ORIGINAL ARTICLE

Effect of Different Occlusal Loads on Periodontium: A Three-dimensional Finite Element Analysis

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ABSTRACT

Introduction: The periodontal tissue reaction to variations in occlusal forces has been described in the literature where in clinical and histologic changes are discussed that produced due to stresses in the periodontal structures. Unfortunately, these stresses are not quantified.

Aim: The aim of this study is to determine the stress produced on various periodontal tissues at different occlusal loads using finite element model (FEM) study.

Materials and methods: Four FEMs of maxillary incisor were designed consisting of the tooth, pulp, periodontal ligament (PDL), and alveolar bone at the various levels of bone height (25, 50, and 75%). Different occlusal loads (5, 15, 24, and 29 kg) at an angle of 50° to the long axis of the tooth were applied on the palatal surface at the level of middle third of the crown. All the models were assumed to be isotropic, linear, and elastic, and the analysis was performed on a Pentium IV processor computer using ANSYS software.

Results: At normofunction load, the stresses were maximum on the mesial side near the cervical region at point D for tooth (−10.93 MPa), for PDL (−4.06 MPa), for bone (−4.3 MPa); with normal bone levels, as the bone levels decreased the stresses increased and the stresses tend to concentrate at the apical region. At any given point, the stresses were increased by 60 and 90% at hyperfunctional loads of 24 and 29 kg respectively, and with hypofunctional load of 5 kg, stresses were reduced by 60%.

Conclusion: Based on the findings of the present study, there is reasonably good attempt to express numerical data of stress to be given normal occlusal and hyperfunctional loads to simulate clinical occlusal situations which are known to be responsible for healthy and diseased periodontium.

Keywords: Finite element analysis, Hyperfunctional load, Hypofunctional load, Stress, Trauma from occlusion.

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INTRODUCTION

Since the beginning of the 20th century, the role of occlusal trauma as the causation of periodontal disease has been studied. The role of occlusion on periodontal health is challenging and the results of research studies are contradictory and inconclusive. Several studies have tried to assess the stress produced by the occlusal forces within the tooth and supporting structures. Finite element analysis is a numerical form of computer analysis using mechanical engineering that allows the stress to be identified and quantified within the structures constructed using elements and nodes.

Few studies on FEM analysis of stress have focused on primary and secondary trauma from occlusion including a two-dimensional (2D) model of postreinforced maxillary central incisor which evaluated the principal stresses in PDL at various base levels.[1,2] Stones[3] utilized a three-dimensional (3D) FEM model (ANSYS 5.4) of natural central incisor to calculate maximum and minimum principal stresses in the PDL of maxillary central incisor with different bone heights. In both the studies, only the PDL stresses were calculated; however, alveolar bone and root stresses are important in alveolar bone destruction due to occlusal trauma either alone or during periodontitis. A study done by Reddy and Vandana[4] was the first report of distribution of stresses in PDL tissues (PDL, root, and alveolar bone) using a 3D FEM.[4]

The adaptive capacity of the periodontal structure (with regard to occlusal forces) varies greatly from individual to individual and even in the same individual from time to time. These occlusal forces may also vary based on somatic and psychic changes in the individual. The periodontal tissue reaction to variations in occlusal forces has been described in the literature wherein clinical and histologic changes are discussed that are produced due to stresses in the periodontal structures. Unfortunately, these stresses are not quantified.

Medline/PubMed search using keywords traumatic occlusion, stress, quantity, alveolar bone, PDL, root, FEM was scanty.

In the present study, first attempt was been made to apply normofunctional (150 N), hypofunctional (50 N), and hyperfunctional (240 and 290 N) occlusal loads on maxillary central incisors with normal alveolar height

and with compromised alveolar height with 25, 50, and 75% bone loss to quantify various stress levels in tooth root, PDL, and alveolar bone.

MATERIALS AND METHODS

The finite element analysis was performed on a personal computer (Pentium III, Intel) using ANSYS software, marketed by ANSYS Inc., USA. In this study, a 3D FEM of the anatomic size and shape of an average Indian maxillary central incisor was constructed. Variable PDL widths were developed at different occlusogingival levels derived from the data of Coolidge.[5] The use of these varying thicknesses makes the model more precise and realistic.

In this study, the analytical model was built by scanning the pictures of the maxillary central incisor in Wheeler's textbook.[6] The geometric model was converted into FEM. The type of element used for modeling was a 3D quadratic tetrahedral element with three degrees of freedom (dof) for each node. The FEM was the representation of geometry in terms of a finite number of elements and nodes, the complete model consisted of 47,229 elements and 68,337 nodes, and Table 1 represents the number of elements and nodes for varying bone levels. The different structures, such as alveolar bone, tooth, and PDL were assigned the material characteristics confirming to the data available in the literature.[7,8] In this study, all the tissues were assumed to be isotropic, homogeneous, and linear (Table 2). The boundary condition of the FEM basically represents the load imposed on the structures under the study and their fixation counterparts. The model was restrained at the base in order to avoid any motion against the loads imposed on the dentoalveolar structures.

The compressive stresses induced within the root, PDL, and alveolar bone due to various loads, namely normofunctional (150 N), hypofunctional (50 N), and hyperfunctional (240 and 290 N) were studied. These loads were applied on the tooth, at varying heights of the alveolar bone levels, i.e., with normal alveolar height and with compromised alveolar height with 25, 50, and 75% bone loss, in a palatolabial direction on palatal surface of crown at an angle of 50° to the long axis of the tooth at the level of the middle third of the crown. The load represents the average angle of contact between maxillary and mandibular incisor teeth in Angle's class I centric occlusion.

The criteria used to evaluate the structure under loading were to judge the compressive stress at certain nodes, and minimum principal stress were used. Five different points positioned on the included tissues D, E, F, G, and H (Fig. 1) were selected for the purpose of stress analysis. The stresses and tooth displacement (Fig. 2) were analyzed using color coding deformation and graphical animations. The tables comprised of numerical values of above information.

RESULTS

The criteria used to evaluate a structure from the stress perspective are the minimal principal stress criteria. The results are summarized in Tables 3 and 4.

STRESSES ON TOOTH

Normal Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point D_t (−1.18 MPa), which is located at the

Table 1: Number of elements and nodes used in different models of the tooth

% bone level reduced	Number of elements	Number of nodes
0	47,229	68,337
25	42,998	62,766
50	35,639	52,740
75	28,147	42,429

Table 2: Material parameters used in finite element model

Material	Young's modulus (kg/mm^2)	Poisson's ratio
Tooth	2.0×10^3	0.30
PDL	6.8×10^{-2}	0.49
Alveolar bone	1.4×10^3	0.30
Pulp	0.2	0.45

Figs 1A to C: (A) Sampling points on tooth; (B) sampling points on PDL; and (C) sampling points on bone
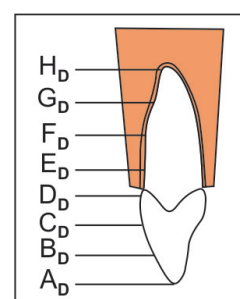

Fig. 2: Sampling points for displacement

CEJ on the mesial side, and maximum stresses are seen at point D_t (−10.93 MPa) on the labial site (Fig. 3 and Table 3).

At a decreased load of 50 N (hypofunction), the minimum (−0.39 MPa) and maximum (−3.64 MPa) stresses were seen at points similar to that seen with normal load.

At an increased load of 240 N (hyperfunction), the minimum (−1.88 MPa) and maximum (−17.49 MPa) stresses were seen at points similar to that seen with normal load.

At another increased load of 290 N (hyperfunction), the minimum (−2.28 MPa) and maximum (−21.13 MPa) stresses were seen at points similar to that seen with normal load.

Compromised Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point E_t (−0.02 MPa) which is located at the cervical third of root on the palatal side, and maximum stresses are seen at point F_t (−140.45 MPa), a point at the junction of middle and apical third on the tooth on the labial site with 75% bone loss (Table 4).

At a load of 50 N (hypofunction), the minimum stresses were seen at point E_t (−0.009 MPa) which is located at the cervical third of root on the palatal side, and maximum stresses are seen at point F_t (−46.81 MPa), a point at midpoint on the root on the labial site with 75% bone loss.

At an increased load of 240 N (hyperfunction), the minimum (−0.04 MPa) and maximum (−224.71 MPa) stresses were seen at points similar to that seen with normal load with similar compromised bone heights.

At another increased load of 290 N (hyperfunction), the minimum (−0.05 MPa) and maximum (−271.53 MPa) stresses were seen at points similar to that seen with normal load with similar compromised bone heights.

STRESSES ON PDL

Normal Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point D_p (−0.005 MPa) which is located at CEJ on the mesial side, and maximum stresses are seen at point H_p (−4.06 MPa) on the palatal side (Fig. 4 and Table 3).

At a decreased load of 50 N (hypofunction), the minimum (−0.001 MPa) and maximum (−1.35 MPa) stresses were seen at points similar to that seen with normal load.

At an increased load of 240 N (hyperfunction), the minimum (−0.008 MPa) and maximum (−6.5 MPa) stresses were seen at points similar to that seen with normal load.

At another increased load of 290 N (hyperfunction), the minimum (−0.01 MPa) and maximum (−7.8 MPa) stresses were seen at points similar to that seen with normal load.

Table 3: Stress values in tooth, PDL, and alveolar bone height at normal bone height at different occlusal loads

	Stress on tooth (normal bone height)						Stress on PDL (normal bone height)						Stress on bone (normal bone height)					
	Minimum			Maximum			Minimum			Maximum			Minimum			Maximum		
	Site	Point	Value (MPa)	Site	Point	Value (MPa)	Site	Point	Value (MPa)	Site	Point	Value (MPa)	Site	Point	Value (MPa)	Site	Point	Value (MPa)
15 kg	Mesial	D(t)	−1.18	Labial	D(t)	−10.93	Mesial	D(p)	−0.005	Distal	H(p)	−4.06	Distal	D(b)	−0.04	Labial	D(b)	−4.3
5 kg	Mesial	D(t)	−0.39	Labial	D(t)	−3.64	Mesial	D(p)	−0.001	Distal	H(p)	−1.35	Distal	D(b)	−0.01	Labial	D(b)	−1.44
24 kg	Mesial	D(t)	−1.88	Labial	D(t)	−17.48	Mesial	D(p)	−0.008	Distal	H(p)	−6.5	Distal	D(b)	−0.06	Labial	D(b)	−6.95
29 kg	Mesial	D(t)	−2.28	Labial	D(t)	−21.33	Mesial	D(p)	−0.01	Distal	H(p)	−7.8	Distal	D(b)	−0.08	Labial	D(b)	−8.4

Table 4: Stress values in tooth, PDL, alveolar bone with compromised bone heights for different occlusal loads

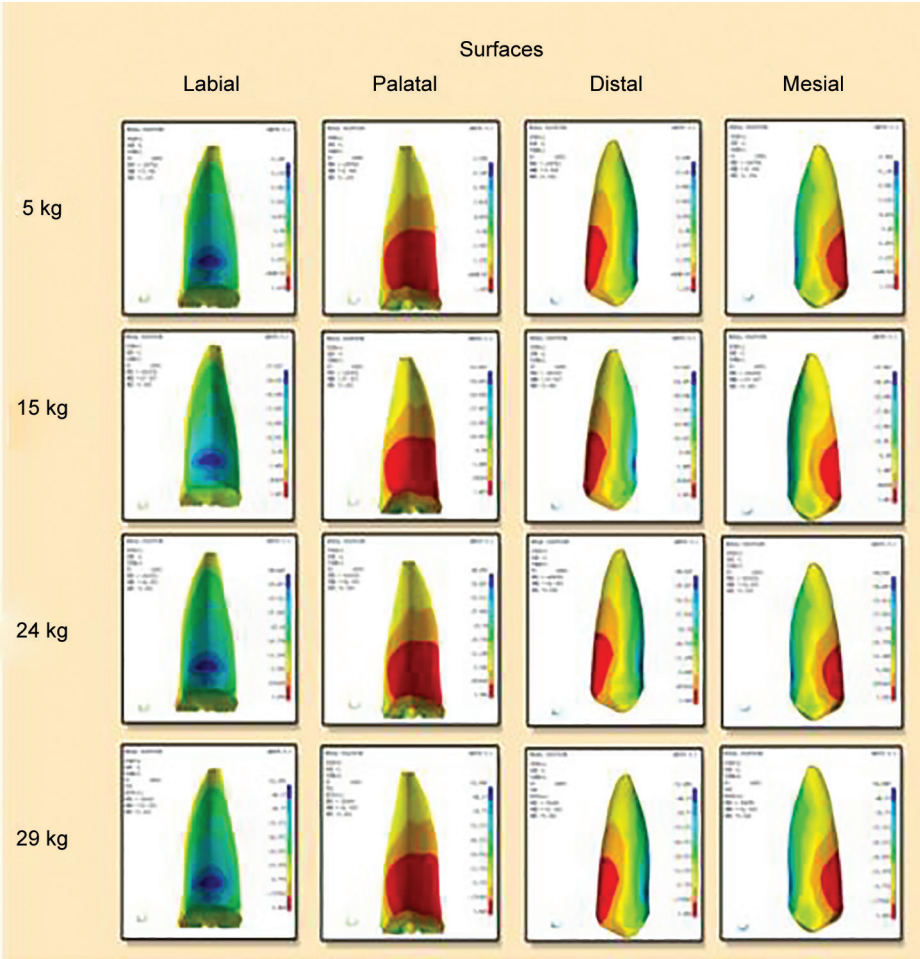|	Stress on tooth (compromised bone height)								Stress on PDL (compromised bone height)								Stress on bone (compromised bone height)							
	Minimum				Maximum				Minimum				Maximum				Minimum				Maximum			
	Site	Point	Value (MPa)	Bone level	Site	Point	Value (MPa)	Bone level	Site	Point	Value (MPa)	Bone level	Site	Point	Value (MPa)	Bone level	Site	Point	Value (MPa)	Bone level	Site	Point	Value (MPa)	Bone level
15 kg	Palatal	E(t)	−0.02	75%	Labial	F(t)	−140.45	75%	Distal	F(p)	−0.02	50%	Palatal	G(p)	−73.46	75%	Distal	F(b)	−0.06	50%	Palatal	G(b)	−44.01	75%
5 kg	Palatal	E(t)	−0.009	75%	Labial	F(t)	−46.81	75%	Distal	F(p)	−0.008	50%	Palatal	G(p)	−24.48	75%	Distal	F(b)	−0.02	50%	Palatal	G(b)	−14.68	75%
24 kg	Palatal	E(t)	−0.04	75%	Labial	F(t)	−224.7	75%	Distal	F(p)	−0.03	50%	Palatal	G(p)	−117.5	75%	Distal	F(b)	−0.1	50%	Palatal	G(b)	−70.46	75%
29 kg	Palatal	E(t)	−0.05	75%	Labial	F(t)	−271.53	75%	Distal	F(p)	−0.04	50%	Palatal	G(p)	−142.0	75%	Distal	F(b)	−0.12	50%	Palatal	G(b)	−85.14	75%

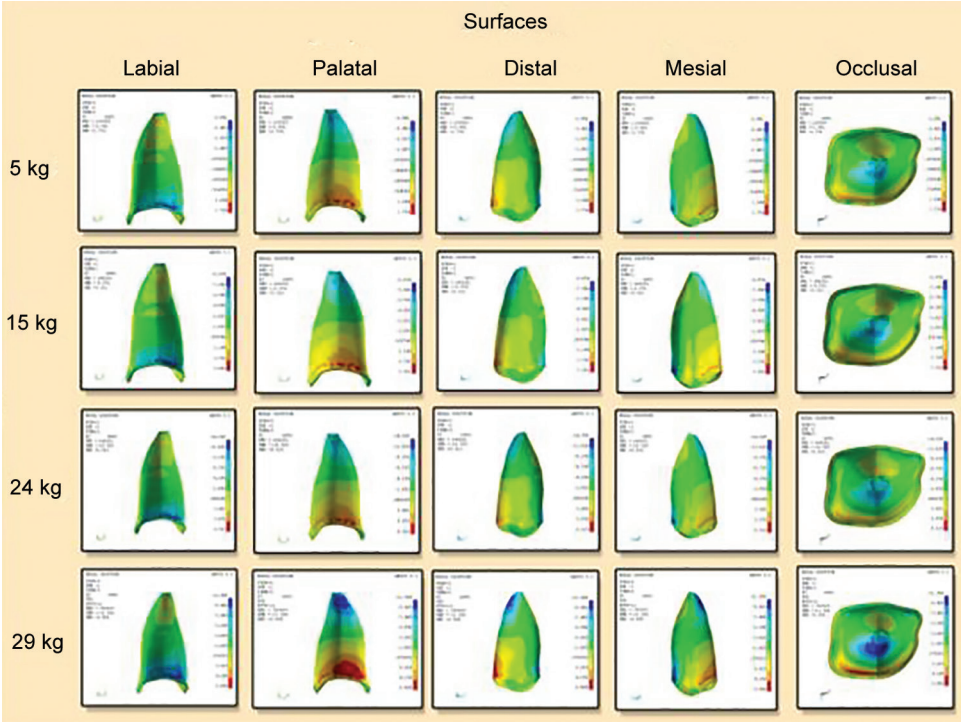**Fig. 3:** Tooth root normal bone height

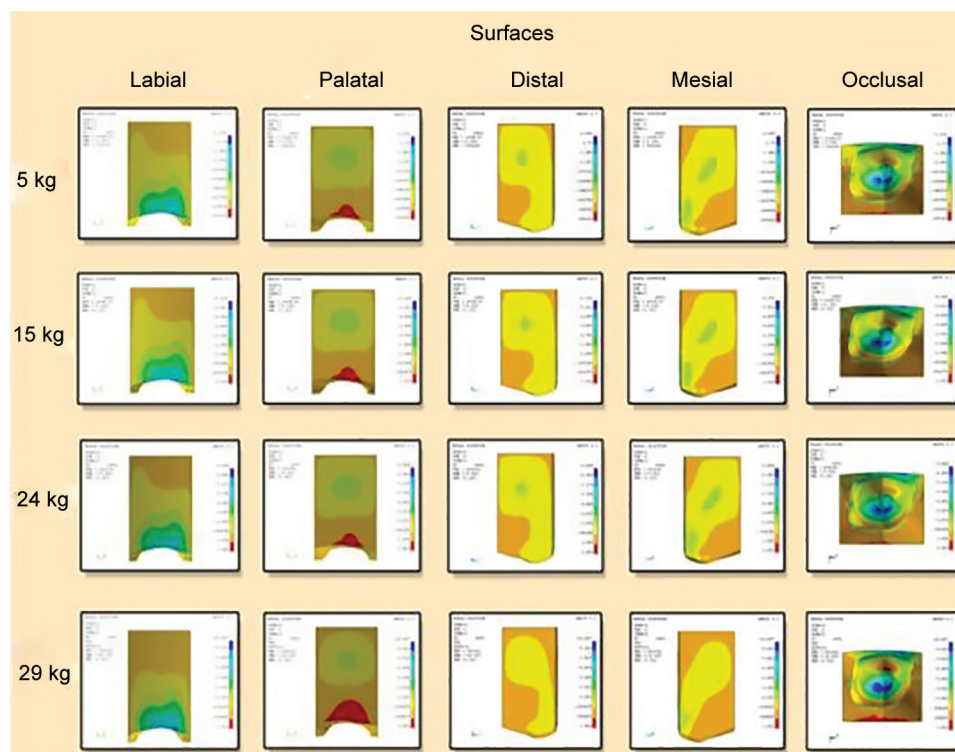**Fig. 4:** Periodontal ligament normal bone height

Fig. 5: Normal bone height

Compromised Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point F_p (−0.02 MPa) which is located at midpoint of PDL on distal side with 50% bone loss, and maximum stresses are seen at point G_p (−73.46 MPa), which is located at the junction of middle apical third of PDL site with 75% bone loss on palatal side (Table 4).

At a decreased load of 50 N (hypofunction), the minimum (−7.8 MPa) and maximum (−24.48 MPa) stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

At an increased load of 240 N (hyperfunction), the minimum (−0.03 MPa) and maximum (−117.54 MPa) stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

At another increased load of 290 N (hyperfunction), the minimum (−0.04 MPa) and maximum (−142.03 MPa) stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

STRESSES ON ALVEOLAR BONE

With Normal Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point D_b (−0.04 MPa) which is located at the crest on the mesial side, and maximum stresses are seen at point D_b (−4.3 MPa) on the labial side (Fig. 5 and Table 3).

At a decreased load of 50 N (hypofunction), the minimum (−0.01 MPa) and maximum (−1.44 MPa)

stresses were seen at points similar to that seen with normal load.

At an increased load of 240 N (hyperfunction), the minimum (−0.06 MPa) and maximum (−6.9 MPa) stresses were seen at points similar to that seen with normal load.

At another increased load of 290 N (hyperfunction), the minimum (−0.08 MPa) and maximum (−8.4 MPa) stresses were seen at points similar to that seen with normal load.

Compromised Bone Height

At a load of 150 N (normofunction), the minimum stresses were seen at point F_b (−0.06 MPa) which is located at the midpoint of the alveolar bone on the distal side with 50% bone loss, and maximum stresses are seen at point G_b (−44.01 MPa), a point at the junction of middle and apical third of the bone on the palatal side with 75% bone loss (Table 4).

At a decreased load of 50 N (hypofunction), the minimum (−0.02 MPa) and maximum (−14.68 MPa) stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

At an increased load of 240 N (hyperfunction), the minimum (−0.1 MPa) and maximum (−70.46 MPa) stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

At another increased load of 290 N (hyperfunction), the minimum (−0.12 MPa) and maximum (−85.14 MPa)

stresses were seen at points similar to that seen with normal load and with similar compromised bone heights.

With decreased load (hypofunction), the stresses were reduced by 66%, and with increase in load to 24 and 29 kg (hyperfunction), the stresses were increased by 60 and 90% respectively, at any given point and at any given bone level.

DISPLACEMENT OF THE TOOTH

The displacement of the tooth due to four different loads applied in the palatal direction, 5, 15, 24 and 29 kg, was measured at eight sampling points: Incisal edge (A_D), junction of the incisal and the middle thirds (B_D), junction of the middle and the cervical thirds (C_D), the cervical third (D_D) of the crown, and the cervical third (E_D), junction of the cervical and the middle thirds (F_D), junction of the middle and the apical thirds (G_D), and the apex (H_D) of the root.

Normal Alveolar Bone Height

Maximum displacement was noted at the incisal edge on the distal surface (A_D), and the minimum displacement was noted at the cervical third of the root (E_D) on the palatal surface for all the loads applied (Graph 1 and Fig. 6).

Maximum displacement on applying 5 kg of load was 0.15106 mm at the incisal edge of the distal surface, and the minimum displacement corresponding to zero was noted at the cervical third of the root. The amount of displacement increased to 0.45319 mm for the maximum and 0.12194 mm for the minimum displacement when

the load was increased to 15 kg. On further increasing the load to 24 and 29 kg, the maximum displacement was 0.7251 and 0.87616 mm respectively, and the minimum displacement was 0.1951 and 0.23575 mm respectively.

25% Loss of Alveolar Bone Height

Maximum displacement was noted at the incisal edge (A_D) on the distal surface, and the minimum displacement was noted at the junction of the cervical and the middle thirds of the root (F_D) on the palatal surface for all the loads applied in the study (Graph 2).

On applying 5 kg of load, maximum displacement measured 0.33895 mm and the minimum amount corresponded to zero. However, the amount of displacement increased to 1.0168 mm being the maximum and the minimum being 0.13522 mm, on applying 15 kg of load.

The amount of displacement further increased to 1.627 and 1.9659 mm on increasing the loads to 24 and 29 kg respectively, with the minimum displacements being 0.21636 and 0.26143 mm for the two loads respectively.

50% Loss of Alveolar Bone Height

Maximum displacement was noted at the incisal edge (A_D) on the distal surface, and the minimum displacement was noted at the junction of the cervical and the middle third of the root (F_D) on the palatal surface (Graph 3).

On applying 5 kg of load, maximum displacement measured was 1.9857 mm with the minimum being 0.15673 which increased to 5.9572 and 0.47019 mm on
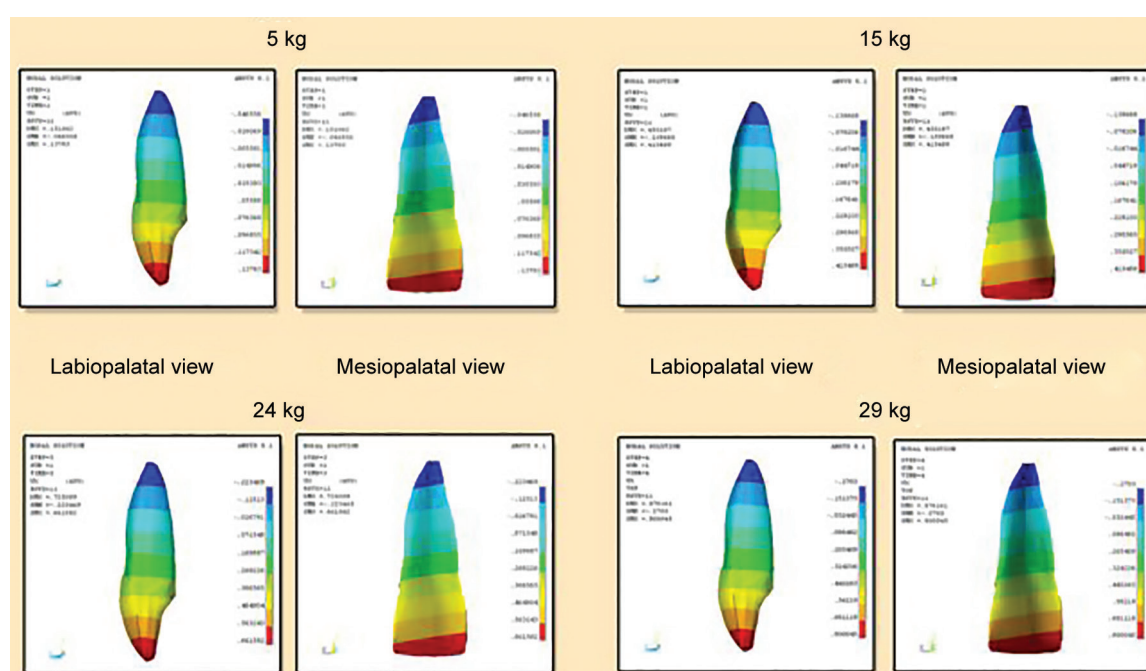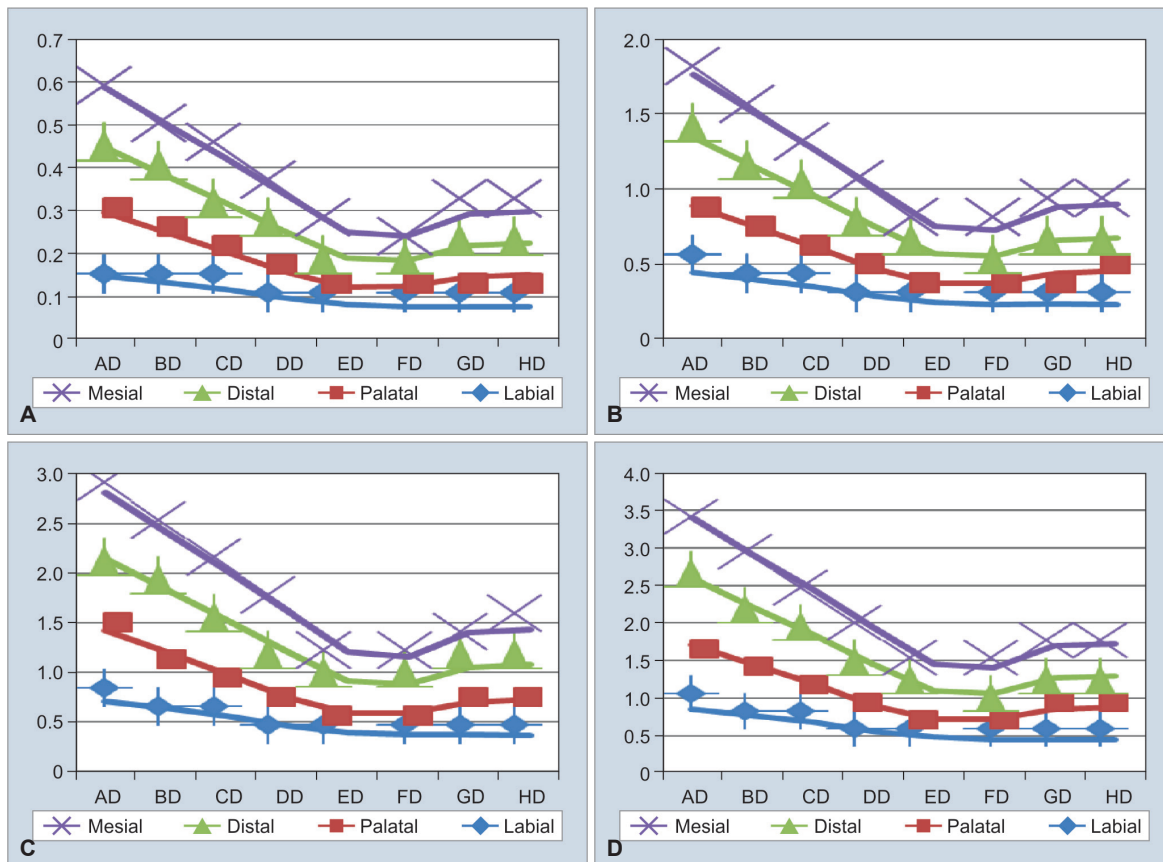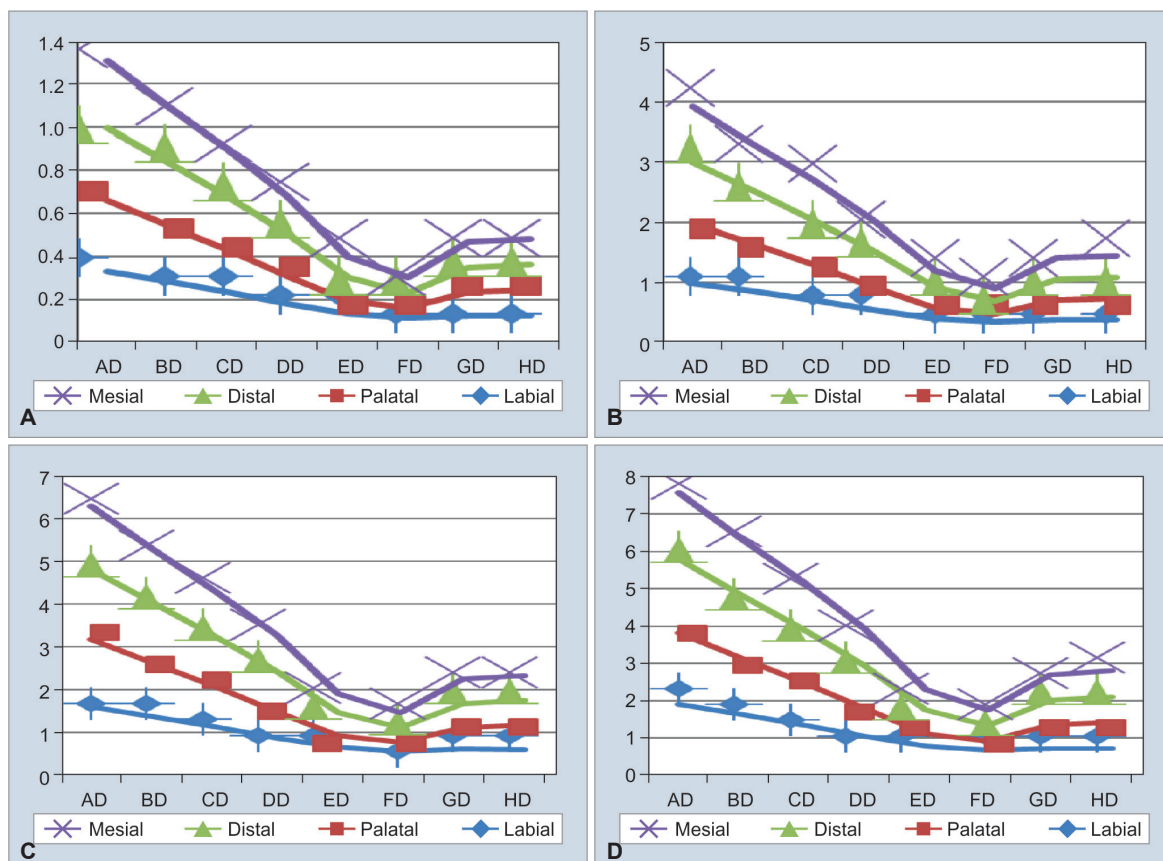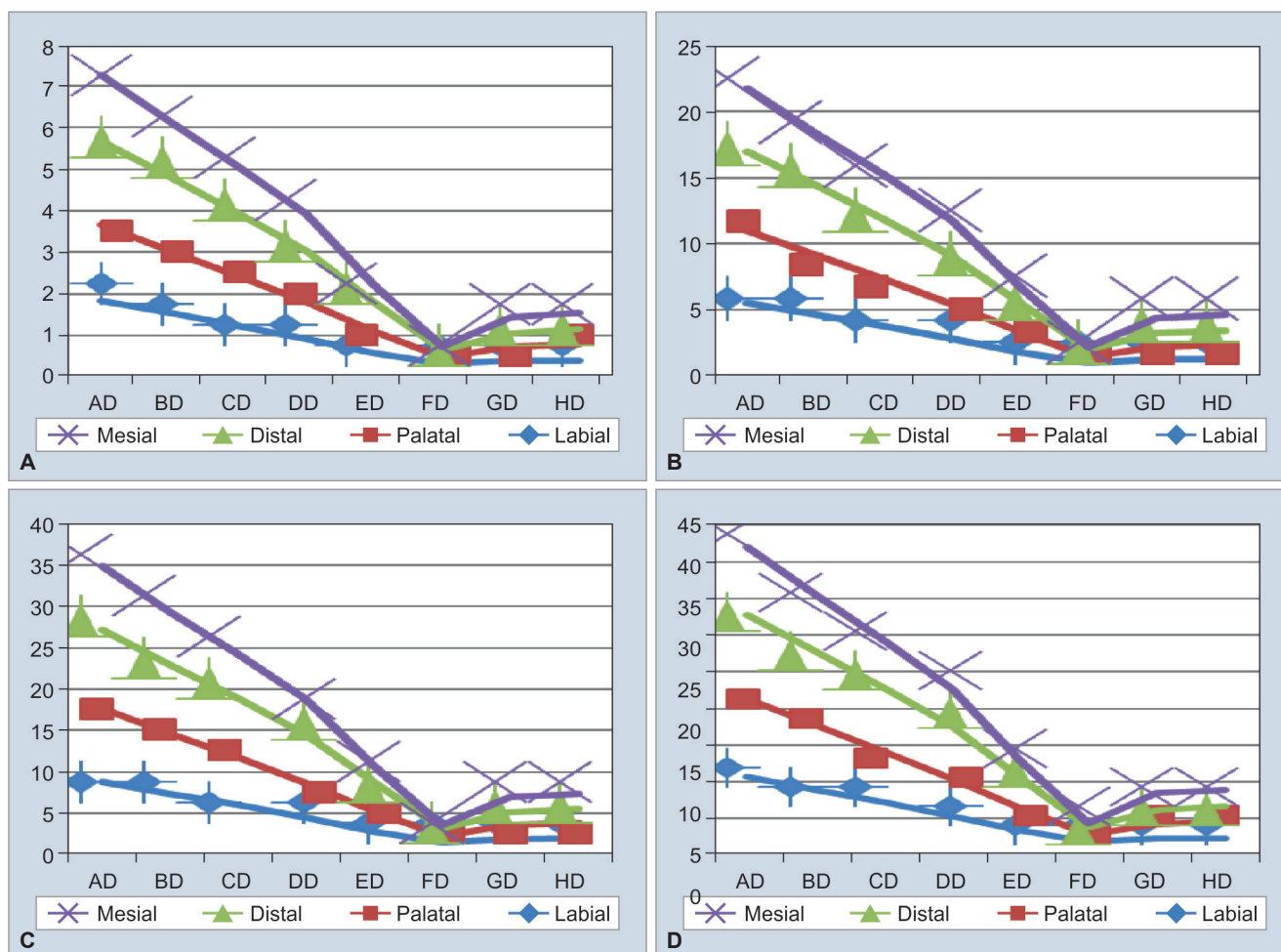

Fig. 6: Displacement normal bone height

Graphs 1A to D: Displacement with normal bone height: (A) 5 kg; (B) 15 kg; (C) 24 kg; and (D) 29 kg

Graphs 2A to D: Displacement with 25% bone loss: (A) 5 kg; (B) 15 kg; (C) 24 kg; and (D) 29 kg

Graphs 3A to D: Displacement with 50% bone loss: (A) 5 kg; (B) 15 kg; (C) 24 kg; and (D) 29 kg

increasing the load to 15 kg. On further increasing the loads to 24 and 29 kg, maximum displacements noted were 9.5315 and 11.517 mm and minimum displacements were 0.7523 and 0.90903 mm respectively.

75% Loss of Alveolar Bone Height

Maximum displacement of the tooth was noted at the incisal edge (A_D) on the distal surface, while the minimum displacement was noted at the junction of the middle and the apical thirds of the root (G_D) on the distal surface (Graph 4).

On applying 5 kg of load, maximum displacement was noted to be 42.258 mm and the minimum displacement to be 2.8274, which increased to 126.77 and 8.4821 mm when the load was increased to 15 kg. When the loads were further increased to 24 and 29 kg, the maximum displacement increased to 202.84 and 245.1 respectively, with the minimum amount of displacement being 13.571 and 16.399 respectively.

DISCUSSION

It has been said that when teeth, jaws, muscles of mastication, and temporomandibular joint (components of

somatognathic system) are in harmonious relationship, this balance will contribute to the health of periodontium. Conversely, it has also been said that when the interrelationship is disturbed, periodontal disease may follow.[9]

More has been written, but perhaps less known, about the precise role of occlusion in periodontal disease than about most other aspects of periodontology. The therapist is confronted with contradictory theories and methods. Therapy must depend on an understanding of the mechanisms of occlusion in both physiologic and pathologic conditions. The development of such an understanding is based on histologic observation of the clinical material, clinical studies, and animal experimentation. From such a basic understanding, one can more meaningfully employ the various methods of dealing with periodontal disease involving occlusion, such as trauma from occlusion. Periodontal trauma is a morbid condition produced by repeated mechanical forces exerted on the periodontium exceeding the physiologic limits of the tissue tolerance and contributing to a breakdown of the supporting tissues of the tooth.[9]

In this study, maxillary central incisor was modeled and analyzed using finite element analysis. The average force of 15 kg (150 N) was applied at the middle third
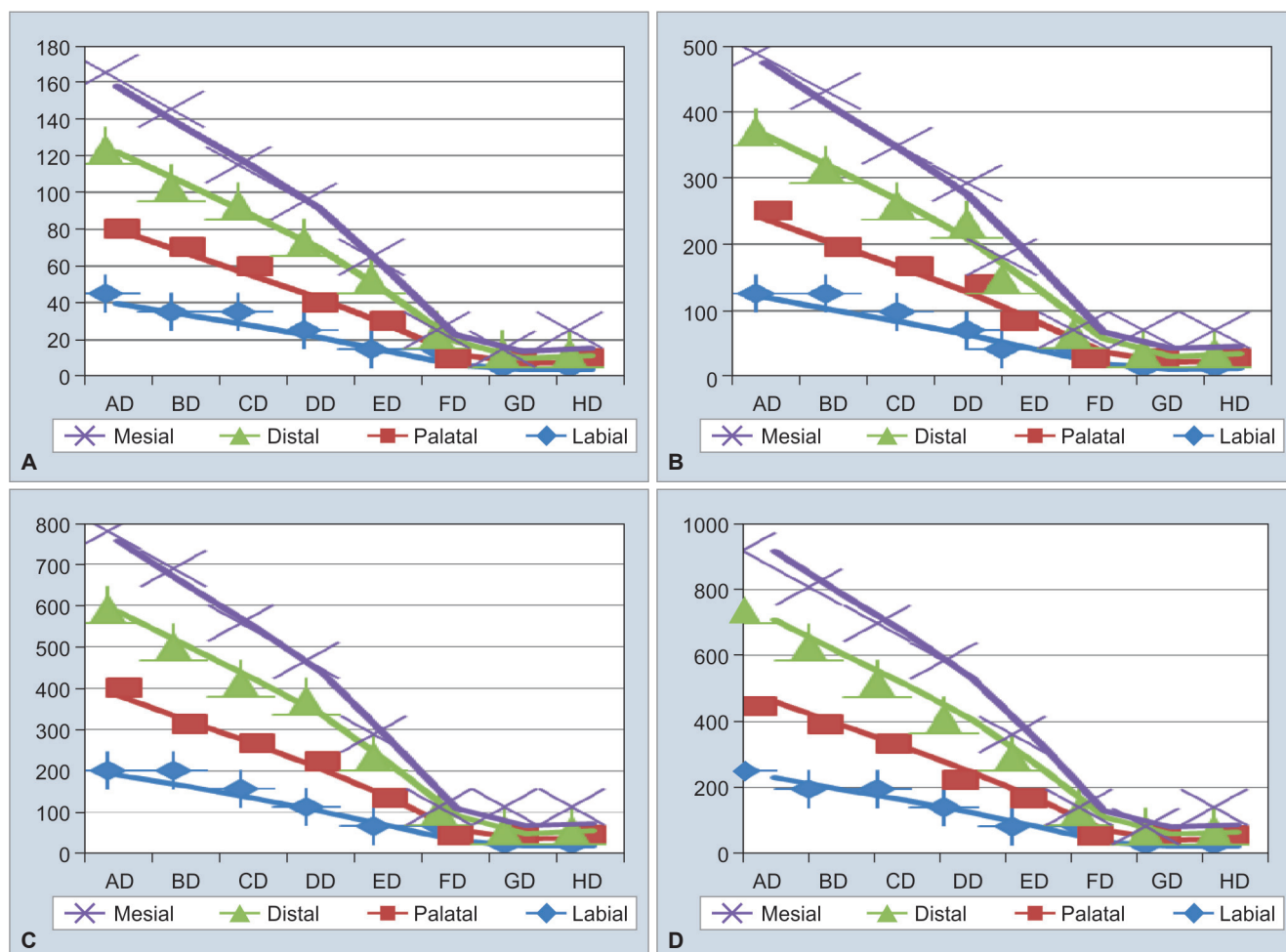
Graphs 4A to D: Displacement with 75% bone loss: (A) 5 kg; (B) 15 kg; (C) 24 kg; and (D) 29 kg

of the crown on the palatal surface at an angle of 50° in a palatolabial direction. This represented the normal occlusion. This normal occlusion was compared with three other loadings at the same direction, 5 kg (50 N) representing the hypofunction, as it is very minimal to the average force on the tooth, and 24 kg (240 N) and 29 kg (290 N) representing hyperfunction, as these loads are excessive. The average force on the bicuspids, cuspids, and incisors is about 300 N (30 kg), 200 N (20 kg), 150 N (15 kg) respectively. These occlusal forces produce a constant stress on the tooth and its supporting structures, so the measurement of the stress produced is mandatory.[10] Since the compressive stresses appear to play the vital role in the initiation of bone resorption and the lesion of the occlusal trauma, these values were the subject of important evaluation in this study.

The results of the following study are discussed as follows:

The compressive stresses induced in the root, PDL, and alveolar bone, by the occlusal load representing the hypofunction (50 N) were 66% or three times lesser than the stresses induced by a normal occlusal load of 150 N. On the contrary, when the occlusal loads increased to 240 and 290 N, the stresses increased in the periodontal structures by 60% or 1.6 times and 93% or 1.9 times respectively, representing primary trauma from occlusion.

Reduction of alveolar bone height representing the weakened periodontal support or more appropriately secondary trauma from occlusion had little effect on the tooth and the supporting tissue when 25% of the bone support was lost; however, the stresses increased dramatically when 50 and 75% of bone support was lost and also shifted apically on the tooth coinciding with the alveolar crest for the amount of bone loss.

Thus, the increase in stresses by 60 and 90% at hyperfunctional loads with normal bone heights explains the reason for primary trauma from occlusion and with compromised bone levels explain the reason for secondary trauma from occlusion.

In compromised bone height levels, the stresses increased as we reached the apex of the tooth, this can be explained as the amount of force distributed in relation to surface area is increased as the loss of the tissues lead to decrease in surface area.

Reinhardt et al[11] studied only the principal PDL stresses in primary and secondary occlusal trauma in a 2D

model of postreinforced maxillary central incisor using finite element analysis. The results showed maximum compressive stresses near the alveolar crest (−0.415 MPa), which were noted to increase dramatically as the bone levels diminished and to a lesser extent in the apical one half (−0.163) of the root for all the loads applied in the study at different bone levels. The study only used 1 N (0.1 kg) of force applied, at an angle of 50° to the long axis of the tooth on the palatal surface in palatolabial direction, at the level of middle third of the crown and at different bone levels.

Reddy and Vandana[4] studied the von mises stresses in a natural model of the maxillary central incisor tooth, PDL, and alveolar bone using a higher load of 24 kg, at an angle of 50° to the long axis of the tooth on the palatal surface in a palatolabial direction, at the level of the middle third of the crown and at different bone levels, using a 3D FEM. The results showed maximum stresses at the cervical region (21.676 MPa) and to a greater extent at the apex of the root (14.061 MPa). Their study used von Mises stresses which are used for ductile materials in which the stresses are normalized in all areas, and compression and tension cannot be interpreted adequately. Since tooth is brittle material, von mises stresses are not ideal to study compression and tension on a tooth.

Therefore, the present study used minimum principal stresses to measure the stresses as it best represents the compression state of the stress.

Geramy and Faghini studied the compression stresses in the labial site of the PDL in 3D FEM model of maxillary central incisor with normal to reducing alveolar bone heights. The highest stress levels were traced in the sub-cervical area, except for model of 8 mm of alveolar bone loss. An increase of compressive stress up to 17.13 times on the cervical and 9.9 times in the apical area was shown as compared with normal bone height model. Based on FEM analysis, 2.5 mm of alveolar bone loss can be considered as limit beyond which stress alterations were accelerated and the alveolar bone loss increases stress produced in PDL.

The possible clinical transfer from the current FEM study are as follows.

The improvements in this study are a 3D modeling of natural tooth, and the PDL width modeled with different widths instead of the average thickness around the root. The types of elements were quadratic tetrahedral which have three dof than do triangular elements.

At all bone heights, the stress values were found to be higher in relation to apex. This may explain the *de novo* occurrence of periapical abscess represented as periapical radiolucency in those teeth with primary trauma from occlusion without periodontitis.

With normal bone heights to compromised bone heights, the maximum values were concentrated on the labial site at point D_t, i.e., near the CEJ, this can explain the occurrence the abfraction process due to the excessive loads.

In few areas, tensile stresses were seen on the palatal, this can be explained as the angulation of the teeth which are naturally labial, and the forces directed onto the tooth tend to displace the tooth in labial direction due to which the compressive stresses result on the labial site and inversely the tensile stresses are seen on the palatal site.

There is a need to quantify stress produced by occlusal load and correlated with histologic findings that gets expressed as clinical and radiographic features.

The various clinical, radiographic, and histologic features of hypofunction resulting in periodontal atrophy characterized by decreased PDL space, thickened cementum, and accompanied bone loss have been presented.[12]

The results of the excessive loads applied in this study (240 and 290 N) would cause development of the typical histologic lesion of primary occlusal trauma. Alterations of the periodontium that have been associated with occlusal trauma will vary with the magnitude and direction of the applied force and location (pressure *vs* tension). These changes may include widening/compression of PDL, bone remodeling (resorption/repair), hyalinization, necrosis, increased cellularity, vascular dilatation/permeability, thrombosis, root resorption, and cemental tears.[2,13-21] Collectively, these changes have been interpreted as an attempt by the periodontium to adapt and undergo repair in response to traumatogenic occlusion. The compressive stress curves for the excessive loads used in this study showed the highest ligament stress at the crest of the alveolar bone. This pattern is consistent with the widening of the PDL space that occurs with the lesion of occlusal trauma. Values decreased when measured in an apical direction but again increased at apex, suggesting a relation of excessive stress at the apex to periodontal destruction in the periapex in primary trauma from occlusion.

As it is accepted, PDL is most formative tissue in the periodontium, and its important role in healing and periodontal regeneration after periodontal therapy has been proved. Reducing the stresses in the PDL may provide a better condition for this tissue to continue its regenerative and physiologic functions. Therefore, when there is occlusal trauma, occlusal therapy may result in better healing and more regeneration than without occlusal therapy. If the maximum allowable/permissible stress in the PDL is found, there can be a more accurate discussion in the findings of this study. Until then, we content ourselves with the multiplications produced at the produced stresses to

be out of the PDL biologic tolerance range. Data provided by this study are in agreement with several studies,[22-25] though not conducted by FEM, and also in disagreement with others.[26,27]

It must be noted that as bone remodeling (resorption and apposition) occurs as a result of compressive stress, the widened PDL would likely retain less compressive stress per unit area, thereby limiting the mechanical stimuli for further resorption. This model, therefore, can only represent the condition that initiates occlusal traumatism and not the dynamic changes that accompany the formation of the lesion. A similar explanation was suggested by Reinhardt et al.[11]

Further improvised finite element studies have modeled the PDL as either a nonlinear fiber-reinforced material or with viscoelastic properties to study orthodontic tooth movement,[28-34] and also FEMs have been based on micro-computed tomography scans for detailed descriptions of both the external geometry and the internal morphology of the alveolar bone.[35]

Displacement

Initial tooth displacement, or mobility of a tooth, has been used for evaluating biophysical properties of the periodontium as a diagnostic aid to periodontal problems.[36-40] Patterns of initial displacement of a tooth may be influenced by such anatomic variables as dimensions of the tooth and alveolar bone, widths of the PDL space, and mechanical properties of the periodontium.[21,41] However, the degree to which the biomechanical responses of the tooth are affected by tooth and bone geometry is not clear.

Different patterns of tooth displacement may be related to a change in the amount of displacement[27,97,109] or a change in the displacement pattern described with the centers of rotation and the center of resistance of the tooth.[41-44]

The maximum tooth displacement for all the loads applied in this study at normal alveolar bone height was noted at the incisal edge with the minimum tooth displacement at the cervical third of the root. The pattern of tooth displacement was changed by different alveolar bone heights. An increase of the tooth displacements was pronounced at the cervical third with the minimum displacement shifting apically to the junction of the cervical and middle thirds for the 25 and 50% loss of bone support, and further shifting apically to the junction of the middle and apical thirds for the 75% alveolar bone loss.

There are no studies of tooth displacement occurring due to the normal occlusal load. However, from the available literature, the results of this study could be correlated to the finite element study by Tanne et al[45] in which an orthodontic load of 100 gm was used to study the effects

of root length and alveolar bone loss on patterns of initial tooth displacement which was found to be 1 mm for average root length of 13 mm.

Clinical Implications

The compressive values were subject of the clinical interest in this study, since these values appear to play the greatest role in the initiation of bone resorption and lesion of occlusal trauma.

During occlusal discrepancy correction, the results of this study will be useful to measure mechanical values of the stress before and after treatment, provided a chairside stress measurement device is made available. The need of the hour is to manufacture a chairside measuring device to help academicians and practitioners.

So far, there is no report of numerical stress values being presented for the changes brought about in supporting periodontal tissues.

It was written subjectively as "more of stresses being produced as the alveolar bone levels reduced."

The advantage of FEM study is that we are able to show the changes in numerical stress values at normo-, hyper-, and hypo-occlusal loads. The normal occlusal load is applied as per the literature information. However, hypofunctional and two hyperfunctional loads are hypothetical situations to determine the possible numerical stress values at that point of static application of occlusal load. In a given mouth, there is a great variation in occlusal load on every tooth depending on the type of food consumed, physiologic activity, muscular action, and other parafunctional habits.

The determination of stress values in a functional mouth is the need of the hour to decipher the role of occlusion on periodontal tissues. The future direction on this raw topic requires the comprehension of biologic adaptation by the mechanical engineers to design a method to evaluate most complex issue of structural and functional interplay.

Further, the histologic evaluation of periodontal tissue changes should be correlated with stress values using FEM.

Limitations

The tooth is treated as pinned to the supporting bone, which is considered to be rigid, and the nodes connecting the tooth to the bone are considered fixed. This assumption will introduce some error; however, maximum stresses are generally located in the cusp/incisal edge area of the tooth.

It must be noted that as bone remodeling (resorption and apposition) occurs as a result of compressive stress, the widened PDL would likely retain less compressive stress

per unit area, thereby limiting the mechanical stimuli for further resorption; this model, therefore, can only represent the condition that initiates occlusal traumatism and not the dynamic changes that accompany the formation of lesion.

The shape of the tooth described in this study represents the most common morphologic feature of maxillary central incisor. However, the variations in the morphologic conditions among the normal individuals may affect the applicability of this analysis. The progress in the finite element analysis will be limited until better-defined physical properties for enamel, dentin, PDL, and cancellous and cortical bone are available.

Despite the disadvantages of FEM, it can be regarded as useful tool to visualize stress in the periodontal structures. The animal studies are also taken into account for assessing most of the periodontal pathology and the main drawback would be the true morphological reflection of the human supporting tissues is not practical. Although no true analog of human periodontal tissues has yet been seen, animal studies are indispensable to study various tissue changes in periodontal literature. To highlight or emphasize the advantage of FEM, the actual physical properties of the materials involved can be simulated, thus this method is the nearest that one could possibly get to simulate the oral environment *in vitro* with the existing computer knowledge.

Further Research

A stress analysis inside the PDL would be possible by means of processing the results obtained with our model, when a micromechanical model of PDL is made available.

Further, no quantitative guidelines exist to assist clinicians in making proper adjustment of traumatic occlusion, orthodontic force, for controlled tooth movements, and placement of implants so that the stress in the supporting structures gets evenly distributed. The FEM has been tried in this aspect but with certain approximations and assumptions. Therefore, further studies should be done to correlate the effects of frictional increases in the loads of the dynamic occlusion to the changes in the periodontal tissues.

CONCLUSION

Based on the findings of the present study, there is reasonably good attempt to express numerical data of stress to be given hypofunctional, normal occlusal, and hyperfunctional loads to simulate clinical occlusal situations which are known to be responsible for healthy and diseased periodontium. An attempt to measure the numerical stress levels to the various loadings definitely demonstrates changes in stress levels within periodontium on different surfaces of the tooth. The study of excessive loads in terms of primary trauma from occlusion and with grades of bone loss for secondary trauma from

occlusion has clearly demonstrated variations in stress reactions of various periodontal tissues. At this juncture, the requirement is to assess the various occlusal forces to its histologic effects in an *in vivo* study. Considering the dynamicity of occlusion, the possibility of studying the histologic changes to representative excessive loads is highly questionable.

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