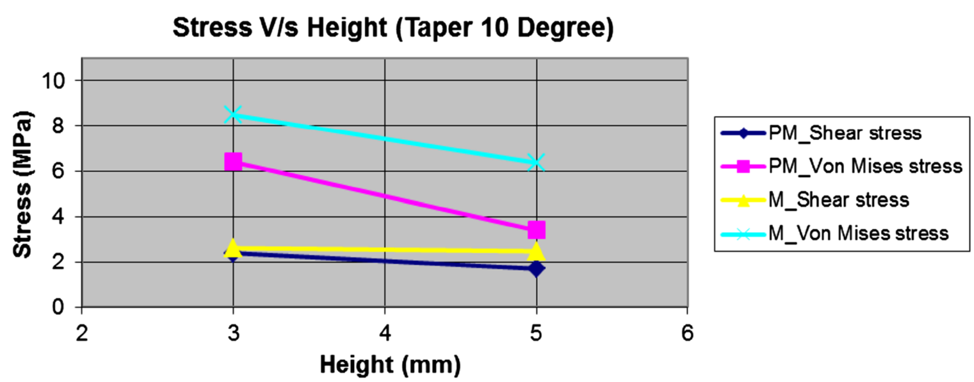

Fig. 5 Variation of maximum shear stress and Von Mises stress with height in maxillary second premolar and maxillary first molar at 10° taper

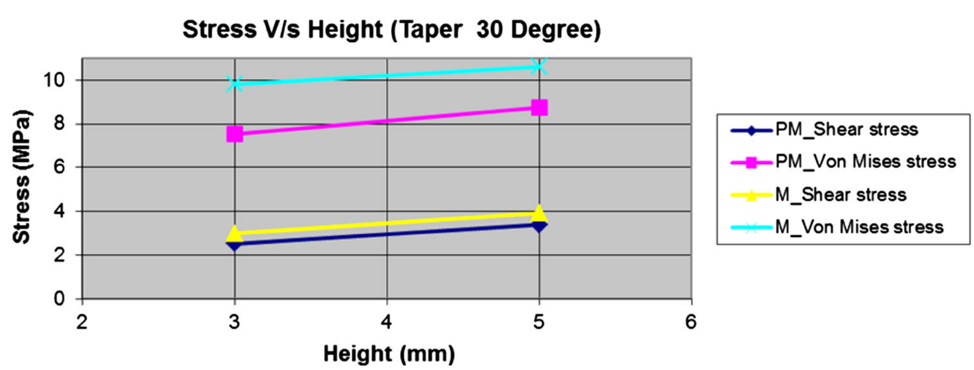

Fig. 6 Variation of maximum shear stress and Von Mises stress with height in maxillary second premolar and maxillary first molar at 30° taper

the maximum Von Mises stresses were in the range 3.39–10.62 MPa. For loadcase 2, which depicts an apically directed occlusal load, the maximum shear stresses were in the range 0.66–3.04 MPa, while the maximum Von Mises stresses were in the range 1.93–8.58 MPa. For loadcase 3, which depicts the load applied by a food bolus, the maximum shear stresses were in the range 0.38–0.87 MPa, while the maximum Von Mises stresses were in the range 1.49–3.57 MPa.

The stress fields in the cement lute were consistently highest under 100 N horizontal Point load followed by 100 N vertical Point load and least under 100 N

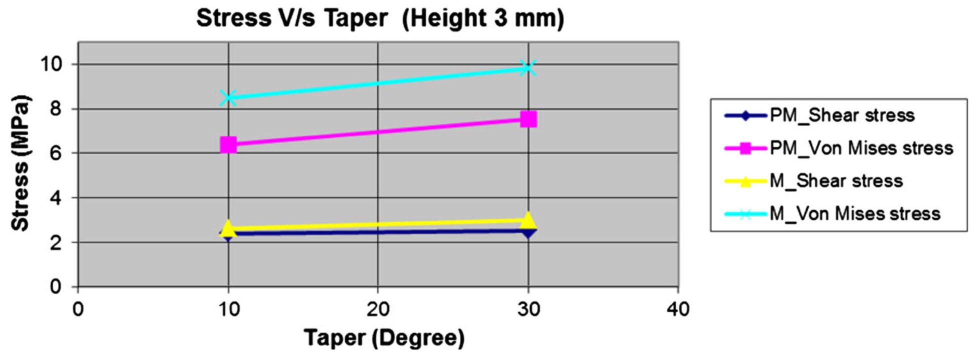

Fig. 7 Variation of maximum Shear stress and Von Mises stress with taper in maxillary second premolar and maxillary first molar at 3 mm height

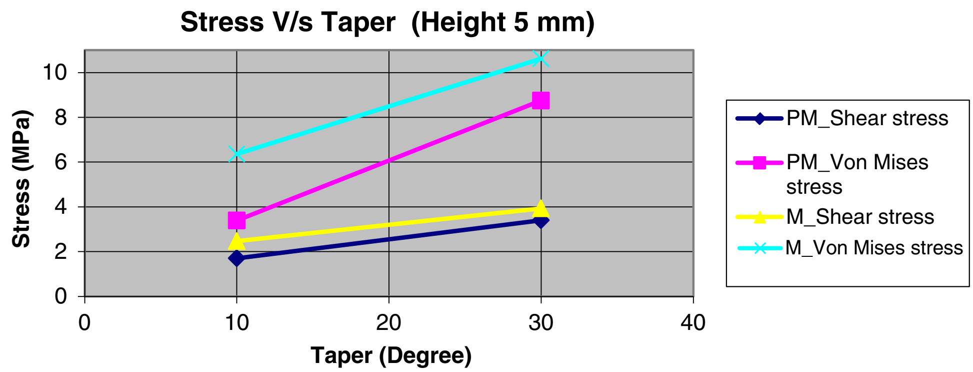

Fig. 8 Variation of maximum shear stress and Von Mises stress with taper in maxillary second premolar and maxillary first molar at 5 mm height

Distributed load for both premolar and molar. In comparison to the 10° taper, the stress fields in the cement lute were consistently higher in premolar and molar with a 30° preparation taper under all three loading conditions. The combination of 10° taper with a long preparation (5 mm) had the lowest stress field and the combination of 30° taper with a long preparation (5 mm) had the highest stress field under all three loading conditions. The stress fields were higher in the chamfer region and at the junction of shoulder and chamfer region in comparison to the shoulder region.

Discussion

The present study investigated the cement lute stress based on preparation height, taper and margin design under different occlusal loading conditions using three dimensional FEA rather than the two dimensional FEA method because it would more accurately model the complex stress distributions encountered clinically and give an actual representation of the stress behaviour in the cement lute. Also the two-dimensional models have fewer elements and structural details and may lead to incorrect interpretation due to higher stress values. In comparison to clinical trials, which could have been complicated and expensive owing to an array of factors affecting microleakage under artificial crowns, advanced computer modeling techniques allow virtual manipulation of clinical variables relating to cement microfracture [26].

Maxillary premolar and molar were chosen for the investigation as various FEA studies have been conducted on the same for stress analysis [19, 26–34]. Also, Trier et al. [37] found that over 95 % of all castings that failed by becoming uncemented lacked resistance form and amongst that 63 % of the failures were molars, 35 % were premolars, and 2 % were anterior teeth. The source of data for the investigation was taken from a CT scan of a human skull to precisely simulate the dimensions and shape of the maxillary second premolar and first molar.

The cement lute width was kept 24 µm wide in accordance to the study conducted by Fusayama and Iwamoto [38] who advocated an optimum cement thickness between 30 and 50 µm. Further, Jorgensen and Esbensen [39] stated that increase in film thickness from 20 to 140 µm decreased the retentive strength of crowns by approximately 33 %. Likewise, study conducted by Mayhew et al. [40] indicated that cast gold crowns cemented with a film thickness of 44 µm were significantly more retentive than those cemented with a film thickness of 113–255 µm.

Convergence, a primary feature of preparation geometry, is the angle between opposing axial walls and has been shown to affect crown retention with optimal retention occurring between 5° and 12°. Clinically, ideal axial wall convergence is not routinely obtained. Studies have reported mean convergence values ranging from 14° to 20°.

Reported mean total occlusal convergence (TOC) angles range from 12.2° to 27°, depending on whether the preparations were completed in the preclinical laboratory or in clinical situation [3, 5, 7–9, 12–14]. Hence in the present study, a convergence angle of 10° and 30° was selected to approximate clinical situation.

The functional masticatory forces exerted by stomatognathic system in the oral cavity are small compared to static isometric closing forces. Under normal masticatory conditions, the forces exerted on the occlusal surface seldom exceed 10–15 pound [17, 18]. In the present study three different occlusal loading conditions were applied i.e. 100 N point horizontal load, 100 N point vertical load and 100 N distributed load across the occlusal surface acting apically to simulate clinical condition.

The first load case where the load was applied horizontally near the occlusal surface, represented a load that could well displace a crown by rotation, particularly where the preparation height was low. The second load case, where a point load was applied to the occlusal surface in an apical direction, represented a typical swallowing load. The third load case where the load was distributed uniformly over the entire occlusal surface of the crown simulated the effect of crushing a food bolus. The results obtained showed that the maximum stress occurred in the first case during horizontal Point loading followed by vertical Point loading and least in distributed loading. The magnitude of the stresses were rather smaller in distributed axial loading than the previous two load cases, as a result of distributing the load over a wider area leading to better stress homogenization. The results of this analysis concur with findings of Kamposiora et al. [26] who concluded that stresses under oblique stressing were higher than under axial stressing.

It would seem logical that the combination of a short preparation with 30° taper would present the most unfavourable stress field. However, the combination of a 30° taper with a long preparation had the most unfavourable stress field which is in direct opposition to the traditional teaching of fixed prosthodontics. One possible explanation for this finding may be the juxtaposition of the 30° taper with the maximum shear stress plane, which occurs at 45° to the vertical axis of the preparation. Therefore, as the taper increased from 10° to 30°, the sampling plane in the cement lute became increasingly influenced by the maximum shear stress [19]. The maximum stresses with the 10° preparation taper were consistently smaller than with the 30° taper. Standard texts on Fixed Prosthodontics [41, 42] suggest that the optimal taper for a crown preparation is between 5° and 10°, this assumption being based on laboratory based biomechanical studies carried out in the 1950’s and 1960’s. The results from this study seem to suggest that a smaller preparation taper of 10° is also biomechanically more acceptable than a 30° taper, as the stress distribution is more favourable.

Various researchers [43, 44] have suggested that the majority of dental structure failures are attributable to a process that finds its catastrophic end only after many years of service. This is where a fissure grows slowly under fatigue stress, structurally weakening the component until final breakage occurs. When combined with the stress analysis results found in this study, such a concept is useful for indicating where fracture initiation is most likely to occur.

The applied stress in this investigation was 100 N. It should be recognized that mastication is a complex procedure influenced by age, gender, food texture, occlusal scheme, time, and presence of temporomandibular disorders. As a consequence, opening and closing velocities, directional changes, and width of lateral movements vary considerably among individuals. Additionally, in vivo temperature fluctuations and chemomechanic and microbiologic influences create a hostile environment for the longevity of the restorations. Although the differences in the magnitude of the stresses were small under the various parameters specified, it is possible that after many years of function, the slightly higher stress field of the 30° taper crown would result in crack initiation and propagation and ultimately loss of retention.

Fatigue, which is defined as progressive fracture under repeated loading, should be used for predicting cement failure for clinical situations involving low stresses and significant cyclic loading. Endurance limit is approximately 40 % of the ultimate strength of the material represents the limit up to which a material can be subjected to an infinite number of cycles without failing [26]. The ultimate tensile strength (UTS) of the cement used in this study ranged from 9 to 20 MPa. Resultant stresses in the present study were very low and did not reach the estimated endurance limit. However under horizontal stressing, stress values were higher than the endurance limit, and some conditions actually exceeded the UTS of the cement.

Further, the results showed that models with chamfer margin exhibited increased stresses in comparison to shoulder margins. There was a relative increase in stress magnitude at the junction of shoulder and chamfer finish line but it was lesser in magnitude compared to chamfer region. This finding is consistent with the study conducted by Kamposiora et al. [26] who stated that stresses at the margins of crowns with chamfer marginal configuration were higher than those with shoulder margins. However few studies also exist which contradict the aforementioned result. Farah and Craig [22] compared chamfer, chisel, and shoulder with a bevel with absence of cement interface and concluded that most uniform stress distribution was exhibited by chamfer finish line. Chai and Steege [45] evaluated the effects of labial margin design (chamfer, rounded shoulder with knife-edge metal finish, rounded shoulder with beveled metal finish) on stress distribution of metal–ceramic crowns, and concluded that there were no differences at the cement–dentin and metal–cement interface of the different margins.

In this study, it is important to analyze qualitative rather than quantitative comparison from the result values because of the simplifications in the modeling procedure owing to some limitations. The stress variations under average conditions have been simulated without attempting to simulate individual clinical situations. Firstly the root, periodontal ligament and supporting alveolar bone were not modeled which could lead to some errors. Secondly, the loads applied were static loads that were different from dynamic loading seen during function. Thirdly, these models examined just one crown in isolation. However, in the mouth, teeth contact adjacent teeth and dissipate some load to them via the proximal contact areas. There is biological variation due to intra-oral factors such as tooth size, arch shape and occlusal load. Finally, in the model used for the study, all material was assumed to be linearly elastic and homogenous in nature whereas, tooth is a viscoelastic, anisotropic, and heterogeneous material. The resultant stress values obtained may not be accurate quantitatively but are generally accepted qualitatively. Further research comprising computerized techniques combined with experimental techniques and long term clinical evaluation should be conducted to establish the true nature of the biologic system.

Within limitations of the study, the following conclusions were drawn (i) The stress fields in the cement lute were highest under horizontal point loading followed by vertical point loading and least under distributed loading. Distributed axial loading shows the least stress, better stress homogenization and gives a favorable prognosis for the fixed prostheses. Therefore the forces should be directed along the long axis of the tooth. This can be achieved by carefully planning the occlusion. (ii) Smaller preparation taper of 10° is biomechanically more acceptable than a 30° taper, as the stress distribution in the cement lute is lesser. (iii) As the maximum stress field was associated with a combination of 30° taper and long preparation (5 mm) it is desirable to decrease taper as height increases. (iv) Increased cement stresses may be associated with chamfer margin design resulting in increased risk of microfracture and failure.

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