EVALUATION OF STRESS DISTRIBUTION IN FIXED PARTIAL DENTURES WITH PIER ABUTMENTS USING RIGID AND NONRIGID CONNECTORS A FINITE ELEMENT ANALYSIS

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CERTIFICATE

This is to certify that this dissertation titled **"Evaluation Of Stress Distribution In Fixed Partial Dentures With Pier Abutments Using Rigid And Nonrigid Connectors – A Finite Element Analysis"** is a bonafide record of work done by **Dr. K.Sobha** under my guidance during her postgraduate period between 2008- 2011. This Dissertation is submitted to **THE TAMILNADU Dr. M.G.R. MEDICAL UNIVERSITY**, in Partial fulfilment of requirements for the Degree of **Master of Dental Surgery in Prosthodontics and Crown & Bridge (Branch I).**

It has not been submitted (partial or full) for the award of any other degree or diploma.

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DECLARATION

I, **Dr. K. SOBHA**, do hereby declare that the dissertation titled "**Evaluation Of Stress Distribution In Fixed Partial Dentures With Pier Abutments Using Rigid And Nonrigid Connectors –A Finite Element Analysis"** was done in the Department Of Prosthodontics, TamilNadu Government Dental College & Hospital, Chennai 600 003. I have utilized the facilities provided in the Government Dental College for the study in partial fulfilment of the requirements for the degree of **Master of Dental Surgery** in the speciality of **Prosthodontics and Crown & Bridge (Branch I)** during the course period 2008-2011 under the conceptualization and guidance of my dissertation guide, **Dr.A.Meenakshi MDS.**

I declare that no part of the dissertation will be utilized for gaining financial assistance for research or other promotions without obtaining prior permission from the TamilNadu Government Dental College & Hospital.

I also declare that no part of this work will be published either in the print or electronic media except with those who have been actively involved in this dissertation work and I firmly affirm that the right to preserve or publish this work rests solely with the prior permission of the Principal, TamilNadu Government Dental College & Hospital, Chennai 600 003, but with the vested right that I shall be cited as the author(s).

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INTRODUCTION

In recent years fixed prosthesis has obtained an increasing acceptance from partially edentulous patients as it regains comfort, masticatory ability, appearance, health and integrity of the dentition. Planning a fixed prosthesis requires a sharp acumen to diagnose the presenting conditions and a thorough knowledge of the available treatment methods.

In the success of fixed partial denture, abutment plays an important role. The type of occlusion, amount of bone present and periodontal health of the tooth also determines the success rate. The health of the periodontium depends upon several factors and one such vital factor is the magnitude and direction of load and the stresses induced thereupon. The vertical stress directing along the long axis of the tooth is less injurious when compared to the oblique force, which is more deleterious to the periodontium[1, 2]. Apart from the load applied, the resilient character of the restoration also plays an appreciable role[3].

The occlusal stress may cause periodontal injury when it goes beyond the adaptive capacity of the periodontium. Torque is the most dangerous force to injure the periodontium to the maximum level due to development of shear stress. Deleterious forces can also cause bone resorption and inflammation of the periodontium[4, 5].

Very often it is necessary either to modify the treatment plan or change the design of restoration depending upon the amount of stress taken up by the individual abutments. The clinical situation of two edentulous spaces with a single tooth (pier) between the spaces is possible. Restoration of two missing teeth and an intermediate pier abutment with a rigid fixed partial denture is not an ideal treatment[6]. High stress concentration may occur at pier abutments and excessive displacements may be observed at terminal abutments resulting in damage to the abutments. It has been said that in such a situation a nonrigid connector can be used to eliminate the fulcrum action of a pier abutment[6].

Hence it was decided to conduct a study on the stress distribution with rigid and nonrigid connectors in fixed partial dentures with pier abutments when they are subjected to constant magnitude of occlusal load applied in the laboratory.

The finite element analysis is one of the most frequently used and accepted method to study stress analysis in both industry and science. Finite Element Analysis is a technique for obtaining a solution to a complex mechanical problem by dividing the problem domain into a collection of much smaller and simpler domains or elements in which the field variables can be interpolated with the use of shape functions[7].

Finite Element Analysis was initially developed in the early 1960's to solve structural problems in the aerospace industry. Later it was used to solve problems in heat transfer, fluid flow, mass transport and electromagnetics[7]. Farah et al introduced finite element method (FEM) study in dentistry for the first time, proving its efficiency to be better than photoelastic study in terms of easy modeling and more defined stress analysis[8]. FEM results do not vary by repetition of the analysis and are restricted by the number of nodules and elements used in the model and the elastic constants attributed to the elements[8].

So in this study, the application of finite element method to analyze the stress distribution to the periodontium with rigid and a different orientation of nonrigid design at various locations for a five unit fixed partial denture with pier abutment has been proposed under different loading methods. The null hypothesis for the study was the use of different orientation of nonrigid design at various locations for a fixed partial denture with pier abutment does not influence the stress distribution to the pier abutment and periodontium.

AIM AND OBJECTIVES OF THIS STUDY

Aim

To evaluate the amount of stress distribution in fixed partial dentures with pier abutments using rigid and nonrigid connectors.

Objectives

1. To evaluate the amount of stress transmitted to the supporting structure by loading a fixed partial denture with pier abutment using a rigid connector design.
2. To evaluate the stress distribution using a different orientation of nonrigid connector design in four locations:

   Distal to mesial abutment (canine)

   Mesial to pier abutment (second premolar)

   Distal to pier abutment (second premolar)

   Mesial to distal abutment (second molar)
3. To evaluate the stress distribution under different loading conditions:

   Loading of all teeth to simulate maximum centric occlusion contacts

   Loading of canine to simulate a single anterior contact

   Loading of second molar to simulate a single posterior contact
4. To compare the stress distribution with rigid and nonrigid design types.

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REVIEW OF LITERATURE

Farah JW, Craig RC (1974)[8] were the first to use finite study in dentistry. They analysed the stresses in a restored axisymmetric molar. They said for calculation following information was needed

1. Total number of nodal points.
2. Total number of elements.
3. A numbering system identifying each element.
4. Young's modulus and poisson's ratio of each element.
5. A numbering system identifying each nodal point.
6. Co-ordinates of each nodal point.
7. Evaluation of strains at external nodes.
8. Types of boundary elements.

Hood .J.A.et al (1975)[9] studied on modification of stresses in alveolar bone induced by a tilted molar. The following conclusions were reached.

1. Altering the angle of the load applied to the unsupported molar from 0 (axial) to 30 degrees resulted in a fourfold increase in compressive stress in the supporting bone mesial to it.

2. Increasing the load from 30 to 90 pounds while maintaining a 30 degree angle of application resulted in a linear shear stress on the supporting bone mesial to the tooth.

3. Following the placement of a fixed partial denture, the induced stress at a point on the mesial aspect of the molar tooth, subjected to a 60 pound load at 30 degrees to the long axis, was reduced from 241 to 43 p.s.i

Yettaram .A.L (1976)[10] studied stress distribution patterns for a normal and restored mandibular second premolar under masticatory type forces using finite element method of stress analysis applied to two dimensional models. Force was applied at two points to simulate active centric occlusion. The structure was also subjected to single point load, which was applied to lingual side of the buccal cusp of the tooth. Results concluded that greater stiffness of enamel over dentin enabled it to react to the larger proportion of the applied loads. Dentin core was relatively lightly stressed.

Sutherland JK, Holland GA (1980)[11] conducted a photoelastic analysis of the stress distribution in bone supporting fixed partial dentures of rigid and nonrigid design. It was concluded that

1. Under conditions of the vertical loading, the rigid fixed partial denture design does not permit independent response by either abutment. The nonrigid fixed partial denture design allows some independence in response to the vertical loading.

2. The stress distributions and concentrations produced in the supporting bone were favourably altered by the placement of a fixed partial denture of rigid or nonrigid design.

3. The distribution of stresses in the supporting bone varies with the number and location of the loading sites.

4. Under conditions of vertical loading, the Ney and Stern nonrigid fixed partial designs exhibit no significant differences in stress distribution or concentration.

Gobind.H.Atmaran Hamdi mohammed (1981)[12] determined the physiological stress values in natural tooth and the underlying bone using finite element analysis. In addition to modeling the periodontal ligament as a continuous structure, periodontal ligament was modeled more accurately in a novel fashion as a fibrous structure. The results indicate that type of periodontal ligament had significant influence on nature and magnitude of alveolar stresses, and that fibrous periodontal ligament modeling shows higher and more widely distributed lateral stresses in alveolar bone than those resulting from continuous periodontal modeling.

Sulik. W.D.et al (1981)[13] investigated stress distributions and concentrations produced in the periodontium of abutment teeth of a fixed partial denture. Stress concentrations produced in the periodontium of abutment teeth were notably altered by a moderate (20%) loss of support. Further (40%) loss of periodontium did not result in appreciably additional change. The stress patterns produced by loss of periodontium were favourably altered by placement of a fixed partial denture.

Anusavice KJ (1986)[14] calculated the stress distribution in anterior metal-ceramic crowns fabricated with either gold alloy or nickel alloy copings of reduced thickness using plane stress analysis. Two dimensional finite element models of three crown designs were subjected to a simulated biting force of 200N which was distributed over porcelain near the lingual metal- ceramic junction. The maximum stresses and strains in porcelain for the crowns with a conventional coping thickness (0.3mm) and a reduced coping thickness (0.1 mm) were not significantly different. All values were below the critical failure rates of porcelain.

Laurell L, Lundgren D (1986)[15] a study was done to elucidate the occlusal force pattern and the functional capability of dentition during chewing and biting. The method was based on the use of strain guage transducers mounted into preformed matrixes evenly distributed over the tooth arch. The magnitude of the occlusal forces developed during chewing and swallowing was below all biting forces.

Farah J.W. R.G Craig (1988)[16] conducted a two dimensional finite element analysis of a mandibular quadrant to examine the stresses and displacements resulting from a 100N load placed as follows: (i) distributed on the second molar, (ii) distributed on the second premolar and second molar, (iii) concentrated at 30 degree to the vertical on second molar. Young's modulus and Poissons ratio for each material were selected from accepted values. The Principal stresses were determined throughout the model, with special emphasis being placed for elements in immediate vicinity of teeth mentioned above. Resulting stresses were approximately 4-5 times greater than those resulting from a vertically distributed load.

Farah J.W Craig R.G (1988)[17] examined the principal stresses from placement of three and four unit bridges, spanned from first premolar to second molar using two dimensional finite element methods. He concluded that the addition of a bridge resulted in lower and better distributed stresses. From a stress standpoint the bridges resulted in a more uniform stress distribution around the abutments and an increase in the tensile stress distal to the abutments. Such findings support the placement of a fixed bridge to maintain bone in an edentulous area.

Zhao.Y.F, et al (1989)[18] conducted a two dimensional finite element method to study the stress distribution in the periodontal supporting tissues at the time when the second bicuspid and the second molar were vertically and obliquely loaded. The conclusion was as follows:

1. When the vertical loading was applied to the occlusal surface of the bicuspid and molar, the stress distribution of periodontal supporting tissues was uniform.

2. The stress concentration was on the marginal ridge and the distal apex of bicuspid under the oblique loading.

3. When the oblique loading was applied to the occlusal surface of the molar, the stress concentration was only on the marginal ridge, and it was small.

Yang HS, Thompson VP (1991)[19] investigated the changes in mechanical behaviour of the supporting structures when a fixed prosthesis replaced a missing mandibular first molar through a finite element method. In the unrestored situation, as the degree of bone resorption increased, there was a corresponding increase of stress in the periodontium. The presence of a fixed prosthesis markedly reduced the magnitude and distribution of stress in periodontium.

Aydin AK, Tekkaya (1992)[20] analyzed stresses and deflections of abutments induced by various loadings with two dimensional finite element models. The biomechanic system consisted of three unit posterior fixed partial denture (1) a distributed force of 600 N (2) concentrated nonaxial and (3) axial 300 N forces at the marginal ridge of the molar; and (4) a concentrated vertical 300 N force at the center of the pontic. All computations were conducted for three different alveolar bone levels. According to the stresses induced in the alveolar bone, the most critical loading was the distributed force. With diminishing periodontal support, stresses elevated in the biomechanic system and critical increases were noted for the concentrated nonaxial load on the molar.

Gary R. Goldstein (1992)[21] evaluated the flexion under compressive load of a four-unit mandibular FPD replacing the second premolar and the first molar using holographic interferometry. The results demonstrated that solder joints at the junction of the premolar and molar pontics flexed under a reduced compressive load and exhibited a higher failure rate than other connector designs.

M.B.Moulding, G.A.Holland (1992)[22] analyzed the advantages and disadvantages of an alternative orientation of nonrigid connectors in fixed partial dentures. Nonrigid connectors have been advocated for fixed partial dentures. However, space limitations may require overreduction of the preparation or overcontouring of the retainer to place the keyway within the retainer wall. An inverted orientation of the nonrigid connector can resolve these problems. With this design, the key is attached to the distal surface of the mesial retainer in a dual-abutment fixed partial denture, and the keyway is incorporated in the mesial surface of the pontic.

Misch CM et al (1993)[23] conducted a three dimensional finite element stress analysis to compare models representing a natural tooth and an integrated implant connected with rigid and nonrigid prosthesis. Based on the similarities in both the patterns of stress contours and the stress values generated in the two models, advocating a nonrigid connection because of a biomechanical advantage may be erroneous.

Seaton P (1994)[24] studied movements caused by the application of chewing loads. The location and magnitude of tensile and shear stresses affecting cement within retainers during mastication was related to the type of movement and determined by differences in mobility of abutments at each end of the fixed partial denture, length of span, and point of chewing load. The incidence of cement failure could be reduced with improved strategic stress resistance.

Junro Yamashita (1997)[25] conducted a study to determine the strain distribution of fixed partial dentures during function and to compare the biomechanical behaviour of fixed partial dentures invivo and invitro. A static load was produced through occlusal force invivo and with a universal testing machine invitro. The measurements were recorded by a strain guage method. The results of this study suggest that serious problems with strain may occur in a long-span mandibular posterior fixed partial denture.

Argiris L. Pissiotis (1998)[26] describes a procedure that uses Ney MS (Minimal Space) intracoronal attachment as an interlock in a pontic. This procedure overcomes the disadvantages associated with the use of the intracoronal attachment, which are (1) excessive tooth reduction often required to place the attachment within the contour of the crown; (2) compromised embrasures, which result in oral hygiene and periodontal problems; and (3) poor esthetics.

Hassan M Zaida (1998)[27] analysed the stresses induced in a pier retainer of an anterior resin-bonded fixed partial denture using a photoelastic study. Isochromatic fringes indicated a stress magnitude at the proximolingual areas of the pontic in the 3-unit resin-bonded fixed partial denture. In the 5-unit resin-bonded prosthesis, the stress pattern appeared to involve the entire surface of the pier retainer. The use of pier abutments should be avoided and it is more favourable to use 3-unit resin bonded fixed partial dentures.

Issac L (1999)[28] conducted a finite element analysis of a three unit fixed partial denture cast with nickel-chromium alloy. A two dimensional mathematical model was generated and a load of 1 kg was applied to the occlusal surface of the casting. Maximum stresses were developed in the pontic and connectors with distal connector experiencing the maximum stresses. Stresses transmitted to the dentin were comparatively lower and more of compressive in nature. The underlying bone experienced moderate amount of both compressive and tensile stresses but the displacement in this tissue were minimal compared to the rest.

Russel D.Nishimura, Kent (1999)[29] measured photoelastically the stress transfer patterns with variable implant support and simulated natural teeth through rigid and nonrigid connection under simulated functional loads. The rigid connector demonstrated more widespread stress transfer. Recommendations for selection of connector design should be based on sound clinical periodontal health of a tooth and the support provided by implants.

Yang HS,et al (1999)[30] conducted finite element stress analysis on the effect of splinting in fixed partial dentures. This study analysed the stress levels in the teeth and supporting structures of a fixed prosthesis and ascertained how the addition of multiple abutments in a fixed prosthesis modifies the stresses and their deflection. A reduction of stress and deflection was observed in the supporting structures when a fixed partial denture was fabricated and teeth were splinted together. Increasing the number of splinted abutments did not reveal proportional reduction of stress in the periodontium. Stress concentrations were seen in the connectors of prosthesis and in the cervical dentin area near the edentulous ridge.

Akpinar I et al (2000)[31] evaluated natural tooth's stress distribution in occlusion with a dental implant. This study investigated stress formed around the implant and the antagonist natural tooth under occlusal force. The results indicate that a bite force of 143N resulted in high compressive stress around the roots of a natural tooth opposing a restoration.

Ciftci. Y. et al (2000)[32] in this study, the effect of various materials used in fabricating super structure for implant retained fixed partial denture on stress distribution around implant tissue were investigated. Gold alloy and porcelain produced the highest stress value. Stress created by acrylic resin and reinforced composite resin were 25% and 15% less, respectively than porcelain or gold alloy.

Duyck J et al (2000)[33] the aim of this study was to investigate the influence of prosthesis material on the distribution and magnitude of load on oral implants carrying a fixed partial prosthesis by invivo quantification and qualification of this load. A significantly better distribution of bending moments with the metal prostheses was observed in the case of three unit prostheses.

Issac L, Joseph M (2000)[34] a two-dimensional finite element analysis was carried out to analyse the stress variations in a mandibular posterior fixed partial denture, made of recast nickel-chromium alloy. The study revealed that the connectors experienced maximum stresses and the generated stress values decreased within the partial denture made of recast Ni-Cr alloy.

Tang L, et al (2000)[35] conducted comparative analysis with stress of the cortical bone beneath different pontics of mandibular posterior fixed bridge using three dimensional finite element method. One vertical load of 20 kg and one horizontal load of 20 kg were applied respectively on the occlusal surface of the abutments and the pontics of three different fixed bridges. Under the vertical loading, the cortical bone beneath the pontics showed compressive stresses. Under the horizontal loading, the cortical bone beneath the pontics exhibited tensile stresses and compressive stresses. The stress in the cortical bone beneath the pontics increased when the area of contact was reduced.

Van Ejiden TM (2000)[36] studied biomechanical behaviour of the mandibular bone tissue, and of the mandibular bone as a whole, in response to external loading. The result was complex pattern of stresses and strains (compressive, tensile, shear, torsional) in the mandible. To be able to resist forces and bending and torsional moments, not only the material properties of the mandible but also its geometrical design is of importance. In the longitudinal direction, the mandible is stiffer than in transverse directions, and the vertical cross-sectional dimension of the mandible is larger than its transverse dimension. These features enhance the resistance of the mandible to the relatively large vertical shear forces and bending moments that come into play in the sagittal plane.

Ziada HM, Barrett BE (2000)[37] conducted an invivo study using a nonrigid connector for a resin bonded bridge. A nonrigid connector within the pontic distal to the pier retainer was constructed and it remained in place without debonding for seven years.

Nakamura T et al (2001)[5] evaluated stress analysis of metal-free polymer crowns using the three dimensional finite element method. The purpose of this study was to evaluate the stress distribution under various loading conditions within posterior metal-free crowns made of new composite materials. A three dimensional finite element model representing a mandibular first molar was constructed. A load of 600N simulating the maximum bite force was applied vertically to the crowns. Loads of 225N, simulating masticatory force, were applied from three directions (vertically, at a 45 degree angle, and horizontally). When the load was applied horizontally, the maximum tensile stress was observed around the loading points on the surface in the case of composite and glass-ceramic crowns, and in the cervical area of the metal coping in the porcelain fused to metal crowns.

Aykul H (2002)[38] conducted a study to calculate stress distribution in metal – porcelain crowns by using a three-dimensional finite element method. The tooth model was crowned with Au-Pd alloy, Ni-Cr alloy and porcelain. A load of 450N, at an angle of 45 degrees to the longitudinal axis was applied on the occlusal margin of the crown tooth. The highest stress values were observed when Ni-Cr alloy and porcelain was used.

Dalkiz M (2002)[39] investigated the designs of osseointegrated prostheses in cases of free-end partial edentulism using comparative stress interpreted with the three-dimensional finite element method. Three free-end fixed osseointegrated prostheses models with various connection designs (i.e; rigidly connected to an abutment tooth and an implant, rigidly connected to an implant and two abutment teeth, and rigidly connected to an implant and three abutment teeth) were studied. The stress values of the three models loaded with vertical, buccolingual, and linguobuccal directions at 30 degrees angled to vertical axis forces were analysed. When the fixed partial denture was connected to the three natural abutment teeth and an implant, the lowest levels of stress in the bone were noted.

Proos KA, et al (2002)[40] conducted finite element analysis of a metal-ceramic crown on a first molar tooth. This study evaluated the stresses developed during loading in a first premolar metal-ceramic crown made of different metal cores. An axial load of 600N was applied vertically downward, over a circular area around the crown's fissure. They concluded the peak maximum principal tensile stress in the porcelain existed on the surface of the crown, partially outside the cusp, with the greatest peak in the gold-porcelain system (15.8MPa). The maximum Von Mises stress existed in the metal coping, in the radial edge at the axial/occlusal line angle, with the highest maximum in the nickel-chromium system (143.9 MPa)

Cheng B, Zhao Y (2003)[41] studied the effects of different occlusal thickness on the stress distribution of all-ceramic crowns of the mandibular first molar using a three dimensional finite element analysis. It was found that under the simulated applied loads, the values of tensile and shear stress varied with the occlusal thickness, and much greater values of such stresses were noticed in the all-ceramic crowns 1.0 mm in occlusal thickness, compared with those in the crowns 1.5 mm and 2.0 mm in occlusal thickness.

Ishigaki. S. et al (2003)[2] the aim of the study was to reveal the biomechanical stress distribution in supporting bone around an implant and natural tooth under chewing function. The tooth model showed smooth stress concentration in the supporting bone with low stress concentration around the neck of the implant.

Lin CL, Wang (2003)[42] analysed the biomechanics in an implant/tooth-supported system under different occlusal forces with rigid and nonrigid connectors by adopting a nonlinear finite element approach. A model containing one Frialit-2 implant (placed in the second molar position) splinted to the mandibular second premolar was constructed. Nonlinear contact elements were used to simulate a realistic interface fixation between the implant body and abutment screw and the sliding keyway stress-breaker function. Stress distributions in the splinting system with rigid and nonrigid connectors were observed when vertical forces were applied to the tooth, pontic, implant abutment, or complete prosthesis in 10 simulated models. Minimization of the occlusal loading force on the pontic area through occlusal adjustment procedures to redistribute stress within the implant system in the maximum intercuspation position for an implant/tooth-supported prosthesis is recommended.

Eraslan O, Sevimay M (2005)[43] studied the stress distribution in distal cantilevered fixed partial dentures that are designed with different cantilever morphology and made from different restorative materials using a finite element study. Von Mises stress values with maximum stress concentrations were observed on connectors of distal cantilevers. Models with premolar cantilever extensions restored with all-ceramic induced lower Von Mises stress values than metal-ceramic restorations, however models with molar cantilever extensions restored with all-ceramic restorations induced higher Von Mises stress values than metal-ceramic restorations.

Chun-Li Lin, SH Chang (2006)[44] investigated the mechanical interactions of implant-teeth splinting systems under different periodontal supports and number of splinted teeth with rigid and non-rigid connectors using non-linear finite element approach. The simulated results indicated that the cross-interaction of the periodontal support and the splinting situation was a major factor affecting the stress value in alveolar bone. An additional splinting decreased the stress values of bone significantly for a compromised periodontal support. Also, the stress values of the implant and prostheses increased, but were decreased in bone when the splinting system used non-rigid connectors. The mobility of natural teeth and the implant system between rigid and non-rigid connections showed only small differences.

Motta AB, Pereira LC (2007)[45] conducted a 2D finite element study to compare the stress distribution on 3-unit all-ceramic and metal-ceramic fixed partial dentures and identified the areas of major risk of failure. Three models were designed: (1) metal-ceramic fixed partial denture (2) all-ceramic fixed partial denture with the veneering porcelain on the occlusal and cervical surface of the abutment tooth; (3) all-ceramic fixed partial denture with the veneering porcelain only on the occlusal surface. A 100 N load was applied in an area of 0.5mm^2 on the working cusps, following these simulations; (1) on the abutment teeth and the pontic; (2) only on the abutment teeth; (3) only on the pontic. In conclusion, the best stress values and distribution were found for the all-ceramic fixed partial denture with the veneering porcelain only on the occlusal surface. However, in under clinical situations, fatigue conditions and restoration defects must be considered.

Ozcelik T, Ersoy AE (2007)[46] examined stresses formed around the implant and natural tooth abutments under occlusal forces, using two dimensional finite element and photoelastic stress analysis methods. Three tooth/implant supported fixed prostheses (screw type implant, 3.75mm× 13mm) models with various connection designs (i.e, rigidly connected to an abutment tooth, connected to an abutment tooth with a non-rigid connector, connected to an abutment implant with a non-rigid connector were studied. The highest level of stresses around the implant abutment was noted on the tooth/implant supported fixed prostheses with the rigid connector. On the other hand, non-rigid connectors incorporated into prostheses at the site of the implant abutment reduced the level of stresses in bone. It was concluded that if tooth and implant abutments are to be used together as fixed prostheses supports, non-rigid connectors should be placed on the implant abutment-supported site.

Tanino F, Hayakawa I (2007)[47] examined the effect of stress-breaking attachments at the connections between maxillary palateless overdentures and implants using a three finite element study. In each model, the influence of the stress breaking attachments was compared by changing the elastic modulus from 1 to 3,000 MPa and the thickness of the stress breaking material from 1 to 3 mm. As the elastic modulus of the stress breaking materials increased, the stress increased at the implant-bone interface and decreased at the cortical bone surface. Moreover, stress at the implant-bone interface with 3 mm thick stress-breaking material was smaller than that with 1 mm thick material.

Selcuk Oruc, Oguz Eraslan (2008)[48] evaluated the effects of nonrigid connectors on fixed partial dentures with pier abutments using a finite element study. It was concluded that the area of maximum stress concentration at the pier abutment was decreased by the use of a nonrigid connector designed as free (non bonded) touching surfaces at the distal region of pier abutment.

Manda M, Galanis C (2010)[49] investigated the effect of increasing the vertical dimension on the maximum stress developed within the connectors during the static loading of a cross-arch fixed partial denture extended as a 1- and 2- unit cantilever using a three dimensional finite element analysis. The connector with the highest risk of failure is the 3mm connector distal to the retaining abutment of the 2-unit cantilever restoration. Increasing the vertical dimension is beneficial for the connector distal to the retaining abutment, while the resultant stress changes are not substantial for the connectors mesial to the retaining abutment.

Ditter MP, Kohorst P (2010)[50] conducted a three dimensional finite element study to investigate the influence of the design and material composition of the supporting structure of a zirconia four-unit fixed partial denture on stress distribution during invitro loading. It was concluded that the choice of material for abutment teeth and the socket, as well as the type of tooth support, significantly influence stresses generated in fixed partial dentures during invitro load tests. To achieve realistic results, fixed partial dentures should be supported by resiliently embedded abutment teeth made of moderately rigid material (eg, polyurethane). In clinical practice, the risk of failure is likely to rise with an increasing resilience of the abutment teeth if occlusal contacts are directed over the pontic/connector region rather than being spread over the retainers.

Teixeira MF, Ramalho SA (2010)[51] evaluated the stress on the cortical bone around single body dental implants supporting mandibular complete fixed denture with rigid or semirigid splinting system after axial and oblique occlusal loading simulation, through a finite element analysis. It was concluded that the use of a semirigid system for rehabilitation of edentulous mandibles by means of immediate implant supported fixed complete denture is recommended, because it reduces stress concentration in the cortical bone.

MATERIALS AND METHODS

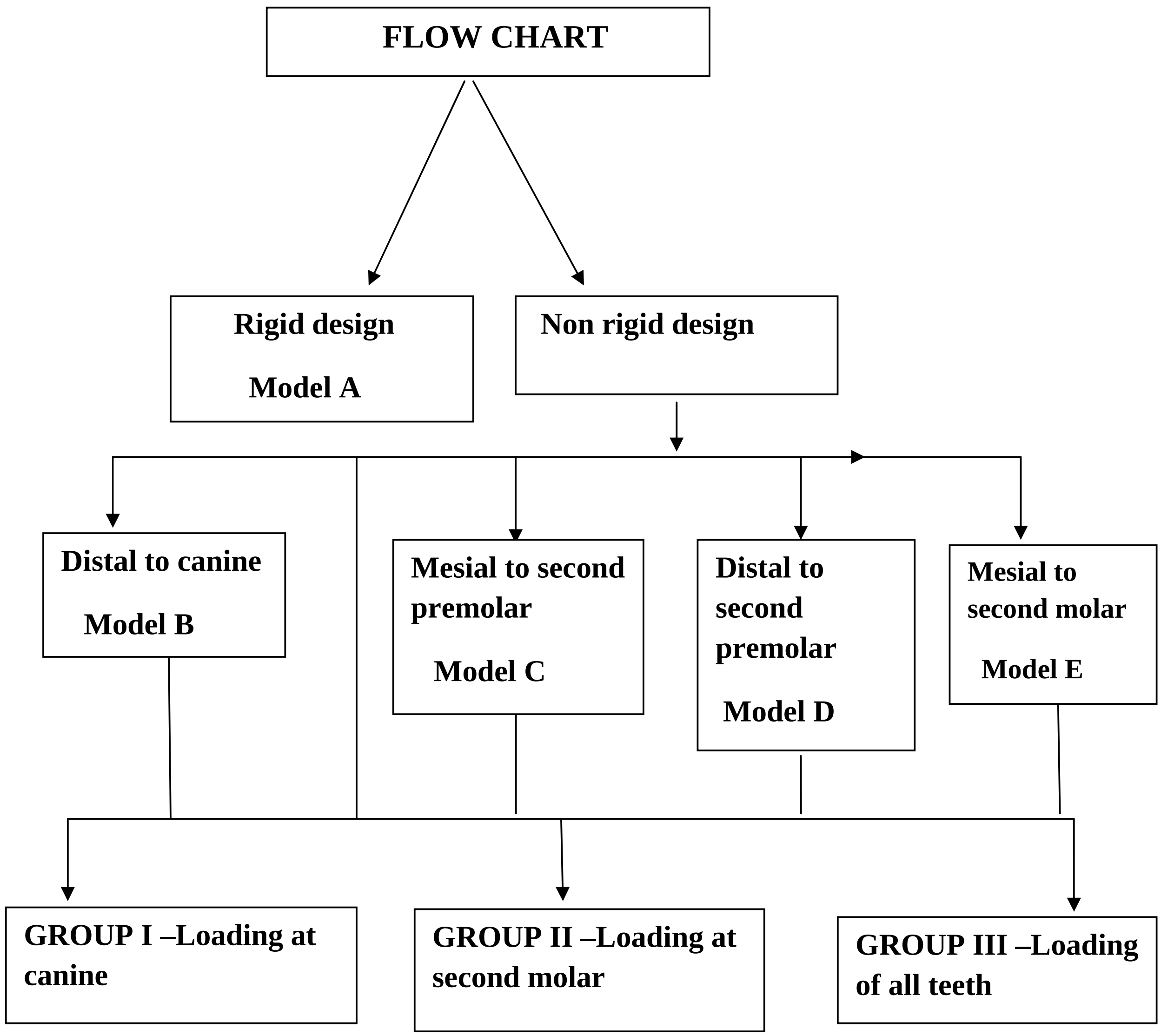
This study was conducted to evaluate the effects of rigid and a different orientation of nonrigid connector design on stress distribution in fixed partial dentures with pier abutments using finite element analysis. The model consisted of a 5-unit metal-ceramic fixed partial denture with the canine, second premolar, and second molar as abutment teeth supported by simulated periodontal and alveolar bone structures.

TABLE I

<table>
<tr><th>S.NO</th><th>5 UNIT FIXED PARTIAL DENTURE DESIGN TYPE</th><th>LOCATION OF NONRIGID CONNECTOR</th><th>MATERIALS</th></tr>
<tr><td>1</td><td>Rigid</td><td>Not applicable</td><td>Ni-Cr alloy, feldspathic porcelain (Ivoclar)</td></tr>
<tr><td>2</td><td>Non rigid</td><td>Distal to canine</td><td rowspan="4">Ni-Cr alloy, feldspathic porcelain (Ivoclar)
Rhein 83 attachment</td></tr>
<tr><td>3</td><td>Non rigid</td><td>Mesial to second premolar</td></tr>
<tr><td>4</td><td>Non rigid</td><td>Distal to second premolar</td></tr>
<tr><td>5</td><td>Non rigid</td><td>Mesial to second molar</td></tr>
</table>

TABLE II

S.NO	MAIN COMPONENT OF FINITE ELEMENT ANALYSIS
1	Elements – Each simple shape
2	Mesh – The whole collection of elements
3	Nodes – the element equations formed by known values of properties at fixed points on the elements


FLOW CHART
Rigid design
Model A
Non rigid design
Distal to canine
Model B
Mesial to second premolar
Model C
Distal to second premolar
Model D
Mesial to second molar
Model E
GROUP I –Loading at canine
GROUP II –Loading at second molar
GROUP III –Loading of all teeth

INSTRUMENTS USED FOR THE STUDY

1. Personal computer configuration:

MONITOR	- IBM TFT MONITOR
CPU	- IBM (INTELLISTATION Z PRO)
PROCESSOR	- INTELXEON (DUAL PROCESSOR)
MEMORY CAPACITY	- PRIMARY- 2GB, SECONDARY- 80 GB
GRAPHICS CARD	- ATI FIREGL V 7100

2. Software specification:

For modeling	- CATIA V5R18
For meshing and analyzing	- ANSYS workbench 12.0,

(ANSYS inc, USA)

METHODOLOGY

This simulation study was conducted to evaluate the influence of rigid and nonrigid connectors in conjunction with pier abutments on stress distribution in the pier abutment and supporting structures. The finite element method 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[7]. It has been suggested as an effective method to determine stress distribution patterns for complex design[8].

A continuous mathematical model was developed for a 5-unit fixed partial denture with rigid and nonrigid designs. The model was subdivided into numerous discrete elements, which are then connected at nodal points. Linear equations were designed to relate the nodal forces to nodal displacements, and they were subsequently solved using a digital computer.

PHILOSOPHY OF FINITE ELEMENT ANALYSIS

The finite element method is an accepted theoretical technique used in the solution of engineering problems. This method has also been used for biomechanical analysis in orthopedic, cardiovascular, and dental structures. The finite element method provides a unique way of determining stress and displacements because of its ability to model geometrically complex structures[8].

Essentially any problem can be split up into a number of smaller problems with finite element method; this is done by considering that a complex geometrical shape is made up of a number of simpler shapes. For e.g. a circle might be approximated by a series of triangles in an attempt to calculate the area of the circle. This is known as “spatial discretization” with each simple shape being known as an “element” and the whole collection of elements being known as “mesh”[7].

Within each element the relevant property of the material is predicted, each element is given life by inducing into them the properties of original material which it represents. 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. This is done without any reference to other elements in the mesh.

The element equations are formed by assuming known values of properties at fixed points on the elements known as nodes. Then the properties of all the elements and the interaction between them are taken into account by assembling the equations and finding a solution to them. Evaluation of these stresses allows the investigator to determine areas of high stress and large deformations.

The 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). The “equivalent stress of von mises” is an expression that yields an effective absolute magnitude of stresses, taking into account principal stresses in three dimensions.

The basic step for conducting this study can be divided into three phases:

1) Pre processing and modeling
2) Processing and meshing
3) Post processing and analysis

Pre processing

An initial working step in finite element analysis is called as preprocessing. This step essentially involves drafting the geometry of the body to be analyzed. In this case the body consisted of a mandibular posterior segment with canine, second premolar, second molar and the 5-unit fixed partial denture. The periphery of the object was plotted as y, z coordinates and converted as points. These points were recognized by the computer when we key in the values and the periphery of the object was plotted on the computer screen.

Working steps in pre-processing consist of obtaining:

1. Geometric data of the structure to be analyzed.
2. Material property of constituent materials.
3. Loading to which the model is to be subjected.
4. Element type.

Geometric data of the structure to be analyzed

In this study five two-dimensional cross-sectional models were fabricated to represent a missing mandibular first premolar and first molar to perform the computer simulation. Each model consisted of a 5- unit metal ceramic FPD with canine, second premolar, and second molar as abutment teeth supported by simulated periodontal ligament and alveolar bone (cortical and trabecular) structures. A standard intraoral radiographic film was used to trace the geometry for the tooth model and a 5-unit FPD with canine and first molar pontic was designed to represent the rigid model. The bone was modelled as a simplified rectangular configuration. For the two dimensional models it was assumed that enamel was completely removed. A metal thickness of 0.3 mm and a ceramic thickness of 1.2 mm were given for the restorations. The interface between the retainers and their abutments was considered to be rigid. No luting cement was included in the models. The average width of the periodontal ligament and cortical bone were 0.25 mm and 1.5 mm respectively. The lower border of the mandible was considered fixed in all directions to resist the finite element method for the occlusal load.

For the nonrigid models, Rhein 83 attachment which consisted of an OT sphere and cap was designed in the following locations:

1. Distal to the canine.

2. Mesial to second premolar (pier abutment).

3. Distal to second premolar (pier abutment).

4. Mesial to second molar.

The OT sphere and cap was measured using a caliper for dimensions. The models were created in the CATIA-V5R18 software by giving various commands. This model was imported to the ANSYS software through IGES (Initial graphic exchange specification) file for further analysis.

Material property of constituent materials:

Finite element analysis assumes the following mechanical properties of the materials comprising the structure.

1. Homogeneous: mechanical properties of the material are the same throughout each structural element.
2. Isotropic: the material properties are the same in all direction of the structural element.
3. Linearly elastic: the deformations or strains of the structure are proportional to the applied loads.

TABLE III Mechanical properties of materials[48]

Material	Elastic modulus (GPa)	Poisson's Ratio (v)
Feldspathic porcelain	82.8	0.35
NiCr alloy	206	0.33
Dentin	18	0.33
Pulp tissue	0.003	0.45
Periodontal ligament	0.069	0.45
Cortical bone	13.7	0.3
Spongy bone	1.37	0.3
Nylon	2.8	0.4

Element type:

The models of the five FPD's and the supporting structures were meshed with eight node quadrilateral elements.

Processing

In this step, all the relevant informations obtained in the pre-processing stage were taken as the control data. This control data forms the basic unit to be analyzed. The finite element software now employs the inbuilt graphic facilities over the geometric data.

This geometric data was put into meshing. Meshing was done by giving a meshing command to the software. Meshing divides the body into finite number of element with each element having nodes and control points. Loads were applied at the control points and displacement seen at the nodes.

Working steps in processing

1. Setting up of a control data.
2. The different layers of the body to be analyzed are represented as different areas.
3. Computer graphic facility of the finite element software is utilized and meshing is done of the different areas.

The meshing divides the whole geometric body and its layers into finite elements and this is then subjected to analysis. The ANSYS 12.0 software computer program was employed to generate input data for the finite element stress analysis. Geometric and elastic parameters of all components were entered into the computer program.

The data included: (1) total number of nodal points, (2) total number of elements, (3) the numbering system identifying each element, (4) young's modulus and poisson's ratio of each element, (5) the numbering system identifying each nodal point, (6) the coordinates of each nodal point, (7) the type of boundary constraints, and (8) the evaluation of the forces at the external nodes.

From the previously generated models the y and z coordinates were determined. When these y and z coordinates were input into the ANSYS 12.0 software program, the periphery of the models were plotted on the computer screen. After all these coordinates were united appropriately the different layers can be appreciated.

The finite element software on which the model was created meshes the different areas independently. Thus the whole model was divided into different nodes and elements. The model thus created was given life like properties by inducing into the different layers their modulus of elasticity and poisson's ratio.

Modulus of elasticity = stress / strain

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These properties when induced in the respective areas of the model can predict the behavior and stress propagation of the material under testing when a load is given to it.

Loading the prepared model

Loading of the 5-unit fixed partial denture with rigid and nonrigid connector designs with a 50-N[58] static vertical occlusal load on the cuspal fossa of each abutment was used to calculate the stress distributions. Three different loading methods were employed:

1. Loading of all teeth to simulate maximum centric occlusion contacts.
2. Loading of the canine to simulate a single anterior contact.
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For each loading five observations were made in the canine cusp, canine distal cervical, canine root surface, second premolar cusp, second molar mesial cervical, second molar mesial cusp, second molar distal cusp and second molar root surface.

The results thus obtained were taken up for interpretation. The model showed propagation of stresses both numerically and by color coding.

Post processing

Once control data was subjected to analysis by the finite element method software, the results were interpreted. This step consisted of the post processing stage. Stress distribution in the finite element model comes in numerical values and in color coding.

Maximum value of Von Mises stress = denoted by red color

Minimum value of Von Mises stress = denoted by blue color

The in-between values were represented by bluish green, green, greenish yellow and yellowish red in the ascending order of stress distribution.

Working steps in post processing

1. Analysis
2. Interpretation of results both numerically and by color coding

The Von Mises equivalent stress (MPa) in the supporting structures was computed using finite element analysis software. This was performed on all the five models and the Von Mises equivalent stress values obtained were tabulated and analyzed for computation of the results.

METHODOLOGY

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Working steps in processing

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The meshing divides the whole geometric body and its layers into finite elements and this is then subjected to analysis. The ANSYS 12.0 software computer program was employed to generate input data for the finite element stress analysis. Geometric and elastic parameters of all components were entered into the computer program.

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Modulus of elasticity = stress / strain

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Stress = force / area

Strain = change in length / original length

These properties when induced in the respective areas of the model can predict the behavior and stress propagation of the material under testing when a load is given to it.

Loading the prepared model

Loading of the 5-unit fixed partial denture with rigid and nonrigid connector designs with a 50-N[58] static vertical occlusal load on the cuspal fossa of each abutment was used to calculate the stress distributions. Three different loading methods were employed:

1. Loading of all teeth to simulate maximum centric occlusion contacts.
2. Loading of the canine to simulate a single anterior contact.
3. Loading of the second molar to simulate a single posterior contact.

For each loading five observations were made in the canine cusp, canine distal cervical, canine root surface, second premolar cusp, second molar mesial cervical, second molar mesial cusp, second molar distal cusp and second molar root surface.

The results thus obtained were taken up for interpretation. The model showed propagation of stresses both numerically and by color coding.

Post processing

Once control data was subjected to analysis by the finite element method software, the results were interpreted. This step consisted of the post processing stage. Stress distribution in the finite element model comes in numerical values and in color coding.

Maximum value of Von Mises stress = denoted by red color

Minimum value of Von Mises stress = denoted by blue color

The in-between values were represented by bluish green, green, greenish yellow and yellowish red in the ascending order of stress distribution.

Working steps in post processing

1. Analysis
2. Interpretation of results both numerically and by color coding

The Von Mises equivalent stress (MPa) in the supporting structures was computed using finite element analysis software. This was performed on all the five models and the Von Mises equivalent stress values obtained were tabulated and analyzed for computation of the results.

RESULTS

This finite element study was performed to evaluate the stress distribution in fixed partial dentures with pier abutments using rigid and a different orientation of nonrigid connector design in five locations. According to the area of loading, three groups were formed. For each group five observations were made in the region of canine cusp, canine distal cervical, canine root, second premolar cusp, second molar mesial cervical, second molar mesial cusp, second molar distal cusp and second molar distal root region.

The results of this study are shown from Table 1 to Table 18. In this study, the maximum value of von mises stress in mega pascal was calculated in the 5-unit metal ceramic FPD and supporting structures.

- Table 1 shows the various models with respect to type and location of connectors.

- Table 2 shows the number of elements and nodes used in the finite element models for rigid and nonrigid design.

- Table 3-7(Annexure) shows the maximum value of von mises stress in the supporting structures for the five models when canine tooth was loaded to simulate anterior loading.

- Table 8-12(Annexure) shows the maximum value of von mises stress in the supporting structures when second molar was loaded to simulate posterior loading.

- Table 13-17(Annexure) shows the maximum value of von mises when all teeth were loaded to simulate maximum centric occlusion contacts.

- Table 18 shows the mean, standard deviation using ANOVA (analysis of variance) test for group I.

- Table 19 shows the mean, standard deviation using ANOVA test for group II.

- Table 20 shows the mean and standard deviation using ANOVA test for group III.

- Table 21(Annexure) shows the results of statistical evaluation using Tukey HSD (highly significant differential) test for multiple comparisons within group I.

- Table 22(Annexure) shows the results of statistical evaluation using Tukey HSD test for multiple comparisons within group II.

- Table 23(Annexure) shows the results of statistical evaluation using Tukey HSD test in group III

The basic data obtained after finite element analysis for this study is presented in Table 18 to Table 23

INFERENCE FROM TABLE 18

The maximum value of von mises stress was found to be statistically significant with a probability value <0.001 for the five models in all the observed areas.

When comparing all the five models after loading the canine tooth with a static vertical occlusal load of 50N, model A exhibited maximum stress distribution followed by model E. Model D exhibited minimum stress distribution on the pier abutment.

INFERENCE FROM TABLE 19

The maximum value of von mises was found to be statistically significant with a probability value <0.001 in all the observed areas for the five models when the second molar was loaded with a static vertical occlusal load of 50N. Model A showed maximum stress distribution followed by model B, model C and model D. Minimum stress distribution was shown by model E.

INFERENCE FROM TABLE 20

The maximum value of von mises stress was statistically significant with a probability value of <0.001. When all teeth were loaded with a static vertical occlusal load of 50N simulating maximum centric occlusion contacts, model A exhibited maximum stress distribution followed by model E. The stress values for model B and C were almost similar and less when compared with model E. Minimum stress distribution was exhibited by model D.

INFERENCE FROM TABLE 21

While comparing the stress distribution within the group for the various models in group I, the mean difference was found to be significant at 1% level.

INFERENCE FROM TABLE 22

When comparing the stress distribution within the group for the five models in group II, the mean difference was significant at1% level

INFERENCE FROM TABLE 23

On multiple comparison analysis of the five models in group III, the mean difference was significant at 0.05 level.

TABLE 1

MODELS	TYPE AND LOCATION OF CONNECTOR
MODEL A	Rigid connector
MODEL B	Non rigid connector distal to canine
MODEL C	Non rigid connector mesial to second premolar
MODEL D	Non rigid connector distal to second premolar
MODEL E	Non rigid connector mesial to second molar

TABLE 2

MODELS	ELEMENTS	NODES
Rigid	9191	11852
Non rigid	9540	12320

TABLE 3

GROUP I- Loading of canine tooth One way ANOVA

	Model										P value
	Model A		Model B		Model C		Model D		Model E		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Canine Cusp	4.310	.010	4.306	.011	4.334	.005	4.134	.011	4.342	.004	<0.001**
Canine Distal Cervical	3.140	.020	3.604	.009	3.808	.011	2.144	.009	2.068	.011	
Second Pre Molar cusp	2.390	.014	.	.	.	.	1.808	.023	2.028	.018	
Second Molar Mesial Cervical	1.748	.020	.	.	.	.	.	.	.	.	
Second Molar Mesial Cusp	2.244	.021	.	.	.	.	.	.	.	.	
Second Molar Distal Cusp	1.468	.018	.	.	.	.	.	.	.	.	
Canine Root	.880	.045	1.232	.022	1.342	.032	1.254	.009	1.148	.011	
Second Molar distal Root	.788	.019	.	.	.	.	.	.	.	.	

** denotes significant at 1% level

TABLE 4

GROUP II- Loading of second molar One way ANOVA

	Model										P value
	Model A		Model B		Model C		Model D		Model E		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Canine Cusp	4.348	.018	.	.	.	.	.	.	.	.	<0.001**
Canine Distal Cervical	3.348	.016	.	.	.	.	.	.	.	.	
Second Pre Molar	4.388	.008	3.184	.032	2.776	.026	.	.	.	.	
Second Molar Mesial Cervical	3.918	.015	2.452	.025	2.202	.004	2.098	.018	.	.	
Second Molar Mesial Cusp	4.414	.022	4.290	.024	4.136	.022	4.414	.026	4.094	.009	
Second Molar Distal Cusp	3.838	.018	3.448	.033	3.688	.018	3.524	.034	2.796	.015	
Canine Root	3.298	.011	.	.	.	.	.	.	.	.	
Second Molar distal Root	3.580	.020	.626	.019	1.840	.014	1.084	.036	1.394	.005	

******denotes significant at 1% level

TABLE 5

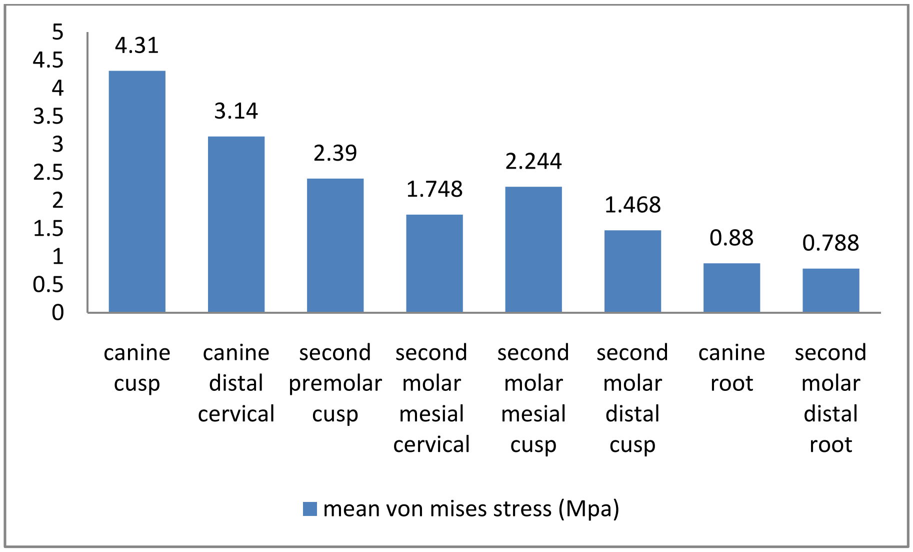
GROUP III- Loading all teeth One way ANOVA

	Model										P value
	Model A		Model B		Model C		Model D		Model E		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Canine Cusp	4.508	.011	4.508	.011	4.516	.015	4.436	.026	4.514	.009	<0.001**
Canine Distal Cervical	4.096	.025	4.480	.019	2.774	.013	2.426	.015	2.768	.011	
Second Pre Molar	4.604	.017	4.588	.011	4.588	.011	3.664	.017	4.574	.005	
Second Molar Mesial Cervical	4.118	.013	4.112	.011	2.778	.004	1.828	.018	2.920	.014	
Second Molar Mesial Cusp	4.622	.013	4.604	.009	4.600	.014	3.714	.017	3.668	.011	
Second Molar Distal Cusp	4.502	.015	4.510	.007	4.378	.015	2.756	.009	3.544	.005	
Canine Root	.936	.017	1.436	.025	1.382	.013	.932	.018	.952	.018	
Second Molar distal Root	.980	.014	.992	.011	1.402	.013	.936	.017	.932	.011	

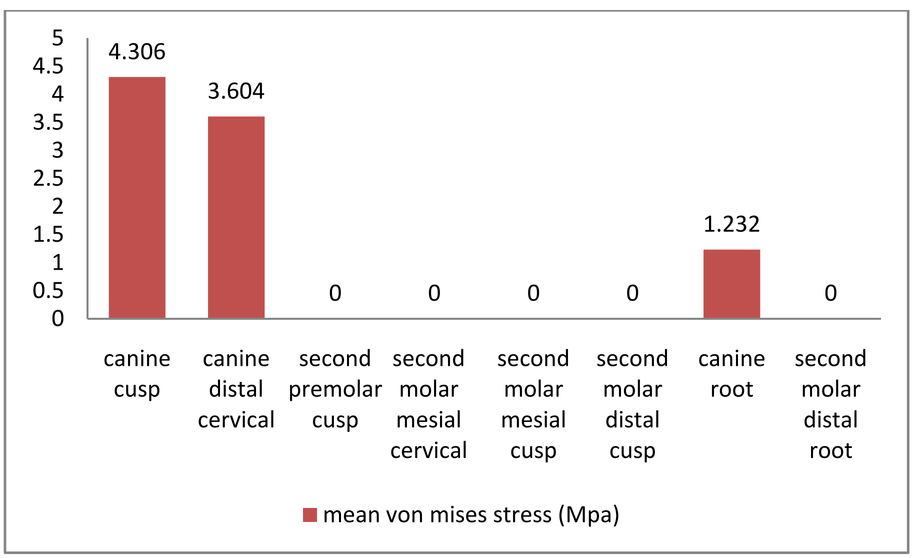
**denotes significant at 1% level

GROUP I – Loading of canine tooth

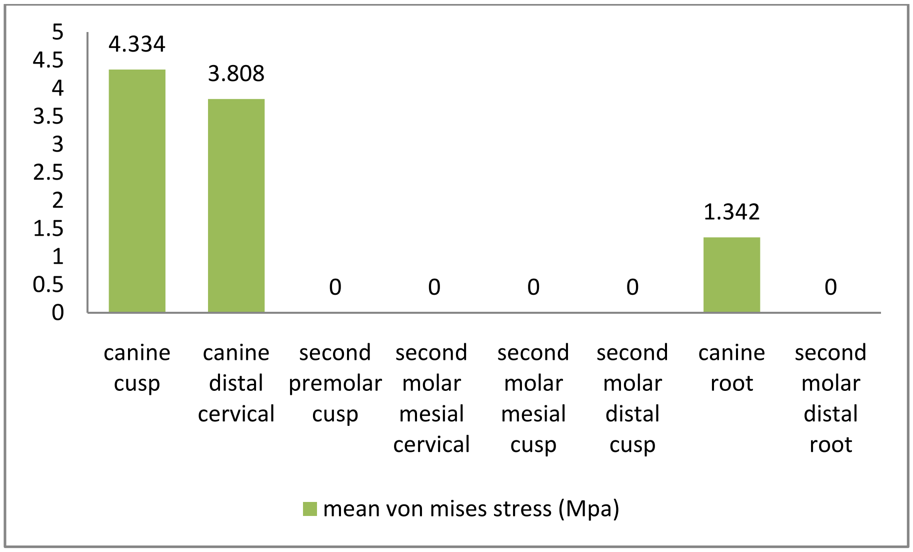
MODEL A- Rigid connector design



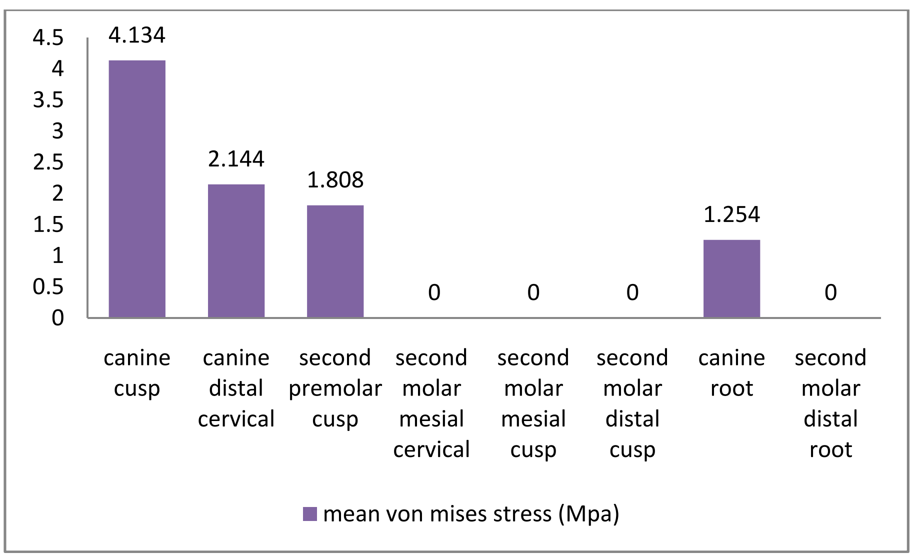
MODEL B- Nonrigid connector distal to canine



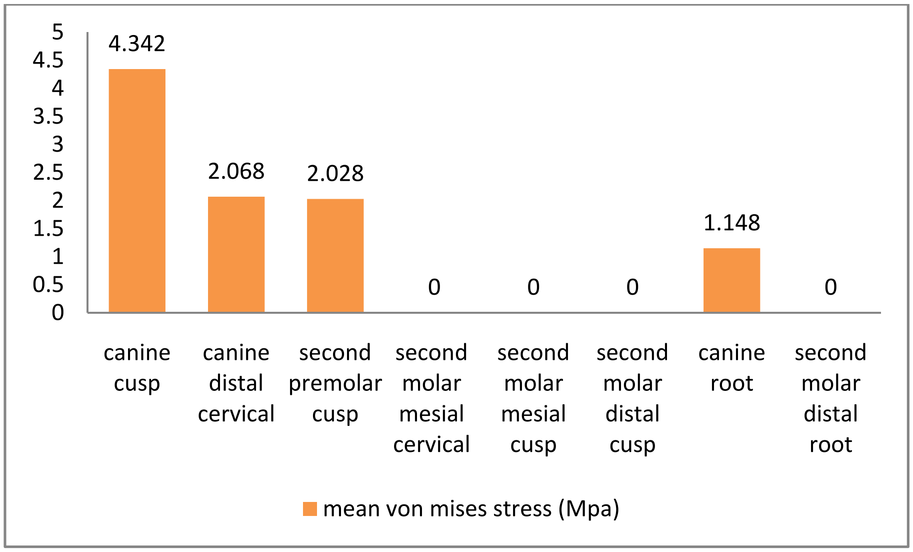
MODEL C – Nonrigid connector mesial to second premolar



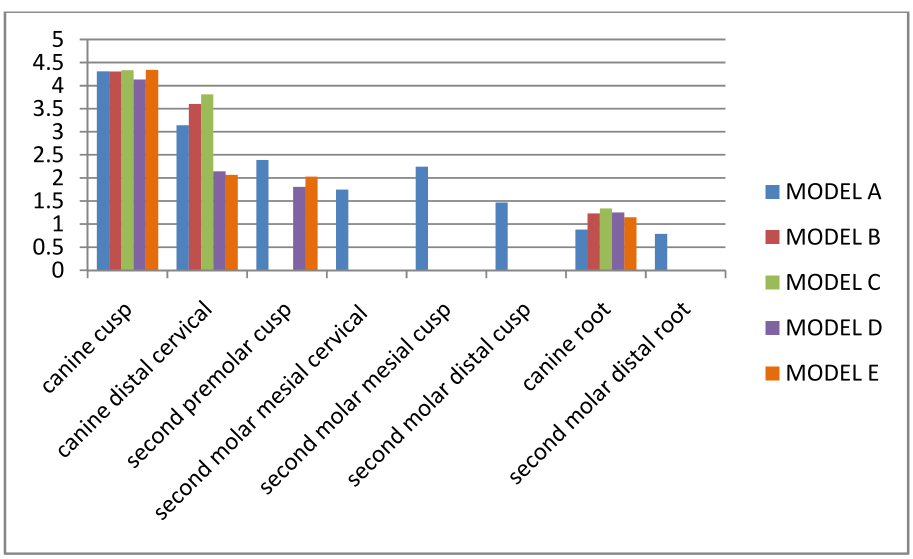
MODEL D – Nonrigid connector distal to second premolar



MODEL E – Nonrigid connector mesial to second molar

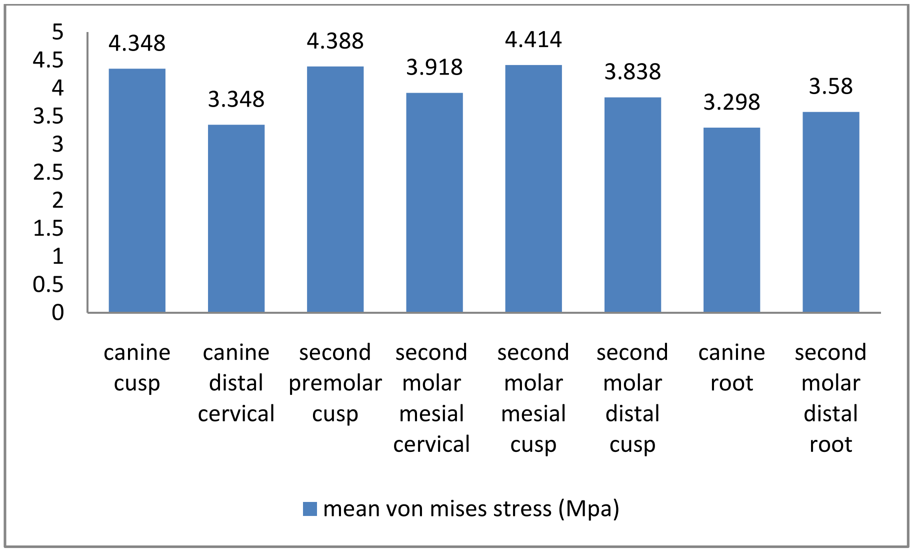


COMPARATIVE VALUEOF VON MISES STRESS WITH CANINE LOADING

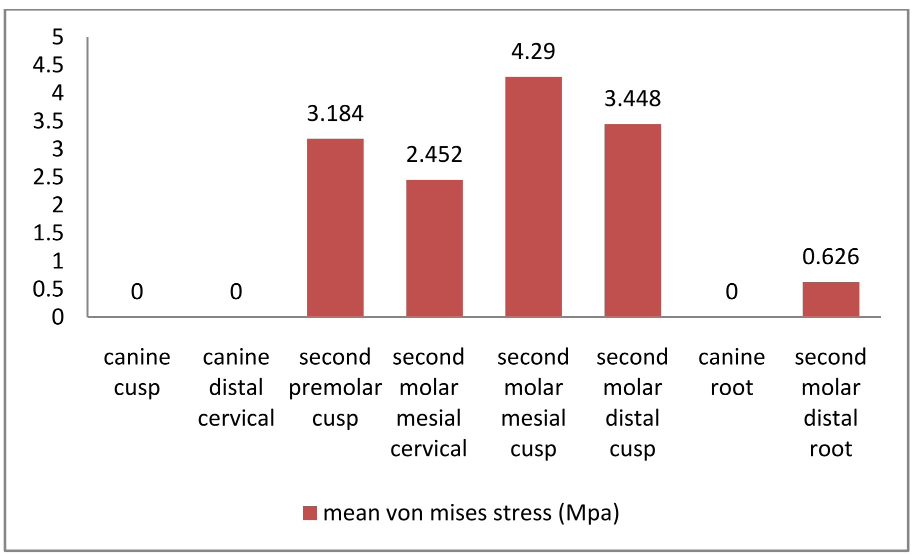


GROUP II – Loading of second molar tooth

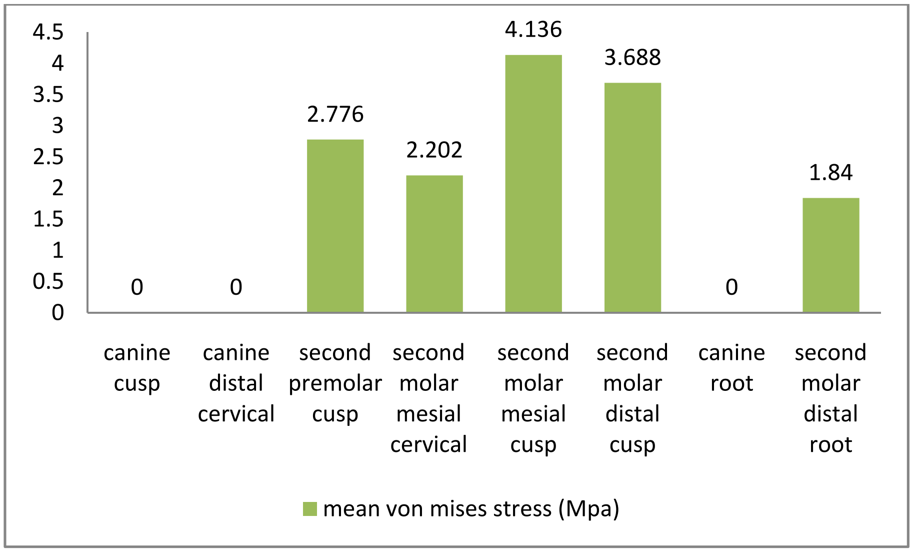
MODEL A – Rigid connector design



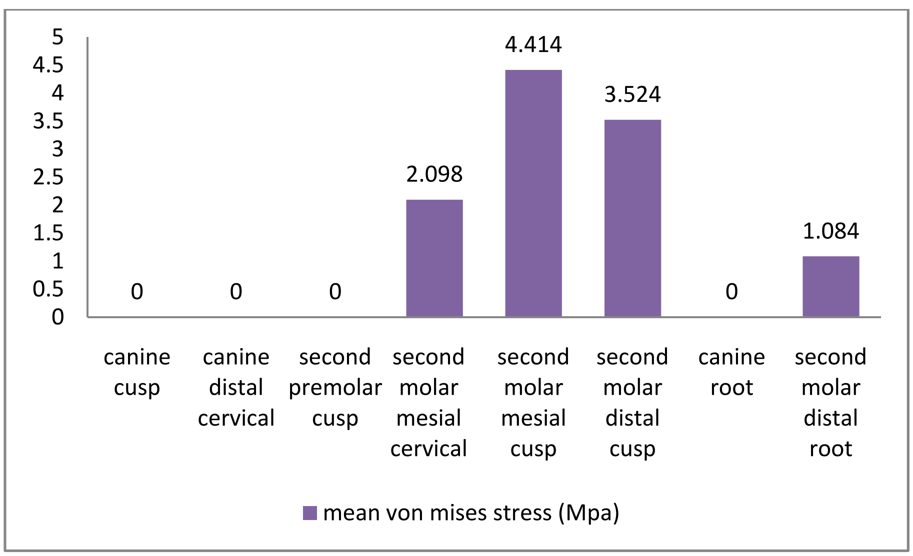
MODEL B – Nonrigid connector distal to canine



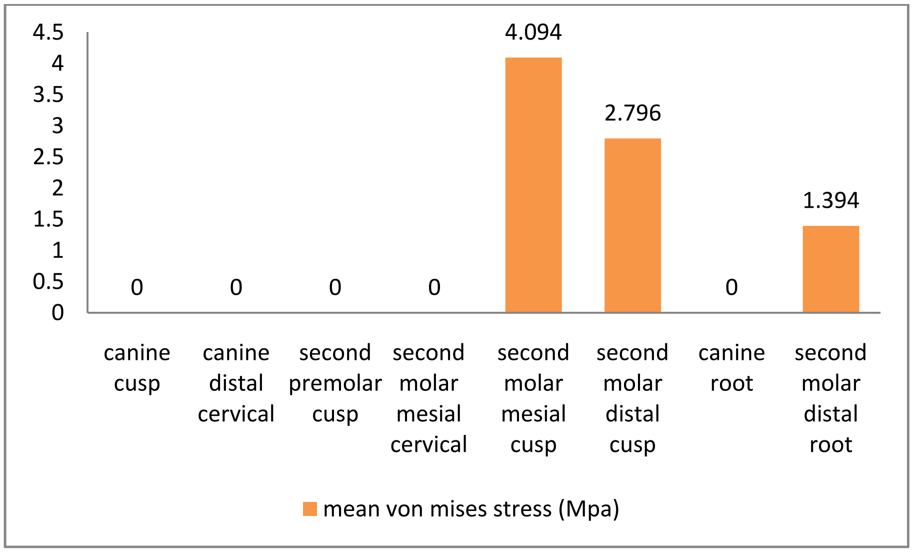
MODEL C – Nonrigid connector mesial to second premolar



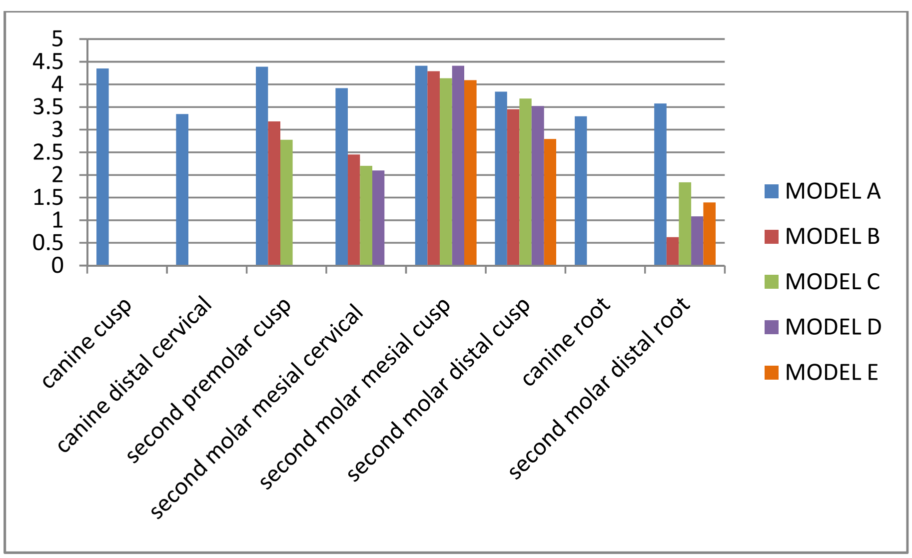
MODEL D – Nonrigid connector distal to second premolar



MODEL E – Nonrigid connector mesial to second molar

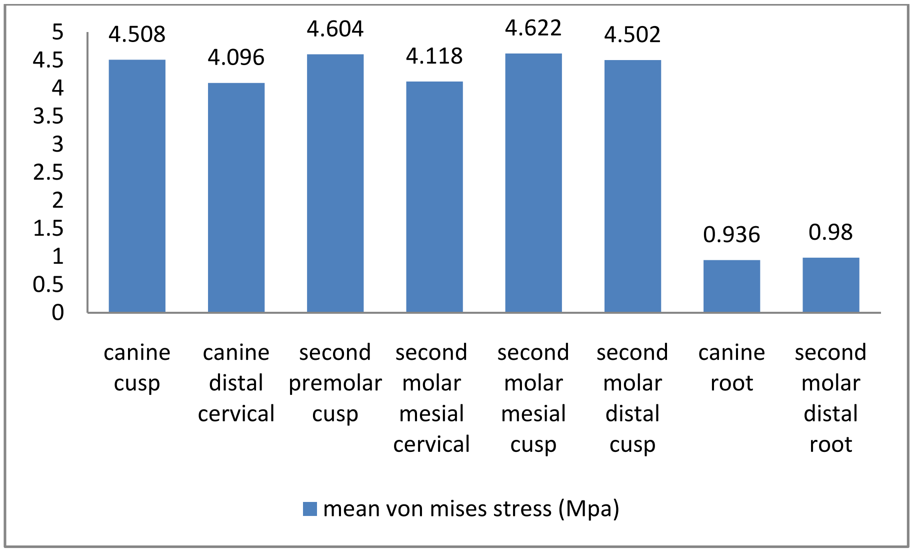


COMPARATIVE VALUES OF VON MISES STRESS WITH POSTERIOR LOADING

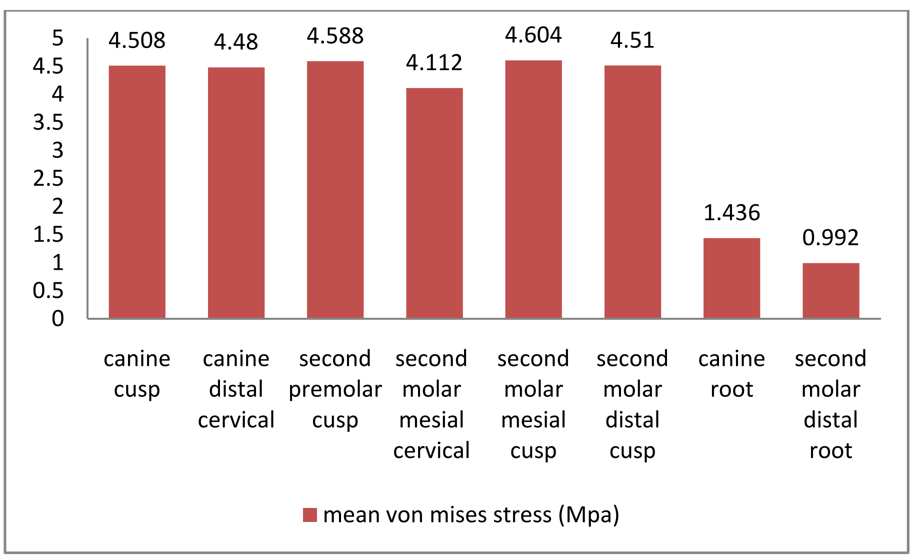


GROUP III – Complete loading of all teeth

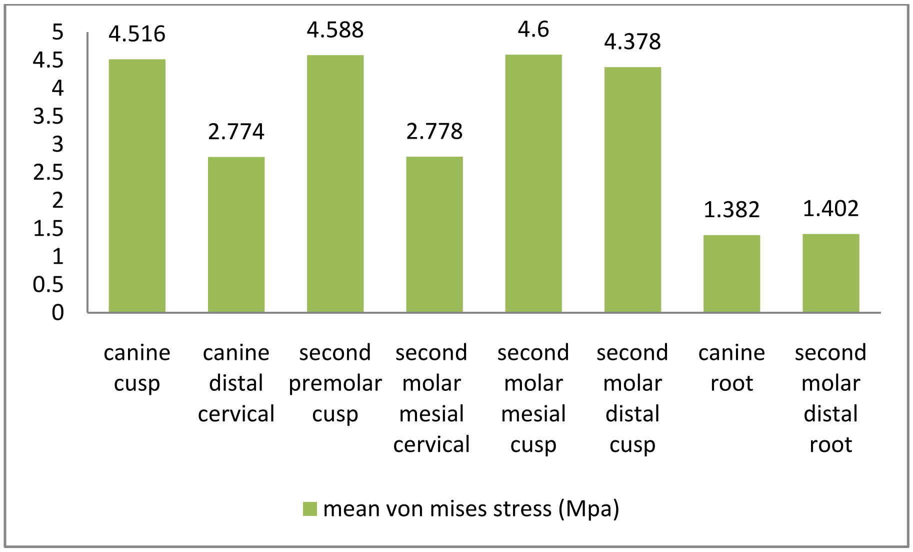
MODEL A – Rigid connector design



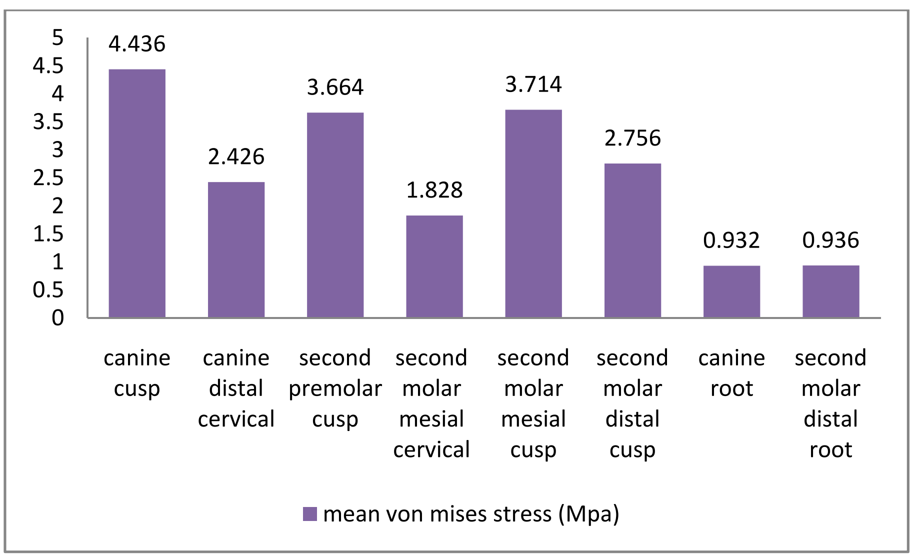
MODEL B – Nonrigid connector distal to canine



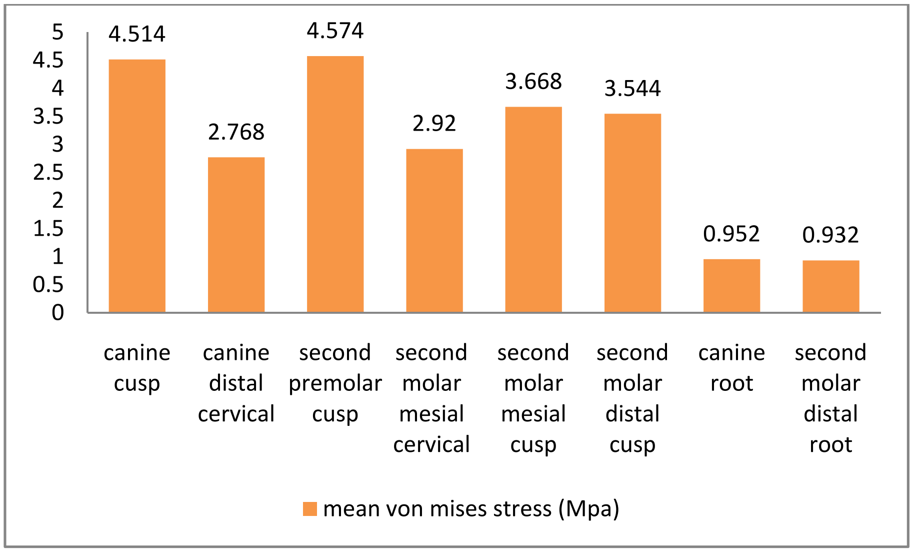
MODEL C – Nonrigid connector mesial to second premolar



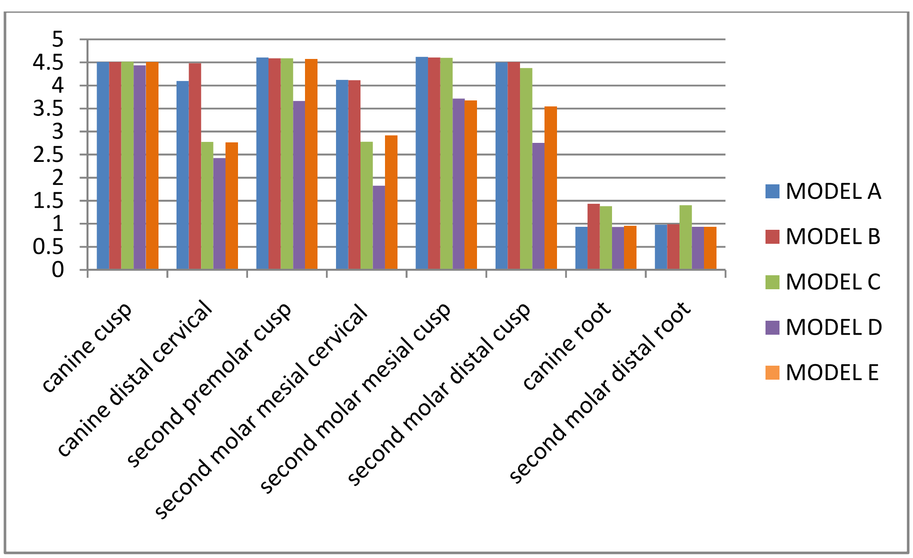
MODEL D – Nonrigid connector distal to second premolar



MODEL E – Nonrigid connector mesial to second molar



COMPARATIVE ANALYSIS OF VON MISES STRESS WITH COMPLETE LOADING



DISCUSSION

In fixed partial denture fabrication, rigid connectors (solder joints) are usually used between pontics and retainers since it provides the desirable strength and stability to the prosthesis while minimizing the stresses associated with the restoration. However, a completely rigid restoration is not indicated for all situations requiring a fixed prosthesis[6]. When it is not possible to prepare two abutments with a common path of placement, a nonrigid connector is indicated. Similarly in a mandibular arch when a complex fixed partial denture consists of anterior and posterior segments a nonrigid connector is used. Rigid fixed partial dentures have been shown to inhibit mandibular flexure, and extensive splints have been shown to flex during forced opening. The associated stresses can cause dislodgement of complex fixed partial dentures[52].

Because of the curvature of the arch, the faciolingual movement of an anterior tooth occurs at a considerable angle to the faciolingual movement of a molar. Studies in periodontometry have shown that the faciolingual movement ranges from 56 to 108μm, and intrusion is 28μm.Teeth in different segments of the arch move in different directions. These movements can create stresses in a long span prosthesis that will be transferred to the abutments as well as between retainers and abutment preparations[6].

Another situation is when edentulous spaces exist on both sides of a tooth creating a lone, free standing pier abutment. A rigid five unit fixed partial denture in such a case is not ideal because of the physiological tooth movement, arch position of the abutments and a disparity in the retentive capacity of the retainers[6]. The existence of “pier” (middle) abutments, promote a fulcrum-like situation that can cause the weakest of the terminal abutments to fail[6].

Standlee and Caputo[53] reported that when an occlusal load is applied to the retainer on the abutment tooth at one end of a fixed partial denture with a pier abutment, the pier abutment may act as a fulcrum. Thus tensile forces may be generated between the retainer and abutment at the other end of the restoration.

Botelho MG, Dyson JE[54] reported that rigid long- span fixed partial dentures are associated with higher debonding rates than short-span prostheses.

Lin et al[44] reported that a nonrigid connector has the ability to separate the splinted units.

Herman E.S.Chayes[55] states that an ideal attachment is one which while providing adequate stability for the appliance would at the same time permit the physiologic movement of both appliance and abutment teeth.

Thus, the use of a nonrigid connector has been suggested and is commonly used with fixed partial dentures.

Nonrigid connectors are classified into intracoronal or extracoronal, custom made or prefabricated[56]. The intracoronal attachments used in fixed partial dentures are generally incorporated entirely within the contour of the crown. So more tooth reduction is required for the attachment to be placed within the confines of the crown and secondly atleast 3mm of height is needed for an attachment to be effective. A consequence of not having this 3mm minimum height is lack of proper embrasures that can result in poor esthetics, compromised oral hygiene and periodontal health[56].

In situations requiring nonrigid connectors most of the clinicians prefer the custom made dovetail type with the keyway within the abutment and key within the pontic. If the key and keyway are not aligned properly it can cause added stress on the pier abutment[56].

Hence a study was performed to evaluate the stress distribution in a fixed partial denture with pier abutment using a prefabricated nonrigid connector design.

In this study an extracoronal attachment from Rhein-83[57] was employed to design the nonrigid connector. Here the key is attached to the abutment tooth and the keyway lies within the pontic. Rhein 83 comes in two types – vertical, horizontal. Here horizontal attachment was used assuming that it will reduce the stress to the abutment.

Rhein 83 attachments have elastic retention so as to eliminate the phenomena of friction. They are neither rigid nor shock absorbing. With elasticity it is possible to control the flexure and construct resilient and shock absorbing prostheses. When these attachments are used in rigid prostheses, the sphere and cap function as a retentive button coupling. Therefore their function will be maintained in a stable position. The elastic materials permit a wide area of retention in the equatorial region of the sphere. The retentive area is surpassed from the cap; the cap will assume its initial shape, on a wide area of the sphere. It will compress and return. The space between the flat sphere and the elastic cap reduces the stresses due to vertical flexion.

In resilient prostheses, they function with a cushion effect, like a shock absorber, due to the flattened head of the sphere and to the retentive elastic caps.

The biomechanical aspects are difficult to evaluate using clinical observation/experimental approaches with limited information and sample variations. ***Farah et al***[8] introduced finite element method (FEM) study in dentistry for the first time, proving its efficiency to be better than photoelastic study in terms of easy modeling and more defined stress analysis. Therefore, finite element analysis (FEA) has generally been accepted as a complementary tool for understanding the detailed mechanical responses for any biologic investigations[7]. The results of the FEA computation depend on many individual factors including material properties, boundary conditions, and interface definitions and also on the overall approach to the model[7].

A 3-dimensional model, although more realistic, would have resulted in coarser meshes due to the computer capacity, which would not have allowed the fine representation of a thin periodontal ligament and cortical bone[30]. Hence in this study a two dimensional finite element model was generated to simulate the bone, abutment teeth and the five unit fixed partial denture. The assumption required for analysis of stress distribution by using a two dimensional finite element method was that stresses along the buccolingual direction were negligible[30]. The structures in the model were all assumed to be homogeneous, isotropic and to possess linear elasticity. Considering the previously mentioned limitations, this finite element analysis only approximated the clinical situation, while a qualitative comparison between the models was performed, rather than focusing on quantitative data.

Finite element models were simulated using CATIA V5R18 software by giving various commands. The results were analyzed and interpreted using ANSYS software through IGES (Initial Graphic Exchange Specification) file.

In this study for the rigid design (model A), when the canine tooth was loaded with a static vertical force of 50N[58], maximum stress was evident in the region of canine cusp (4.31MPa) and a low stress concentration in the second molar cusp (2.244MPa) and apical third of distal root(0.788 MPa). With posterior loading, second molar mesial cusp (4.414MPa) showed maximum stress concentration. The mesial terminal abutment also showed a comparable amount of stress along the cusp and cervical regions. When all teeth were loaded, maximum stress concentration was observed at the cusp tips and cervical region of abutments. Stress on the pier abutment was 4.604 MPa.

With the nonrigid connector distal to canine (model B), anterior loading showed maximum stress concentration in the cusp, cervical region and root surface of the mesial terminal abutment only. Posterior loading showed no stresses on the anterior abutment. When all teeth were loaded, the stress distribution was almost similar to that of the rigid design with a stress concentration of 4.588MPa on the pier abutment

When the anterior terminal abutment was loaded, the nonrigid connector at the mesial region of the pier abutment (model C) showed more stress on the anterior tooth when compared to model A and B. Posterior loading was similar to that of model B but the amount of stress concentration on the pier abutment was less. With complete loading stress on the pier abutment was comparable to model B (4.588MPa).

The nonrigid connector design at the mesial region of second molar tooth (model E) showed no stress concentration on the posterior terminal abutment with anterior loading and less stress distribution on the pier abutment (2.028MPa) when compared with the rigid design (2.390MPa). Posterior loading showed stress concentration only on the posterior terminal abutment. With complete loading the results were almost similar to that of the rigid design but the stress on the pier abutment (4.574MPa) was slightly less.

For the nonrigid connector design distal to the pier abutment (model D), anterior loading showed only less stress distribution on the anterior terminal abutment and pier abutment (1.808MPa) when compared to the rigid design and the nonrigid designs in other locations. Posterior loading was similar to that of model E. On loading all cusps stress concentration on the pier abutment (3.664MPa) was found to be the least when compared with the other models.

Data was analyzed with one way ANOVA (analysis of variance) and Tukey highly significant differential tests. The maximum value for von mises stress was found to be statistically significant with a probability value of <0.001 for all the models .The results support rejection of the null hypothesis.

Moulding MB et al[58] in his study reported that the stress fields change depending on the location of nonrigid connectors. Also the authors stated that the rigid fixed partial denture distributes stresses vertically and evenly, and the nonrigid connectors located at the distal of the canine and at the mesial of molar designs distribute stresses evenly.

Sutherland JK et al[11] conducted a photoelastic analysis of the stress distribution in bone supporting fixed partial dentures of rigid and nonrigid design. It was concluded that under conditions of the vertical loading, the rigid fixed partial denture design does not permit independent response by either abutment. The nonrigid fixed partial denture design allows some independence in response to the vertical loading. He also reported that rigid and nonrigid connectors exhibit differences in stress distributions and concentrations within the supporting bone structure.

Selcuk Oruc et al[48] in a study comparing the stress distribution with the nonrigid connectors assigned as free (non bonded) touching faces for different design types reported that high stress values were located at the connectors and cervical regions of abutment teeth, especially at the pier abutment. However, with the use of a nonrigid connector at the distal region of the pier abutment, the area of maximum concentration for the pier abutment was reduced.

The present study also correlates with the previous studies[11, 48] showing minimum stress distributions in the pier abutment with complete loading of all cusps when the nonrigid connector was positioned distal to second premolar. Also, the stress concentration on anterior abutment was minimized with posterior loading and vice versa. This may be an indication of the nonrigid design's influence on prevention of the lever effect with five unit prosthesis.

In the study conducted earlier by ***Selcuk Oruc etal***[48] with the nonrigid connector distal to pier abutment and a static vertical load applied to the cusp tips of all teeth, root surfaces and apical areas of the pier abutment showed less stress concentration and a relatively high stress concentration was observed along the mesial root surface of the distal terminal abutment.

In this study a newly designed nonrigid attachment at the distal side of second premolar was simulated and a static vertical load was applied to the cusp fossa of all the teeth. Several assumptions were made for the model used in this study regarding simulated structures as mentioned above[30]. On focusing the stress pattern on the pier abutment, the stress value was found to be nil on the root surface of pier abutment and a minimum stress on the root surface of the distal terminal abutment when compared to other locations.

This study also shows minimal stress pattern when the nonrigid connector was placed in the posterior locations that is; distal to second premolar and mesial to second molar when compared to anterior locations like distal to canine and mesial to second premolar[58].

Future studies need to be conducted to compare the stress distribution with an axial and oblique loading on the cusp and other areas on the occlusal surfaces. Further clinical trials can ultimately confirm the predictions made from finite element analysis presented here.

SUMMARY AND CONCLUSION

This study was done to evaluate the stress distribution in fixed partial dentures with pier abutments using rigid and a different orientation of nonrigid connector designs. A finite element study was done to determine the amount of stress distribution and to find out the ideal location of nonrigid connector. Finite element models were simulated using CATIA V5R18 software by giving various commands.

All the five models were loaded with a static vertical occlusal force of 50N to the cusp fossa to simulate anterior, posterior and complete loading. The results were analyzed and interpreted using ANSYS software through IGES (Initial Graphic Exchange Specification) file. The data obtained were tabulated and statistical analysis was done.

The findings of the present study support the following conclusions.

1. The stress distribution in fixed partial denture with pier abutment was affected by the presence and location of a nonrigid connector.
2. The area of minimum stress concentration occurs in pier abutments when the nonrigid connector was positioned distal to the pier abutment.

3. The overall stress distribution on the pier abutment was decreased when a nonrigid connector with an elastic cap was used.

Finite element analysis suffers from several limitations, mostly related to necessarily simplified assumptions due to the lack of information about material properties, uncertainty of correct load distribution, assignment of the proper boundary conditions and creation of a valid mesh[30].

Future scope needs to be on research and development coupled with controlled, prospective clinical studies to guide the clinician in near future.

ANNEXURE

GROUP I

TABLE-3

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.32	3.16	2.40	1.73	2.25	1.46	0.9	0.79
B	4.31	3.62	_	_	_	_	1.22	_
C	4.33	3.80	_	_	_	_	1.34	_
D	4.13	2.15	1.80	_	_	_	1.26	_
E	4.35	2.07	2.04	_	_	_	1.16	_

TABLE 4

MODEL	Canine cusp	Canine distal cervical	second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.30	3.12	2.38	1.73	2.21	1.48	0.9	0.78
B	4.31	3.60	_	_	_	_	1.21	_
C	4.34	3.80	_	_	_	_	1.34	_
D	4.13	2.13	1.78	_	_	_	1.25	_
E	4.34	2.07	2.04	_	_	_	1.14	_

TABLE 5

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.30	3.16	2.40	1.77	2.26	1.48	0.9	0.78
B	4.32	3.60	_	_	_	_	1.22	_
C	4.33	3.82	_	_	_	_	1.34	_
D	4.12	2.15	1.80	_	_	_	1.24	_
E	4.34	2.05	2.02	_	_	_	1.14	_

TABLE 6

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.32	3.12	2.40	1.74	2.26	1.44	0.9	0.82
B	4.30	3.60	_	_	_	_	1.26	_
C	4.34	3.82	_	_	_	_	1.39	_
D	4.14	2.15	1.84	_	_	_	1.26	_
E	4.34	2.08	2.04	_	_	_	1.16	_

TABLE 7

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.31	3.14	2.37	1.77	2.24	1.48	0.8	0.77
B	4.29	3.60	_	_	_		1.25	
C	4.33	3.80	_	_			1.30	
D	4.15	2.14	1.82	_			1.26	
E	4.34	2.07	2.00	_			1.14	

GROUP II

TABLE 8

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.36	3.35	4.39	3.91	4.40	3.82	3.30	3.56
B			3.15	2.43	4.28	3.45		0.61
C			2.76	2.21	4.12	3.68		1.84
D				2.08	4.40	3.51		1.04
E					4.10	2.78		1.39

TABLE 9

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.32	3.32	4.38	3.92	4.40	3.82	3.28	3.56
B			3.15	2.43	4.26	3.42		0.60
C			2.76	2.20	4.12	3.66		1.82
D				2.08	4.38	3.48		1.06
E					4.09	2.78		1.39

TABLE 10

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.36	3.36	4.39	3.92	4.42	3.84	3.30	3.58
B			3.20	2.45	4.28	3.42		0.64
C			2.82	2.20	4.12	3.70		1.84
D				2.10	4.42	3.52		1.08
E					4.10	2.81		1.39

TABLE 11

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.36	3.36	4.40	3.94	4.40	3.86	3.31	3.60
B			3.22	2.49	4.31	3.45		0.64
C			2.76	2.20	4.16	3.70		1.86
D				2.11	4.45	3.57		1.13
E					4.08	2.81		1.40

TABLE 12

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.34	3.35	4.38	3.90	4.45	3.85	3.30	3.60
B			3.20	2.46	4.32	3.50		0.64
C			2.78	2.20	4.16	3.70		1.84
D				2.12	4.42	3.54		1.11
E					4.10	2.80		1.40

GROUP III

TABLE 13

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.52	4.07	4.62	4.11	4.63	4.51	0.92	0.98
B	4.50	4.49	4.60	4.12	4.62	4.51	1.41	0.98
C	4.51	2.78	4.60	2.78	4.60	4.37	1.39	1.39
D	4.42	2.41	3.68	1.84	3.71	2.76	0.92	0.92
E	4.52	2.78	4.57	2.92	3.68	3.55	0.94	0.92

TABLE 14

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.50	4.07	4.58	4.11	4.60	4.48	0.92	0.98
B	4.50	4.46	4.58	4.10	4.60	4.50	1.46	1.00
C	4.51	2.76	4.58	2.78	4.58	4.40	1.39	1.42
D	4.40	2.43	3.66	1.84	3.70	2.76	0.92	0.96
E	4.50	2.76	4.57	2.92	3.66	3.54	0.94	0.94

TABLE 15

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.50	4.12	4.62	4.14	4.63	4.50	0.94	1.00
B	4.52	4.49	4.58	4.12	4.60	4.52	1.41	1.00
C	4.54	2.76	4.58	2.78	4.62	4.38	1.38	1.39
D	4.46	2.44	3.68	1.80	3.72	2.74	0.92	0.92
E	4.52	2.78	4.58	2.90	3.66	3.55	0.94	0.94

TABLE 16

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.52	4.12	4.60	4.12	4.62	4.52	0.96	0.98
B	4.50	4.46	4.60	4.12	4.60	4.51	1.44	1.00
C	4.52	2.78	4.58	2.77	4.60	4.38	1.36	1.41
D	4.46	2.44	3.64	1.84	3.74	2.76	0.96	0.94
E	4.52	2.76	4.57	2.94	3.68	3.54	0.98	0.94

TABLE 17

MODEL	Canine cusp	Canine distal cervical	Second premolar cusp	Second molar mesial cervical	Second molar mesial cusp	Second molar distal cusp	Canine root surface	Second molar distal root
A	4.50	4.10	4.60	4.11	4.63	4.50	0.94	0.96
B	4.52	4.50	4.58	4.10	4.60	4.51	1.46	0.98
C	4.50	2.79	4.60	2.78	4.60	4.36	1.39	1.40
D	4.44	2.41	3.66	1.82	3.70	2.76	0.94	0.94
E	4.51	2.76	4.58	2.92	3.66	3.54	0.96	0.92

TABLE 21 – MULTIPLE COMPARISONS (TUKEY HSD) for GROUP I

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Canine Cusp	Model A	Model B	.0040*	.00573	.004	-.0131	.0211
		Model C	-.0240(*)	.00573	.004	-.0411	-.0069
		Model D	.1760(*)	.00573	.000	.1589	.1931
		Model E	-.0320(*)	.00573	.000	-.0491	-.0149
	Model B	Model A	-.0040*	.00573	.004	-.0211	.0131
		Model C	-.0280(*)	.00573	.001	-.0451	-.0109
		Model D	.1720(*)	.00573	.000	.1549	.1891
		Model E	-.0360(*)	.00573	.000	-.0531	-.0189
	Model C	Model A	.0240(*)	.00573	.004	.0069	.0411
		Model B	.0280(*)	.00573	.001	.0109	.0451
		Model D	.2000(*)	.00573	.000	.1829	.2171
		Model E	-.0080*	.00573	.000	-.0251	.0091
	Model D	Model A	-.1760(*)	.00573	.000	-.1931	-.1589
		Model B	-.1720(*)	.00573	.000	-.1891	-.1549
		Model C	-.2000(*)	.00573	.000	-.2171	-.1829
		Model E	-.2080(*)	.00573	.000	-.2251	-.1909
	Model E	Model A	.0320(*)	.00573	.000	.0149	.0491
		Model B	.0360(*)	.00573	.000	.0189	.0531
		Model C	.0080*	.00573	.000	-.0091	.0251
		Model D	.2080(*)	.00573	.000	.1909	.2251
Canine DC	Model A	Model B	-.4640(*)	.00800	.000	-.4879	-.4401
		Model C	-.6680(*)	.00800	.000	-.6919	-.6441
		Model D	.9960(*)	.00800	.000	.9721	1.0199
		Model E	1.0720(*)	.00800	.000	1.0481	1.0959
	Model B	Model A	.4640(*)	.00800	.000	.4401	.4879
		Model C	-.2040(*)	.00800	.000	-.2279	-.1801
		Model D	1.4600(*)	.00800	.000	1.4361	1.4839
		Model E	1.5360(*)	.00800	.000	1.5121	1.5599
	Model C	Model A	.6680(*)	.00800	.000	.6441	.6919
		Model B	.2040(*)	.00800	.000	.1801	.2279
		Model D	1.6640(*)	.00800	.000	1.6401	1.6879
		Model E	1.7400(*)	.00800	.000	1.7161	1.7639
	Model D	Model A	-.9960(*)	.00800	.000	-1.0199	-.9721
		Model B	-1.4600(*)	.00800	.000	-1.4839	-1.4361
		Model C	-1.6640(*)	.00800	.000	-1.6879	-1.6401
		Model E	.0760(*)	.00800	.000	.0521	.0999
	Model E	Model A	-1.0720(*)	.00800	.000	-1.0959	-1.0481
		Model B	-1.5360(*)	.00800	.000	-1.5599	-1.5121
		Model C	-1.7400(*)	.00800	.000	-1.7639	-1.7161
		Model D	-.0760(*)	.00800	.000	-.0999	-.0521

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Canine Root	Model A	Model B	-.3520(*)	.01718	.000	-.4034	-.3006
		Model C	-.4620(*)	.01718	.000	-.5134	-.4106
		Model D	-.3740(*)	.01718	.000	-.4254	-.3226
		Model E	-.2680(*)	.01718	.000	-.3194	-.2166
	Model B	Model A	.3520(*)	.01718	.000	.3006	.4034
		Model C	-.1100(*)	.01718	.000	-.1614	-.0586
		Model D	-.0220*	.01718	.004	-.0734	.0294
		Model E	.0840(*)	.01718	.001	.0326	.1354
	Model C	Model A	.4620(*)	.01718	.000	.4106	.5134
		Model B	.1100(*)	.01718	.000	.0586	.1614
		Model D	.0880(*)	.01718	.000	.0366	.1394
		Model E	.1940(*)	.01718	.000	.1426	.2454
	Model D	Model A	.3740(*)	.01718	.000	.3226	.4254
		Model B	.0220*	.01718	.004	-.0294	.0734
		Model C	-.0880(*)	.01718	.000	-.1394	-.0366
		Model E	.1060(*)	.01718	.000	.0546	.1574
	Model E	Model A	.2680(*)	.01718	.000	.2166	.3194
		Model B	-.0840(*)	.01718	.001	-.1354	-.0326
		Model C	-.1940(*)	.01718	.000	-.2454	-.1426
		Model D	-.1060(*)	.01718	.000	-.1574	-.0546

TABLE 22 –MULTIPLE COMPARISONS (TUKEY HSD) for GROUP II

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Second Molar M. Cusp	Model A	Model B	.1240(*)	.01362	.000	.0832	.1648
		Model C	.2780(*)	.01362	.000	.2372	.3188
		Model D	.0000*	.01362	.000	-.0408	.0408
		Model E	.3200(*)	.01362	.000	.2792	.3608
	Model B	Model A	-.1240(*)	.01362	.000	-.1648	-.0832
		Model C	.1540(*)	.01362	.000	.1132	.1948
		Model D	-.1240(*)	.01362	.000	-.1648	-.0832
		Model E	.1960(*)	.01362	.000	.1552	.2368
	Model C	Model A	-.2780(*)	.01362	.000	-.3188	-.2372
		Model B	-.1540(*)	.01362	.000	-.1948	-.1132
		Model D	-.2780(*)	.01362	.000	-.3188	-.2372
		Model E	.0420(*)	.01362	.041	.0012	.0828
	Model D	Model A	.0000*	.01362	.000	-.0408	.0408
		Model B	.1240(*)	.01362	.000	.0832	.1648
		Model C	.2780(*)	.01362	.000	.2372	.3188
		Model E	.3200(*)	.01362	.000	.2792	.3608
	Model E	Model A	-.3200(*)	.01362	.000	-.3608	-.2792
		Model B	-.1960(*)	.01362	.000	-.2368	-.1552
		Model C	-.0420(*)	.01362	.041	-.0828	-.0012
		Model D	-.3200(*)	.01362	.000	-.3608	-.2792
Second Molar D. Cusp	Model A	Model B	.3900(*)	.01567	.000	.3431	.4369
		Model C	.1500(*)	.01567	.000	.1031	.1969
		Model D	.3140(*)	.01567	.000	.2671	.3609
		Model E	1.0420(*)	.01567	.000	.9951	1.0889
	Model B	Model A	-.3900(*)	.01567	.000	-.4369	-.3431
		Model C	-.2400(*)	.01567	.000	-.2869	-.1931
		Model D	-.0760(*)	.01567	.001	-.1229	-.0291
		Model E	.6520(*)	.01567	.000	.6051	.6989
	Model C	Model A	-.1500(*)	.01567	.000	-.1969	-.1031
		Model B	.2400(*)	.01567	.000	.1931	.2869
		Model D	.1640(*)	.01567	.000	.1171	.2109
		Model E	.8920(*)	.01567	.000	.8451	.9389
	Model D	Model A	-.3140(*)	.01567	.000	-.3609	-.2671
		Model B	.0760(*)	.01567	.001	.0291	.1229
		Model C	-.1640(*)	.01567	.000	-.2109	-.1171

		Model E	.7280(*)	.01567	.000	.6811	.7749
	Model E	Model A	-1.0420(*)	.01567	.000	-1.0889	-.9951
		Model B	-.6520(*)	.01567	.000	-.6989	-.6051
		Model C	-.8920(*)	.01567	.000	-.9389	-.8451
		Model D	-.7280(*)	.01567	.000	-.7749	-.6811

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Second Molar Root	Model A	Model B	2.9540(*)	.01368	.000	2.9131	2.9949
		Model C	1.7400(*)	.01368	.000	1.6991	1.7809
		Model D	2.4960(*)	.01368	.000	2.4551	2.5369
		Model E	2.1860(*)	.01368	.000	2.1451	2.2269
	Model B	Model A	-2.9540(*)	.01368	.000	-2.9949	-2.9131
		Model C	-1.2140(*)	.01368	.000	-1.2549	-1.1731
		Model D	-.4580(*)	.01368	.000	-.4989	-.4171
		Model E	-.7680(*)	.01368	.000	-.8089	-.7271
	Model C	Model A	-1.7400(*)	.01368	.000	-1.7809	-1.6991
		Model B	1.2140(*)	.01368	.000	1.1731	1.2549
		Model D	.7560(*)	.01368	.000	.7151	.7969
		Model E	.4460(*)	.01368	.000	.4051	.4869
	Model D	Model A	-2.4960(*)	.01368	.000	-2.5369	-2.4551
		Model B	.4580(*)	.01368	.000	.4171	.4989
		Model C	-.7560(*)	.01368	.000	-.7969	-.7151
		Model E	-.3100(*)	.01368	.000	-.3509	-.2691
	Model E	Model A	-2.1860(*)	.01368	.000	-2.2269	-2.1451
		Model B	.7680(*)	.01368	.000	.7271	.8089
		Model C	-.4460(*)	.01368	.000	-.4869	-.4051
		Model D	.3100(*)	.01368	.000	.2691	.3509

TABLE 23 –MULTIPLE COMPARISONS (TUKEY HSD) for GROUP III

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
						Lower Bound	Upper Bound
Canine Cusp	Model A	Model B	.0000*	.00992	.000	-.0297	.0297
		Model C	-.0080*	.00992	.004	-.0377	.0217
		Model D	.0720(*)	.00992	.000	.0423	.1017
		Model E	-.0060*	.00992	.004	-.0357	.0237
	Model B	Model A	.0000*	.00992	.000	-.0297	.0297
		Model C	-.0080*	.00992	.004	-.0377	.0217
		Model D	.0720(*)	.00992	.000	.0423	.1017
		Model E	-.0060*	.00992	.004	-.0357	.0237
	Model C	Model A	.0080*	.00992	.004	-.0217	.0377
		Model B	.0080*	.00992	.004	-.0217	.0377
		Model D	.0800(*)	.00992	.000	.0503	.1097
		Model E	.0020*	.00992	.000	-.0277	.0317
	Model D	Model A	-.0720(*)	.00992	.000	-.1017	-.0423
		Model B	-.0720(*)	.00992	.000	-.1017	-.0423
		Model C	-.0800(*)	.00992	.000	-.1097	-.0503
		Model E	-.0780(*)	.00992	.000	-.1077	-.0483
	Model E	Model A	.0060*	.00992	.004	-.0237	.0357
		Model B	.0060*	.00992	.004	-.0237	.0357
		Model C	-.0020*	.00992	.000	-.0317	.0277
		Model D	.0780(*)	.00992	.000	.0483	.1077
Canine DC	Model A	Model B	-.3840(*)	.01099	.000	-.4169	-.3511
		Model C	1.3220(*)	.01099	.000	1.2891	1.3549
		Model D	1.6700(*)	.01099	.000	1.6371	1.7029
		Model E	1.3280(*)	.01099	.000	1.2951	1.3609
	Model B	Model A	.3840(*)	.01099	.000	.3511	.4169
		Model C	1.7060(*)	.01099	.000	1.6731	1.7389
		Model D	2.0540(*)	.01099	.000	2.0211	2.0869

		Model E	1.7120(*)	.01099	.000	1.6791	1.7449
	Model C	Model A	-1.3220(*)	.01099	.000	-1.3549	-1.2891
		Model B	-1.7060(*)	.01099	.000	-1.7389	-1.6731
		Model D	.3480(*)	.01099	.000	.3151	.3809
		Model E	.0060*	.01099	.004	-.0269	.0389
	Model D	Model A	-1.6700(*)	.01099	.000	-1.7029	-1.6371
		Model B	-2.0540(*)	.01099	.000	-2.0869	-2.0211
		Model C	-.3480(*)	.01099	.000	-.3809	-.3151
		Model E	-.3420(*)	.01099	.000	-.3749	-.3091
	Model E	Model A	-1.3280(*)	.01099	.000	-1.3609	-1.2951
		Model B	-1.7120(*)	.01099	.000	-1.7449	-1.6791
		Model C	-.0060*	.01099	.004	-.0389	.0269
		Model D	.3420(*)	.01099	.000	.3091	.3749

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
Second Pre Molar	Model A	Model B	.0160*	.00815	.000	-.0084	.0404
		Model C	.0160*	.00815	.004	-.0084	.0404
		Model D	.9400(*)	.00815	.000	.9156	.9644
		Model E	.0300(*)	.00815	.011	.0056	.0544
	Model B	Model A	-.0160*	.00815	.000	-.0404	.0084
		Model C	.0000*	.00815	.000	-.0244	.0244
		Model D	.9240(*)	.00815	.000	.8996	.9484
		Model E	.0140*	.00815	.001	-.0104	.0384
	Model C	Model A	-.0160*	.00815	.004	-.0404	.0084
		Model B	.0000*	.00815	.000	-.0244	.0244
		Model D	.9240(*)	.00815	.000	.8996	.9484
		Model E	.0140*	.00815	.006	-.0104	.0384
	Model D	Model A	-.9400(*)	.00815	.000	-.9644	-.9156
		Model B	-.9240(*)	.00815	.000	-.9484	-.8996
		Model C	-.9240(*)	.00815	.000	-.9484	-.8996
		Model E	-.9100(*)	.00815	.000	-.9344	-.8856
	Model E	Model A	-.0300(*)	.00815	.011	-.0544	-.0056
		Model B	-.0140*	.00815	.006	-.0384	.0104
		Model C	-.0140*	.00815	.006	-.0384	.0104
		Model D	.9100(*)	.00815	.000	.8856	.9344
Second Molar MC	Model A	Model B	.0060*	.00815	.004	-.0184	.0304
		Model C	1.3400(*)	.00815	.000	1.3156	1.3644
		Model D	2.2900(*)	.00815	.000	2.2656	2.3144
		Model E	1.1980(*)	.00815	.000	1.1736	1.2224
	Model B	Model A	-.0060*	.00815	.004	-.0304	.0184
		Model C	1.3340(*)	.00815	.000	1.3096	1.3584

		Model D	2.2840(*)	.00815	.000	2.2596	2.3084
		Model E	1.1920(*)	.00815	.000	1.1676	1.2164
	Model C	Model A	-1.3400(*)	.00815	.000	-1.3644	-1.3156
		Model B	-1.3340(*)	.00815	.000	-1.3584	-1.3096
		Model D	.9500(*)	.00815	.000	.9256	.9744
		Model E	-.1420(*)	.00815	.000	-.1664	-.1176
	Model D	Model A	-2.2900(*)	.00815	.000	-2.3144	-2.2656
		Model B	-2.2840(*)	.00815	.000	-2.3084	-2.2596
		Model C	-.9500(*)	.00815	.000	-.9744	-.9256
		Model E	-1.0920(*)	.00815	.000	-1.1164	-1.0676
	Model E	Model A	-1.1980(*)	.00815	.000	-1.2224	-1.1736
		Model B	-1.1920(*)	.00815	.000	-1.2164	-1.1676
		Model C	.1420(*)	.00815	.000	.1176	.1664
		Model D	1.0920(*)	.00815	.000	1.0676	1.1164

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
Second Molar M. Cusp	Model A	Model B	.0180*	.00825	.226	-.0067	.0427
		Model C	.0220*	.00825	.004	-.0027	.0467
		Model D	.9080(*)	.00825	.000	.8833	.9327
		Model E	.9540(*)	.00825	.000	.9293	.9787
	Model B	Model A	-.0180*	.00825	.006	-.0427	.0067
		Model C	.0040*	.00825	.000	-.0207	.0287
		Model D	.8900(*)	.00825	.000	.8653	.9147
		Model E	.9360(*)	.00825	.000	.9113	.9607
	Model C	Model A	-.0220*	.00825	.000	-.0467	.0027
		Model B	-.0040*	.00825	.001	-.0287	.0207
		Model D	.8860(*)	.00825	.000	.8613	.9107
		Model E	.9320(*)	.00825	.000	.9073	.9567
	Model D	Model A	-.9080(*)	.00825	.000	-.9327	-.8833
		Model B	-.8900(*)	.00825	.000	-.9147	-.8653
		Model C	-.8860(*)	.00825	.000	-.9107	-.8613
		Model E	.0460(*)	.00825	.000	.0213	.0707
	Model E	Model A	-.9540(*)	.00825	.000	-.9787	-.9293
		Model B	-.9360(*)	.00825	.000	-.9607	-.9113
		Model C	-.9320(*)	.00825	.000	-.9567	-.9073
		Model D	-.0460(*)	.00825	.000	-.0707	-.0213
Second Molar D. Cusp	Model A	Model B	-.0080*	.00693	.006	-.0287	.0127
		Model C	.1240(*)	.00693	.000	.1033	.1447
		Model D	1.7460(*)	.00693	.000	1.7253	1.7667
		Model E	.9580(*)	.00693	.000	.9373	.9787
	Model B	Model A	.0080*	.00693	.006	-.0127	.0287
		Model C	.1320(*)	.00693	.000	.1113	.1527
		Model D	1.7540(*)	.00693	.000	1.7333	1.7747

		Model E	.9660(*)	.00693	.000	.9453	.9867
	Model C	Model A	-.1240(*)	.00693	.000	-.1447	-.1033
		Model B	-.1320(*)	.00693	.000	-.1527	-.1113
		Model D	1.6220(*)	.00693	.000	1.6013	1.6427
		Model E	.8340(*)	.00693	.000	.8133	.8547
	Model D	Model A	-1.7460(*)	.00693	.000	-1.7667	-1.7253
		Model B	-1.7540(*)	.00693	.000	-1.7747	-1.7333
		Model C	-1.6220(*)	.00693	.000	-1.6427	-1.6013
		Model E	-.7880(*)	.00693	.000	-.8087	-.7673
	Model E	Model A	-.9580(*)	.00693	.000	-.9787	-.9373
		Model B	-.9660(*)	.00693	.000	-.9867	-.9453
		Model C	-.8340(*)	.00693	.000	-.8547	-.8133
		Model D	.7880(*)	.00693	.000	.7673	.8087

Dependent Variable	(I) Model	(J) Model	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
Canine Root	Model A	Model B	-.5000(*)	.01173	.000	-.5351	-.4649
		Model C	-.4460(*)	.01173	.000	-.4811	-.4109
		Model D	.0040*	.01173	.001	-.0311	.0391
		Model E	-.0160*	.01173	.001	-.0511	.0191
	Model B	Model A	.5000(*)	.01173	.000	.4649	.5351
		Model C	.0540(*)	.01173	.001	.0189	.0891
		Model D	.5040(*)	.01173	.000	.4689	.5391
		Model E	.4840(*)	.01173	.000	.4489	.5191
	Model C	Model A	.4460(*)	.01173	.000	.4109	.4811
		Model B	-.0540(*)	.01173	.001	-.0891	-.0189
		Model D	.4500(*)	.01173	.000	.4149	.4851
		Model E	.4300(*)	.01173	.000	.3949	.4651
	Model D	Model A	-.0040*	.01173	.004	-.0391	.0311
		Model B	-.5040(*)	.01173	.000	-.5391	-.4689
		Model C	-.4500(*)	.01173	.000	-.4851	-.4149
		Model E	-.0200*	.01173	.004	-.0551	.0151
	Model E	Model A	.0160*	.01173	.004	-.0191	.0511
		Model B	-.4840(*)	.01173	.000	-.5191	-.4489
		Model C	-.4300(*)	.01173	.000	-.4651	-.3949
		Model D	.0200*	.01173	.001	-.0151	.0551
Second Molar Root	Model A	Model B	-.0120*	.00844	.001	-.0372	.0132
		Model C	-.4220(*)	.00844	.000	-.4472	-.3968
		Model D	.0440(*)	.00844	.000	.0188	.0692
		Model E	.0480(*)	.00844	.000	.0228	.0732
	Model B	Model A	.0120*	.00844	.001	-.0132	.0372
		Model C	-.4100(*)	.00844	.000	-.4352	-.3848

		Model D	.0560(*)	.00844	.000	.0308	.0812
		Model E	.0600(*)	.00844	.000	.0348	.0852
	Model C	Model A	.4220(*)	.00844	.000	.3968	.4472
		Model B	.4100(*)	.00844	.000	.3848	.4352
		Model D	.4660(*)	.00844	.000	.4408	.4912
		Model E	.4700(*)	.00844	.000	.4448	.4952
	Model D	Model A	-.0440(*)	.00844	.000	-.0692	-.0188
		Model B	-.0560(*)	.00844	.000	-.0812	-.0308
		Model C	-.4660(*)	.00844	.000	-.4912	-.4408
		Model E	.0040*	.00844	.004	-.0212	.0292
	Model E	Model A	-.0480(*)	.00844	.000	-.0732	-.0228
		Model B	-.0600(*)	.00844	.000	-.0852	-.0348
		Model C	-.4700(*)	.00844	.000	-.4952	-.4448
		Model D	-.0040*	.00844	.004	-.0292	.0212

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