ORIGINAL ARTICLE

Mechanical responses to orthodontic loading: A 3-dimensional finite element multi-tooth model

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Introduction: The initial mechanical response to orthodontic loading comprises biologic reactions that remain unclear, despite their clinical significance. We used a 3-dimensional finite element analysis to investigate the stress-strain responses of teeth to orthodontic loading. **Methods:** The model was derived from computed tomography data, with adequate boundary conditions and tissue characterization, with orthodontic hardware to provide a more accurate reflection of events during orthodontic therapy. This study also incorporated the adjacent dentition. Two cases were analyzed: a single-tooth system with a mandibular canine, and a multi-tooth system consisting of the mandibular incisor, the canine, and the first premolar, subjected to orthodontic tipping forces. **Results and Conclusions:** The systems experienced elevated distortion strain energies in the alveolar crest, whereas the tensile and compressive stresses coincided with the apical sites clinically associated with root resorption. Stress levels were considerably greater in the multi-tooth system than in the single-tooth system. The results for the single-tooth model agree with those previously reported. The numeric studies show how orthodontic tooth movement develops different stress fields and how root resorption might occur as a result of hydrostatic compressive stress-induced tissue necrosis. (Am J Orthod Dentofacial Orthop 2009;135:174-81)

Orthodontic forces generate a compound set of mechanical stimuli triggering biologic reactions in associated periodontal ligament (PDL) and dentoalveolar bone, thereby causing teeth to move to more appropriate positions in the jaw. For this reason, the biomechanics of orthodontic tooth movement has been an increasing area of investigation in the dental research community as part of a continuous effort to improve the clinical efficiency and outcomes of the treatment.[1] However, due to the anatomic and material complexity of tooth-supporting apparatus (alveolar bone, PDL, and root cementum), it is not easy to quantitatively determine mechanical responses to orthodontic loads without developing a realistic biomechanical model.

Orthodontic tooth movement (OTM) is the result of biologic reactions to externally applied mechanical stimuli.[2] The optimal application of orthodontic force enables maximum movement of teeth with minimal irreversible damage of the PDL, alveolar bone, and teeth. Therefore, knowing the relationships between the change in the induced mechanical strain fields and corresponding biologic reaction is crucial. In contrast, disregarding the mechanical response during orthodontic therapy not only might make OTM inefficient and extend its duration unnecessarily, but also might cause negative consequences. It was shown that excessive stresses during orthodontic loading can lead to substantial degradation of tooth tissue, reduced functionality, and disappointing esthetic and clinical outcomes.[3]

Tooth tipping is the most widely used approach in orthodontic treatment.[4] The relationship between the applied orthodontic force systems and the resulting OTM has been established empirically, largely derived from clinical experiences of various scenarios. However, a recent study linked the stress or strain induced in

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Submitted, October 2006; revised and accepted, March 2007.
0889-5406/$36.00

doi:10.1016/j.ajodo.2007.03.032

the PDL and the surrounding alveolar bone by orthodontic forces to the bone remodeling process in a quantitative fashion.[5]

Typically, 4 phenomenologic phases are used to describe remodeling related to OTM, each specifically linked to particular biologic reactions: initial tooth movement, arrest of tooth movement, acceleration of tooth movement, and constant linear tooth movement.[2]

In the initial phase, peak movement occurs for the root subjected to the highest pressure concomitant with a reduction of thickness of the PDL. The clinical data indicate that, for a maxillary canine, the peak movement coincides with peak stress induced by orthodontic tipping forces, and it is accommodated by bone remodeling of the alveolar crest.[6] In contrast, during the final OTM phase, the PDL and the surrounding alveolar bone experience continuous remodeling that yields a constant rate of tooth movement. However, a previous study suggested that the stress field during initial loading is the primary implication to trigger bone remodeling and determine OTM.[7] This confirms Wolff's law, which is the basis for analysis of the remodeling of bone after orthopedic procedures.[8,9]

According to Chan and Darendeliler,[10] a potential consequence of orthodontic treatment is root resorption that often is described as surface craters or root shortening. It was further shown that root resorption develops in the areas with higher pressures (hydrostatic stresses) around the root surface.[11] However, root resorption induced by orthodontic treatment is variable and lacks a quantitative basis, and, if severe, it can largely compromise the expected tooth movement.[12,13]

During orthodontic treatment, the PDL is exposed to considerable strains, which, combined with its relative incompressibility, greatly increase the level of hydrostatic pressure. If hydrostatic pressure exceeds systolic pressure (120 mm of mercury or 16 kPa), necrosis can occur in the PDL region.[14] The associated deprivation of tissue nutrition triggers the formation of giant cells to consume the necrotic tissue, and raw, unprotected cementum surfaces of the root are susceptible to the attack of resorptive cells.[15,16] Thus, the resulting focal necrosis in the PDL might trigger resorptive reactions of the root and breakdown of the alveolus. The unfavorable clinical outcomes of necrosis in the PDL were thoroughly addressed by Chan and Darendeliler[17] and Dye et al.[18] Thus, it is necessary to explore a quantitative relationship between stress magnitudes and root resorption.

A precise quantification of the stress-strain response of tooth, PDL, and bone to orthodontic loading cannot be achieved without a biomechanical model able to capture the anatomic complexity and biologic variability of the dental system involved. Moreover, it is suggested that the sophistication and accuracy of a reconstructed dental model can supplement an in-depth analysis of the mechanisms responsible for bone remodeling and root resorption during OTM.

Our aim in this study was to provide a better understanding of the stress-strain patterns in the initial stage of OTM. We used a single-tooth model and then extended to a multi-tooth model while incorporating orthodontic hardware, thus yielding a more realistic representation of the oral loading environment during OTM. Based on the mechanics and models established, the stresses were interpreted in the light of previous investigations of biologic reactions during OTM.

MATERIAL AND METHODS

To acquire the geometry, an adult mandible was precisely reconstructed based on computed tomography imaging. The skeletal specimen was obtained under the standard protocol of the New Zealand Anatomy Act. The computed tomography images (Somaton Plus Imager, Siemens, Greenville, Wis) were taken at 1-mm thickness with 0.5-mm interpolations, yielding a stack of 117 slices.

These slices were processed by using digital edge detection technology for the cortical bone, alveolar bone, enamel, and dentin. The edges were then used for generating the 3-dimensional geometry with commercial computer-aided design software (Rhinoceros 3D, Robert McNeel & Associates, Seattle, Wash). The model was completed by adding the orthodontic hardware that was modeled with the same software based on the manufacturer's technical specifications. Two cases were considered in this study: a single-tooth system and a multi-tooth system. The orthodontic hardware included bracket, adhesive, and wire (Fig 1).

A commercial finite element model (FEA) software package was used for the numeric analysis (CosmosWorks 2004, Structural Research and Analysis, Los Angeles, Calif). A particular aspect in modeling biologic structures such as dental devices is a proper approximation of the complex geometry involved to capture stress gradients.[19,20] In this study, the models were meshed by using 10 node tetrahedral elements, whose quadratic shape allowed capturing the complex curved surfaces in the model and provided better modeling accuracy theoretically and practically.[21] The reference mesh size in both the single-tooth and the multi-tooth models was 1.2 mm, and the element size was varied to accommodate the small features in the models—eg, the PDL.

A convergence test of linear elastic analysis was used to estimate the effect of various mesh densities on

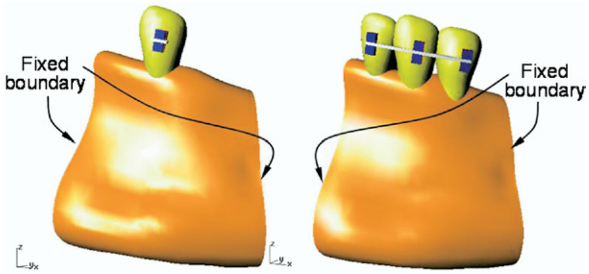

Fig 1. Single-tooth (left) and multi-tooth (right) models reconstructed for FEA with boundary conditions in the extents of cortical and alveolar bones.

discretization approximation in terms of the total strain energy.[21] After this trial, the single-tooth model consisted of 23,565 elements, and the multi-tooth model contained 32,812 elements. In the multi-tooth system, a surface-to-surface contact between the adjacent teeth was applied to model their interactions under orthodontic loading; this requires a nonlinear finite element solver to accommodate the unilateral contact condition.[21]

The FEA models replicate an orthodontic tipping system by using uniform loading conditions directed normally to the wire section. Thus, the orthodontic force applied to the wire is distributed effectively by the bracket and bonding cement onto the crown of the tooth and the surrounding dental structures. The far-field extents of the cortical bone and alveolar bone are fully fixed (clamped) as the boundary condition to prevent displacements from loading as shown in Figure 1. A force component of 0.35 N was applied normally to the wire through the bracket in both the translational and rotational directions.[22] The resultant orthodontic tipping force experienced by each tooth was 0.5 N, which is achieved by combining the component forces as:

$$F = \sqrt{F_r^2 + F_t^2} = \sqrt{0.35^2 + 0.35^2}$$

where F_r and F_t are rotational and translational force components applied.

It was assumed that the structures responded to tipping load linearly, and all materials in this analysis were modeled as isotropic and elastic. Although the PDL is known to be a nonlinear visco-elastic material,[23,24] most finite element models incorporate linear elastic properties as pointed out by Cattaneo et al.[22] We used a similar approach in this study to consider the tipping force and stress at the initial stage of OTM, when the linear elastic properties of the PDL reflect the same stiffness as the initial behavior of the nonlinear PDL.[22]

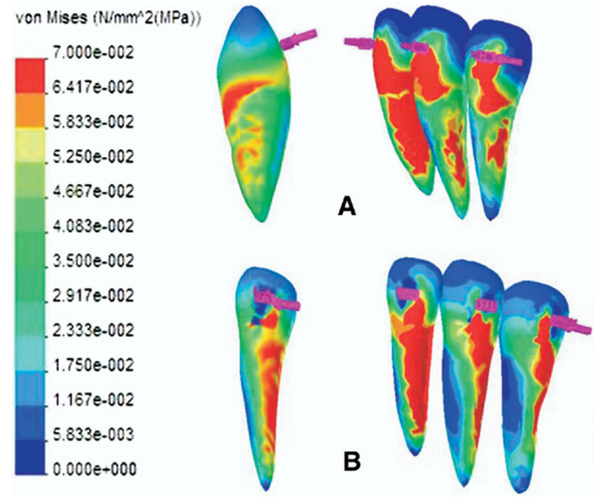

Fig 2. A, von Mises stress contours within the cementum structure for both the single and multi-tooth systems. High stress concentrations are present at the alveolar crest. **B,** The first principal stresses within the cementum structure (front view). Tensile stresses (red) are shown on the side where the orthodontic load is applied and compressive stresses (blue) are shown on the opposite.

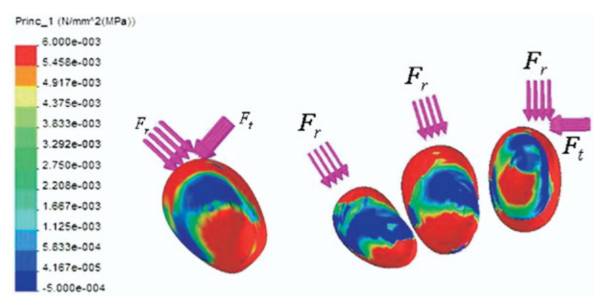

Fig 3. The first principal stresses experienced within the PDL root apex. Tensile (red) and compressive (blue) stresses are present (apical view) in the single tooth system (left) and the multiple tooth system (right) under translational and rotational orthodontic force components.

The material properties for the ligament and other dental components were taken from O'Brien[25] and are summarized in Table I.

In the postprocessing, the biomechanical responses in the cementum, PDL, and alveolar bone to initial OTM were quantified by analyzing and comparing the von Mises stresses, principal stresses, strains, and displacements in both the single-tooth and multi-tooth systems. We paid attention also to the von Mises or the equivalent stress-strain as a nondirectional measure of the distortion energy in the calcified tissue. These quantities are associated with the overall changes of

Table I. Material properties required in the FEA model*

Material	*Young's modulus (MPa)*	*Poisson's ratio*
Enamel	84 100	0.33
Dentin	14 700	0.31
PDL	1.18	0.45
Alveolar bone	490	0.30
Cortical bone	14 700	0.30
Orthodontic wire	210 000	0.30
Orthodontic bracket	210 000	0.30
Bonding cement	2 140	0.31

*Data from Poppe et al.[25]

shape in these biomaterials and the microdamage in the tissues.[26,27] More importantly, the microdamage, effective strain, and strain energy density can be treated as mechanical stimuli under multi-axial stress states resulting from orthodontic forces. In a biomechanical sense, the mechanical stimuli lead to biologic remodeling and drive the desirable OTM.

Of particular interest were the magnitudes of the pressure inside the PDL, especially in comparison with the systolic pressure, assessed by using hydrostatic stress determined with the following formula:

$$\text{Hydrostatic stress} = \frac{(\sigma_1 + \sigma_2 + \sigma_3)}{3}$$

where σ_1, σ_2, and σ_3 are the first, second, and third principal stresses, respectively.

RESULTS

In the single-tooth system, Figure 2, *A*, left side, shows the von Mises stress contours in the cervical region on the outer surface of the root (corresponding to the cementum layer), where they are concentrated (88 kPa). The tensile stresses appear on the distal side of the root (corresponding to the application site of the orthodontic force) with a maximum value of 33.52 kPa, whereas the compressive stresses occur on the opposing side with a maximum magnitude of 9.35 kPa (Fig 2, *B*, left). The peak von Mises stress in the cortical bone reaches 236.3 kPa compared with 34.2 kPa in the cancellous bone region.

The PDL exposed to an orthodontic tipping load undergoes both tensile and compressive principal stresses (Fig 3, left). The maximum tensile stress in the PDL structure is 110 kPa, whereas the maximum compressive stress is 12.8 kPa. Since the strain distribution in the PDL is of special interest in determining remodeling, the peak tensile strain is recorded in the cervical and apical regions (Fig 4, left).[8]

In the alveolar bone, the zones of tensile and compressive stresses can be clearly observed in the distal area, which correlates to the direction of force application (Fig 5, left). In the cancellous bone, the maximum tensile and compressive principal stresses are 20.2 and 12.7 kPa, respectively (Fig 5, *A*, left). In contrast, the maximum tensile and compressive stresses for the cortical bone are 110.1 and 9.2 kPa, respectively, situated at the alveolar crest (Fig 5, *B*, left). Significant differences of the peak tensile stresses and a small difference of the peak compressive stresses can be observed between the cancellous and cortical bones. It was also found that the hydrostatic compressive

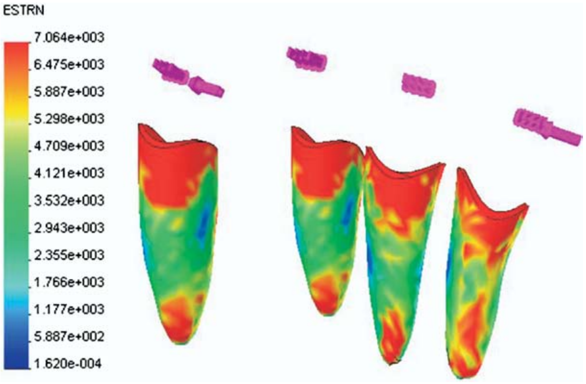

Fig 4. Major strains induced within PDL as a result of orthodontic loading in the single tooth system (left) and multiple tooth system (right). The areas of high strain are present at the alveolar crest and root apex.

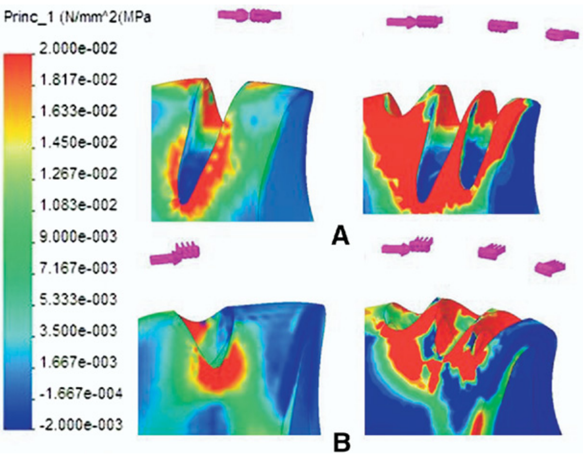

Fig 5. A, The first principal stresses within the cancellous bone for the single (left) and multiple tooth (right) systems. Tensile stresses (red) suggest bone formation while compressive stresses (blue) suggest bone resorption (back view). **B,** The first principal stresses within the cortical bone for the single (left) and multiple (right) tooth systems. Tensile stresses (red) suggest bone formation while compressive stresses (blue) suggest bone resorption (back view).

stress of the single-tooth system was 6 kPa at the root apex.

In the multi-tooth system, the structural responses differ from those of the single-tooth system because the adjacent teeth significantly alter the stress distributions and their magnitudes.

Figure 2, *A*, right, shows that the maximum von Mises stress in the cementum is 192.6 kPa and is situated at the right central incisor. The distribution of von Mises stress reflects the additional influence of adjacent teeth. Again, the von Mises stresses in the cortical bone exceed those in the cancellous bone. The peak von Mises stress in the cortical bone structure reached 287.8 kPa compared with 125.6 kPa in the cancellous bone. The maximum von Mises stresses in both cancellous and cortical bones are located near the alveolar crest of the right central incisor, whose magnitudes are higher than those in the single-tooth system.

The distributions of tensile and compressive stresses in the multi-tooth cementum are similar to those in the single-tooth cementum (Fig 2, *B*, right), where the maximum tensile and compressive stresses in the multi-tooth cementum are 132.3 and 14.6 kPa, respectively. The increase of the stress magnitude in the PDL was also found in the multi-tooth system (Fig 3, *right*); the maximum tensile and compressive principal stresses were 324.8 and 126.3 kPa, respectively. As shown in Figure 5 (right), the peak tensile stress in the cancellous bone was 114.5 kPa, and the maximum compressive stress was 8.4 kPa (Fig 5, *A*, right). The maximum tensile and compressive stresses in the cortical bone were 135.5 and 11.4 kPa, respectively (Fig 5, *B*, right). The hydrostatic stresses of the multi-tooth system at the root apex were 32 kPa in the right central incisor, 4 kPa in the right canine, and 0.8 kPa in the right first premolar.

The displacements of the multi-tooth system (Fig 6, *right*) also surpassed those of the single-tooth system (Fig 6, *left*): 7.0 and 6.2 μm, respectively. The maximum displacement was located in the crown component of the tooth, whereas the minimum displacement was near the center of resistance (or center of rotation[28]). There were differences in crown displacements between the single-tooth and the multi-tooth systems that reflect the effect of the additional dentition on the stiffness of the dental system.

A comparative view of the numeric results of both the single-tooth and multi-tooth systems are summarized in Tables II and III, respectively.

DISCUSSION

We analyzed the biomechanical changes induced by orthodontic loading in the supporting apparatus of the tooth. The results provide important information to assist in interpreting some complex biologic reactions from complex orthodontic force systems.

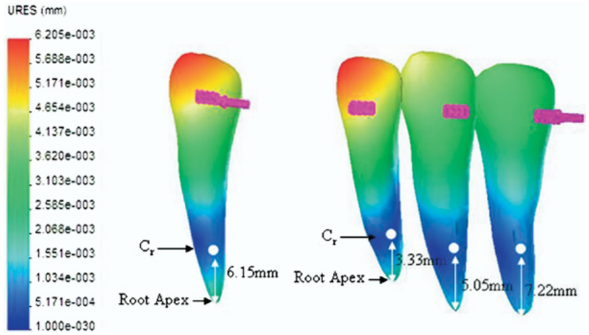

Fig 6. The displacement within the single tooth system (left). The maximum displacement (red) occurs in the crown while the minimum displacement (blue) is predominant in the root (front view). The distance between the center of rotation (C_r) and the root apex is also shown. The displacement within the multi-tooth system (right). The maximum displacement (red) occurs in the crowns while the minimum displacement (blue) is predominant in the roots (front view). The distances between the center of rotation (C_r) and the root apex of each tooth are also shown, where the larger the distance, the smaller the displacement in the crown.

This study complements the previous FEA analysis of OTM by modeling the detailed periodontal and alveolar support and considering the orthodontic loading hardware.[1,4,5,6,29] Furthermore, after the study by Cattaneo et al,[22] the use of high-order elements in this study allowed for a more precise analysis of the curved boundaries of natural teeth and also the stress-strain gradients that develop.

The distribution of the stresses in the single-tooth system agrees with the studies by Shaw et al,[6] Rudolph et al,[4] and Tanne et al.[29] The disto-cervical region of the root experiences the greatest stress, corresponding to the orthodontic force application area. This finding confirms the results of a previous investigation by Polson et al.[30] Such stress concentrations on the tooth surface are closely associated with root resorption craters found in an earlier study.[10]

This study has further highlighted the significant role of the PDL in bone remodeling after OTM. It is shown that compressive hydrostatic stresses are generated in the PDL under orthodontic tipping load, and thus the ligament might recruit mechanical stimuli to start tooth movement. Hydrostatic stresses greater than systolic pressures provide an indication of the rather large constant deformation of the PDL required in recruiting osteoblasts to facilitate OTM. These observations supplement the previous biologic findings of Melsen[5] and Shaw et al.[6]

Table II. FEA results of single-tooth system with orthodontic loading conditions

	von Mises stress (kPa)	*First principal tensile stress (kPa)*	*First principal compressive stress (kPa)*	*Strain (%)*	*Displacement (μm)*
Dentin	87.6	33.5	−9.34	0.00003	0.621
PDL	235.2	110.1	−12.8	0.2	
Alveolar bone	34.2	20.2	−12.7	0.00075	
Cortical bone	236.3	110.1	9.24	0.00025	

Table III. FEA results of multi-tooth system with orthodontic loading conditions

	von Mises stress (kPa)	*First principal tensile stress (kPa)*	*First principal compressive stress (kPa)*	*Strain (%)*	*Displacement (μm)*
Dentin	192.6	132.3	−14.6	0.0001	7.00
PDL	324.0	324.8	−126.2	0.2	
Alveolar bone	125.6	114.5	−8.4	0.0008	
Cortical bone	287.8	135.5	−11.4	0.0003	

The low elastic modulus of the PDL results in considerable strains after orthodontic loading.[7] The high strains in the PDL indicate the potential sites for the bone remodeling process. Melsen[5] showed that the areas stretched in the PDL are closely related to the bone remodeling process and suggested that the stretched fibers of the PDL would most likely lead to the generation of sufficient strain energy to induce remodeling. The strain energy distribution in the PDL influences the adjacent stresses in bordering calcified tissues that indicate remodeling.

The maximum strains in the PDL occur at the alveolar crest and the root apex; these sites have the greatest deformation and bone remodeling rates. From a clinical perspective, moderate peak strains about the center of resistance of the teeth is desired to prevent orthodontically induced root resorption.[30]

In this study, consequent to the direction of applied orthodontic force, the principal stress distributions in cancellous and cortical bone point to the regions of bone formation (tension) and bone resorption (compression). These different stress outcomes anticipate the direction of tooth movement; this is consistent with the results of Bonafe-Oliveira et al.[11] This relationship between the type of orthodontic load and bone biomechanical response is now a general consensus in orthodontics.[2]

In contrast to the single-tooth system, the multi-tooth system showed overall higher structural responses to orthodontic tipping loading. Although the distribution pattern of stresses and strains in the multi-tooth system was remarkably similar to that in the single-tooth system, their magnitudes were considerably greater. This can be explained in part by the presence and the rigidity of the adjacent teeth.

The compressive strains induced in the PDL anticipate the direction of tooth movement as a consequence of orthodontic loading. More importantly, the adjacent dentition causes greater compressive stresses because the PDL is compacted by the rigidity of the neighboring dentin structure; this demonstrates the necessity of considering mutual influences between the teeth.

In the multi-tooth system, the teeth are in contact with one another, and hence the orthodontic forces are transmitted throughout the entire system. In this study, the central incisor had no contact with its homologue across the midline, and thus it experienced the greatest displacement that gave rise to the highest mechanical stress. Although the 3-tooth system is limited in terms of orthodontic force transmission along the entire dental arch, it yields a more realistic kinematic relationship compared with previous studies of a single-tooth system.[1,5,29,31] This study suggests that multi-tooth systems should be considered for further biomechanical analysis because they replicate clinical situations more realistically.

Particular interests are in the determination of the hydrostatic stresses in the PDL. Although in the single-tooth system magnitudes of only 6 kPa occur in the root apex, in the multi-tooth system they reach 32 kPa at the root apex in the central incisor, 4 kPa in the canine, and 0.8 kPa in the first premolar. These hydrostatic stress results for the root apex of the central incisor are much greater than the systolic pressure (16 kPa), and, therefore, this tooth is susceptible to root resorption. These preliminary findings necessitate further investigations into hydrostatic pressure during orthodontic treatment that might provide new methods to predict and prevent root resorption.

A major implication of this study is the need to connect the application of forces more closely to actual orthodontically induced biologic processes by quantifying the biomechanical responses. Dental hard tissues possess physiologic features that can mediate the effect of the forces on the tissues.[32] Therefore, further work is required to establish more precise relationships between the various dental tissues and in-vivo force application systems. One perspective can be to incorporate the nonlinearity of the PDL into the future model.[22-24] It is suggested that the development of complex mathematical models, with experimentally determined material and anatomic data to depict a clinical situation realistically, might lead to a more accurate means to predicting biologic reactions and subsequently improve clinical outcomes.

CONCLUSIONS

We provided an anatomically accurate model and description of the stresses that develop during the initial stage of OTM. The finite element investigation initially considered a single-tooth model and then a multi-tooth model. The model incorporated orthodontic hardware, thus yielding a more realistic orthodontic environment during OTM. Within the limitations of our study, we concluded the following.

1. A multi-tooth system generates a more relevant model of the biomechanical environment by taking into account the force transmission between the teeth and the surrounding bone.
2. Hydrostatic stresses in the PDL can be ued as markers for predicting the potential sites of root resorption.
3. The initial distortion energy distribution in the dental structure provides an indication of the consequential biologic reactions of OTM such as bone remodeling.

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