Dynamic Finite Element Analysis of the Human Maxillary Incisor Under Impact Loading in Various Directions

Haw-Ming Huang, MS, Keng-Liang Ou, PhD,† Wei-Nang Wang, DDS,‡,§ Wen-Ta Chiu, MD, PhD,* Che-Tong Lin, DDS, PhD,† and Sheng-Yang Lee, DDS, PhD**

Abstract

The aim of this study was to investigate fracture patterns occurring when a human upper central incisor is subjected to impact loadings at various angles. A two-dimensional finite element (FE) model of the maxillary incisor and surrounding tissues was established. The structural damping factor for the tooth was then calculated and assigned to the model. Dynamic FE analysis was performed to stimulate the associated impacts. Time-dependent traumatic forces at 0°, 45°, and 90° labially to the long axis of the tooth were applied to the model. Von Mises's equivalent stress contours within the FE models were calculated. Our results indicated that tooth damping lagged behind peak stress by 0.05 ms. In addition, we found that impact direction played an important role in terms of outcome for the fractured incisor. These results can, in part, explain the mechanisms underlying the alternative outcomes when upper incisors are subjected to impact.

Key Words

Finite element, fracture, incisor, dynamic analysis, stress

From the *Graduate Institute of Oral Sciences, †Graduate Institute of Oral Rehabilitation Sciences, ‡Institute of Injury Prevention & Control, Taipei Medical University, Taipei, Taiwan; and §Department of Neurosurgery, Municipal Wan Fang Hospital, Taipei, Taiwan.

Address requests for reprint to Dr. Lee, Sheng-Yang, DDS, Graduate Institute of Oral Rehabilitation Sciences, Taipei Medical University, 250 Wu-Hsing Street, Taipei, Taiwan. E-mail address: seanlee@tmu.edu.tw.

Copyright © 2005 by the American Association of Endodontists

The upper central incisor is the most frequently involved tooth in frontal impact. Clinical findings demonstrate that the outcomes for frontal tooth impacts typically involve crown, oblique root, oblique crown-root, or neck fractures (1). Although theoretical study indicates that force direction plays an important role in the propagation of fracture lines in the impacted tooth (2), the exact relationship between the angle of impact and the resultant fracture lines in the maxillary incisor remain unclear and without experimental evidence.

With the rapid advancement and development of computer technology, finite element analysis (FEA) is now widely used as an effective method for dental trauma analysis. Most FEA studies are simplified and involve a static force applied to the tooth (3–7), however, with the applying force assumed to be unchanged during the impact period and the damping effects of the tissues ignored. In the real world, however, traumatic injuries to the teeth typically result from a dynamic force, the magnitude of which is altered over time. Therefore, for traumatic analysis of the tooth, time-dependent behavior should be considered for different rates of loading (8). To assess the process of stress growth and fracture-line propagation in an impacted tooth, dynamic FEA can provide greater insight into the issue.

To conduct dynamic FEA, the viscoelastic properties of the test subject are needed for computation. The viscoelasticity of a material can be separated into two components: one is a perfect elastic solid, and the other is a viscous liquid. When a viscous material is subjected to an impact, the strain energy can be gradually converted to another energy form. Because of the reduction in the strain energy, the response, such as the deformation of the material, gradually decreases. The mechanism by which the strain energy is gradually converted to another energy form is known as damping. When an impact force is applied, the resultant stress of an elastic material is directly proportional to the strain. However, the resultant stress caused by an external strain is dictated by the rate of deformation (9). Therefore, damping is a nonlinear material property of a viscous material.

Because of the lack of quantitative scientific data, however, few dental studies have assigned nonlinear mechanical properties to their FE model (8, 10–12). In these nonlinear studies, the damping properties of periodontal ligament (PDL) and the intact tooth were evaluated by means of curve-fitting the experimental data to the vibrational behavior of mechanical models. Analyzing the results from these studies reveals that nonlinear FE analysis overcomes the problems inherent in approximation resulting from the adoption of simplified models.

Although the damping property of PDL is the main contributor to tooth viscoelasticity, the cushioning effect of other damping material, such as pulp, can affect internal stress distribution within the impacted structure. To better understand the viscous properties of the human tooth, in our previous study the damping ratio of the human maxillary central incisor was quantified by means of modal testing experiments (13). In the current investigation, we incorporated this experimental damping-ratio data for the incisor into an FE model. Dynamic FEA was used to investigate the stress concentrations and fracture-line propagation in an upper central incisor subjected to dynamic loads in various directions.

Materials and Methods

In this study, the finite element analysis package, ANSYS (Swanson Analysis System Inc., Houston, PA), was used to perform the transient dynamic analysis on a personal

Copyright © by the American Association of Endodontists. Unauthorized reproduction of this article is prohibited.

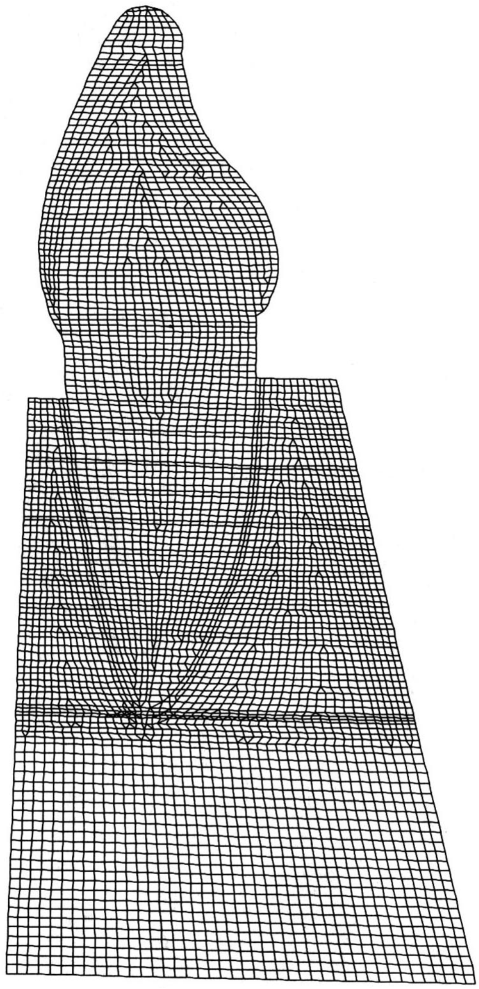

Figure 1. The 2-D strain finite element model used in this study.

computer. Transient dynamic analysis (also called time-history analysis) is a technique used to determine the dynamic response of a structure under the action of any general time-dependent loads. This technique is used for structural analysis of damping effects are considered to be important. To mesh the FE model more finely, a 2-D plane strain FE model of the human maxillary central incisor, containing enamel, dentin, pulp, periodontal membrane, alveolar bone, compact bone, and spongy bone was established (Fig. 1). The geometry and relevant dimensions of the incisor, including the length (23.5 mm) and thickness of the periodontal membrane (0.25 mm) were obtained from an atlas with anthropometry data sourced from a previous study (14). The alveolar process was located 2 mm apically from the cementoenamel junction (CEJ). Our model had a total of 5373 nodes and 5274 2-D quadrilateral elements, with the boundary conditions defined to prevent free body motion. The nodes on the base surface of the alveolar bone were fixed.

As shown in Table 1, the material properties for the FE model were adopted from the literature (14). In addition, the structural damping factor (β) for the model was derived according to the following formula:

$$\beta = \frac{\xi}{\pi f} \tag{1}$$

where ξ is the damping ratio, and f is the first resonance frequency of the upper central incisor (15, 16). In this study, the damping ratio

TABLE 1. Material properties used in the finite element model

	Young's Modulus (GPa)	Density (g/cm^3)	Poisson's Ratio
Enamel	77.90	3.00	0.33
Dentin	16.6	2.20	0.31
Pulp	0.00689	1.00	0.45
PDL	0.05	1.10	0.45
Alveolar Bone	3.50	1.40	0.33
Cortical Bone	10.00	1.40	0.26
Cancellous Bone	0.50	1.40	0.38

(14.6%) (13) and resonance frequency of human maxillary incisor (1388 Hz) (17) were adopted from experimental data. Accordingly, the structural damping factor of the upper central incisor assigned in our model was 0.33×10^{-4}.

A sinusoidal force with a peak of 800 N (18, 19), a rise time of 2 ms, and a total duration of 4 ms (20, 21), was chosen and imposed on a node on the facial crown. To assess the influence of impact direction on stress distribution and fracture-line propagation in the impacted incisor, three impact forces were applied using the model (Fig. 2): F1 simulated the traumatic force and acted horizontally to the labial crown; F2 was a traumatic force at 45° labial to the incisal edge; and F3 was a vertical force acting at the labial middle crown. In the various impact situations, Von Mises' equivalent stress contours within the FE models were displayed for comparison. The stress distribution at the root apex,

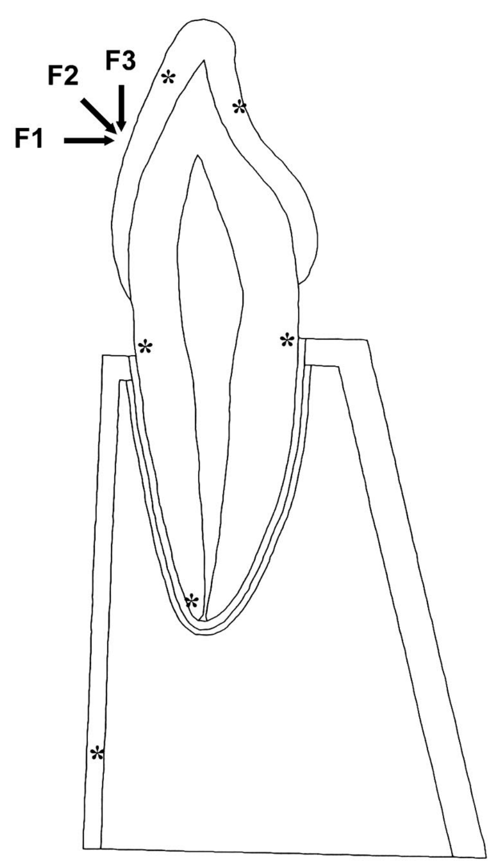

Figure 2. Three loading forces are applied to the labial middle crown of the model. F1, F2, and F3 represent loading at 0°, 45°, and 90°, respectively. Asterisks denote locations where Von Mises' stress was computed for comparison.

Copyright © by the American Association of Endodontists. Unauthorized reproduction of this article is prohibited.

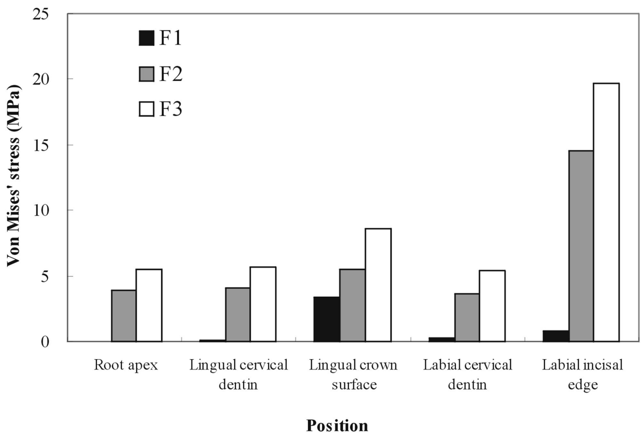

Figure 3. Maximum stresses at various positions are shown. This result appears to be consistent with the clinical observation that the outcomes for incisor fractures are dependent on impact direction.

lingual cervical dentin, lingual crown surface, labial cervical dentin, and labial incisal edge were calculated for fracture-line analysis.

Results

The Von Mise stress developed at various impact angles is depicted in Fig. 3. In each position, a vertical impact caused the highest stresses in the incisor. With the exception of the lingual crown surface, however, the stresses caused by horizontal impact are remained below 1 MPa.

During the horizontal impact (F1), stresses were concentrated in the crown area (Fig. 4*A*). Immediately after impact, stress greater than 10 MPa was demonstrated around the impact site. At 0.25 ms, the stresses were concentrated at the labial crown surface, with a tendency to propagate toward the lingual side. The maximum developed stress in the model occurred at 2.05 ms, with the lingual crown surface independently exhibiting stresses over 3 MPa at this time. Thereafter, the stress decreased over time.

When the upper incisor was subjected to an oblique impact (F2), high stresses first developed in the area around the impact site, then at the incisal edge, lingual crown surface, and the area around the root apex (Fig. 4*B*). At the impact site, the stress was 10 MPa at 0.25 ms, climbing to over 50 MPa at 1 ms. Stresses developed independently at the lingual crown surface and root apex at 1 ms reaching peaks of 10 and 5 MPa, respectively, at 2.05 ms, and decreasing subsequently. At 3.5 ms, only the stress around the impact site remained high ($\geq$10 MPa).

When a vertical load (F3) was applied to the model, the stress distribution patterns were similar to those generated with the oblique impact (Fig. 4*C*). The magnitudes of the concentrated stresses were higher for the vertical load, however. Interestingly, at 2.05 ms, highly concentrated stresses of over 5 and 10 MPa developed on both sides of the cervical dentine and the labial side of alveolar socket wall, respectively. Continuously tracing the time profiles of the stress contours revealed that the stresses in each area were developed independently and were propagated simultaneously.

Discussion

The finite element model used in this study is a 2-D *plane-strain* model. Mathematically, a 2-D plane-strain FE model is a pseudo 3-D model used to provide a simplified 3-D calculation.

Material damping factor is defined as the fraction of strain energy lost in one full cycle of deformation (22). In our dynamic simulations of tooth impact, therefore, the damping properties of all structures were taken into consideration. We found that for each loading condition, peak stress (occurring at 2.05 ms) always lagged behind the maximum applied loading by 0.05 ms. That is, an impacted incisor with greater damping can lower the concentrated stresses by dispersing the strain energy over a longer period (13). This cushioning effect is the main principle behind the effectiveness of a mouth guard that is used for protection of tooth from severe impact injury (23).

The most affected teeth in dental trauma are the maxillary centrals in general. It was observed that the falls or collisions were the main causes (24). Clinically, fractures caused by frontal impacts fall into four categories according to the direction and position of fracture lines: (a) horizontal crown fracture; (b) horizontal fractures at the neck of the teeth; (c) oblique crown-root fractures; and (d) oblique root fractures (1). When a maxillary incisor is subjected to a horizontal force, stresses developing at the labial crown demonstrate a tendency to propagate toward the lingual side (Fig. 4*A*). Further, the lingual crown surface independently exhibits high stresses during the impact process. These phenomena, undoubtedly, result in horizontal fracture of the crown. Interestingly, except for the crown area, the stresses developed at the root and alveolar process were lower than 1 MPa (Fig. 3). Clinical observation also revealed that the crown fractures resulting from high-velocity impacts are usually not associated with damage to the supporting structures (25). Based on analysis of previous clinical profiles, it appears that the energy of the impact is absorbed by the supporting structures during tooth displacement (1). For tooth-trauma analysis, therefore, the damping properties of the tooth should be considered and the cushioning effects on strain energy dispersion cannot be ignored. Clinical retrospective studies have shown that endodontically treated teeth are more prone to fracture than vital teeth (26–28). Many studies discussed that endodontically treated teeth are weak because of loss of tooth structure. However, we found that the reduction in the capability of strain energy dispersion because of pulp tissue extraction may also play a role.

When our upper incisor was subjected to a vertical force, significant stress concentrations were demonstrated at the root apex and cervical dentin (Fig. 4*C*). Connecting imaginary lines between these high stress areas conformed to horizontal fracture lines appearing at the tooth neck, and oblique crown-root fractures and oblique root fractures. When upper central incisors are subjected to impact, therefore, the vertical component of the impacting force will contribute to formation of these three fracture types. On the other hand, previous studies have demonstrated that vertical root fracture seems to typically occur in a buccolingual direction through the thickest part of dentin (29). These findings match our finite element data well.

Apart from the four categories discussed above, enamel fracture is a frequent injury appearing as crazing within the enamel substance that does not cross the dentin-enamel junction (30). Clinical findings show that enamel fractures are caused by direct impact, frequently occurring on the labial surface of the upper incisor (31). Our numerical simulations show that the occurrence of such injury is caused by vertical impact force. Further, fractures of the alveolar socket wall are predominantly noted clinically in the upper incisor region (32). In fact, we found that occurrence is also associated with the vertical component of impact force. When connecting a pseudo fracture line between the labial alveolar socket wall and root apex (Fig. 4*C*), a pattern characteristic of marginal periodontal bone defect appears.

Based on dynamic FE analysis of the human central incisor, the following conclusions can be drawn:

1. Dynamic analysis may provide more realistic simulation of the mechanism of tooth injury because of traumatic impact. The method used in this study can serve as a useful reference for future advanced investigations.

 Copyright © by the American Association of Endodontists. Unauthorized reproduction of this article is prohibited.

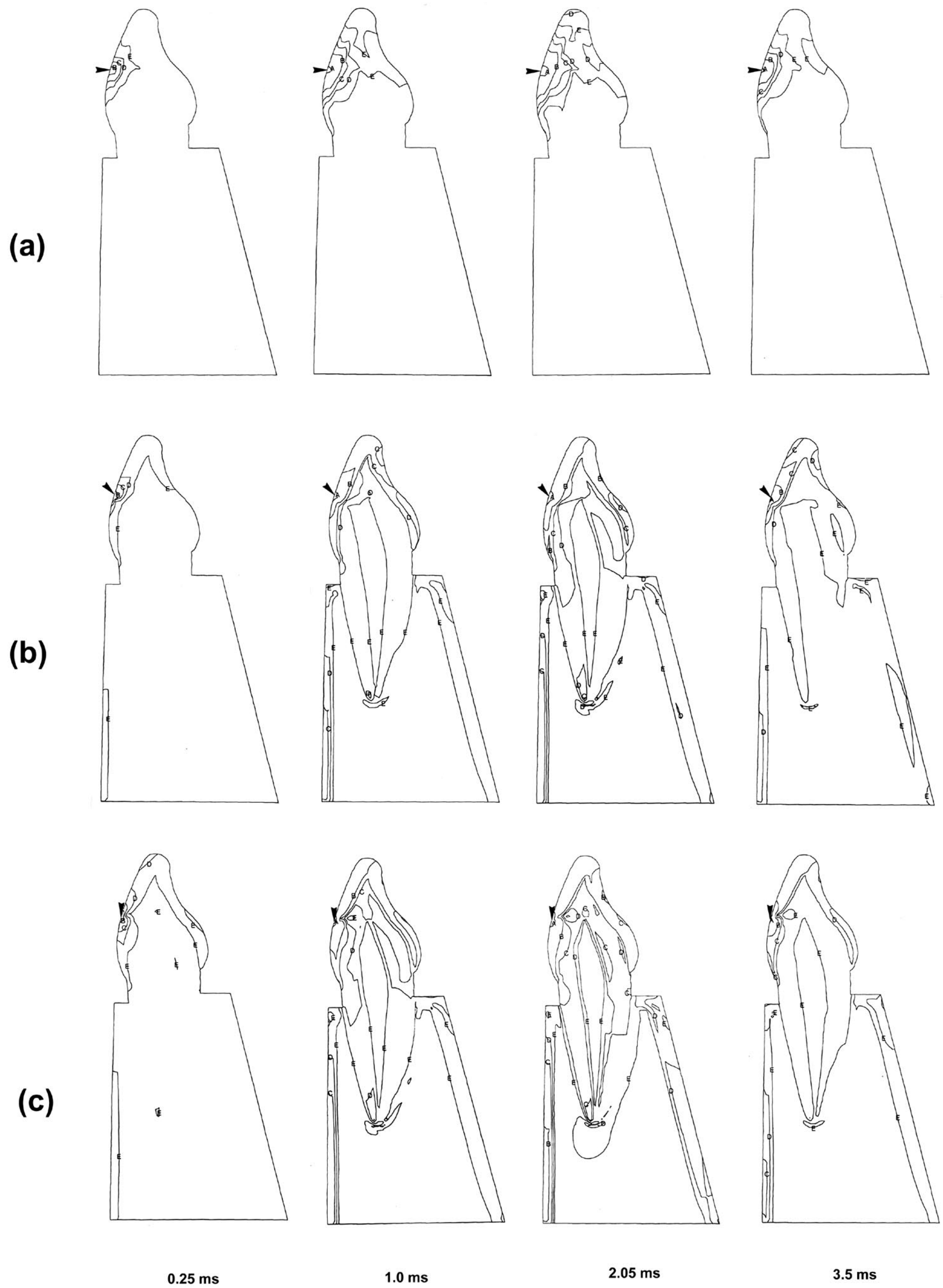

Figure 4. Contours of maximum stresses at 0.25, 1.0, 2.05, and 3.5 ms after onset of (*a*) horizontal impact, (*b*) oblique impact, and (*c*) vertical impact. Symbols A, B, C, D, and E denote 1, 3, 5, 10, and 50 MPa, respectively.

2. The stress contours produced when an upper central incisor was subjected to impact showed that stress concentrations at various positions developed independently.
3. Horizontal impact forces tend to cause horizontal crown fractures. The vertical component of the impact force tends to cause the horizontal fractures at the neck of the teeth, oblique crown-root fractures and oblique root fractures.

Acknowledgments

This study was supported by a grant (TMU90-Y05-A106) from Taipei Medical University, Taipei, Taiwan, and, in part, by a grant (NHRI-EX93-9106PN) from the National Health Research Institutes, Taipei, Taiwan.

References

1. Andreasen JO, Andreasen FM. Classification, etiology and epidemiology. In: Andreasen JO, Andreasen FM, ed. Textbook and color atlas of traumatic injuries to the teeth, 3rd ed. Missouri: Mosby Inc., 1994:151–80.
2. Lee HE, Lin CL, Wang CH, Cheng CH, Chang CH. Stresses at the cervical lesion of maxillary premolar: a finite element investigation. J Dent 2002;30:283–90.
3. Ricks-Williamson LJ, Fotos PG, Goel VK, Spivey JD, Rivera EM, Khera SC. A three-dimensional finite-element stress analysis of an endodontically prepares maxillary central incisor. J Endod 1995;21:362–7.
4. Ho MH, Lee SY, Chen HH, Lee MC. Three-dimensional finite element analysis of the effects of posts on stress distribution in dentin. J Prosthet Dent 1994;72: 367–72.
5. Kishen A, Kumar GV, Chen NN. Stress-strain response in human dentine: rethinking fracture predilection in postcore restored teeth. Dent Traumatol 2004;20:90–100.
6. Eskitascioglu G, Belli S, Kalkan M. Evaluation of two post and core systems using two

Copyright © by the American Association of Endodontists. Unauthorized reproduction of this article is prohibited.

different methods (fracture strength test and a finite element stress analysis). J Endod 2002;28:629–33.
7. Lertchirakarn V, Palamara JE, Messer HH. Patterns of vertical root fracture: factors affecting stress distribution in the root canal. J Endod 2003;29:523–8.
8. Natali AN, Pavan PG, Scarpa C. Numerical analysis of tooth mobility: formulation of a non-linear constitutive law for the periodontal ligament. Dent Mater 2004;20:623–9.
9. Chern Lin JH, Greener EH, Hanawa T, Okabe T. Dynamic mechanical properties of dental amalgams. Dent Mater 1990;6:41–4.
10. Middleton J, Jones M, Wilson A. Three-dimensional analysis of orthodontic tooth movement. J Biomed Engl 1990;12:319–27.
11. Okazaki M, Fukumoto M, Takahashi J. Damped oscillation analysis of natural and artificial periodontal membranes. J Biomed Engl 1996;24:234–40.
12. Toms SR, Eberhardt AW. A nonlinear finite element analysis of the periodontal ligament under orthodontic tooth loading. Am J Orthod Dentofacial Orthop 2003;123: 657–65.
13. Yao WC, Lee HF, Lee SY, et al. The effects of damping property on the response of maxillary incisor subjected to traumatic impact force. Chin Dent J 2004 (in press).
14. Lee SY, Huang HM, Lin CY, Shih YH. In vivo and in vitro natural frequency analysis of periodontal conditions, an innovative method. J Periodontol 2000;71:632–40.
15. Petyt M. Force response. In: Petyt M, ed. Introduction to finite element vibration analysis, 1st ed. United Kingdom: Cambridge University Press, 1990;387–9.
16. Rao SS. Fundamentals of vibration. In: Rao SS, ed. Mechanical vibrations, 2nd ed. NY: Addison-Wesley Inc., 1990:26–37.
17. Huang HM, Lee SY, Yeh CY, Wang MS, Chang WJ, Lin CT. Natural frequency analysis of periodontal conditions in human anterior teeth. Ann Biomed Engl 2001;29:915–20.
18. McDonald AV, King PA, Setchell DJ. An in vitro study to compare impact fracture resistance of intact root-treated teeth. Int Endod J 1990;23:304–12.
19. Johnston T, Messer LB. An in vitro study of the efficacy of mouthguard protection for dentoalveolar injuries in deciduous and mixed dentitions. Endod Dent Traumatol 1996;12:277–85.
20. Chu CS, Lin MS, Huang HM, Lee MC. Finite element analysis of cerebral contusion. J Biomech 1994;27:187–94.
21. Bobak V, Christiansen RL, Hollister SJ, Kohn DH. Stress-related molar responses to the transpalatal arch: a finite element analysis. Am J Orthod Dentofac Orthop 1997; 112:512–8.
22. Oka H, Yamamoto T, Saratani k, Kawazoe T. Application of mechanical mobility of periodontal tissue to tooth mobility examination. Med Biol Engl Comput 1989;27: 75–81.
23. Chapman PJ. Concussion in contact sports and importance of mouthguards in protection. Aust J Sci Med Sport 1985;17:25–8.
24. Celenk S, Sezgin B, Ayna B, Atakul F. Causes of dental fractures in the early permanent dentition: a retrospective study. J Endod 2002;28:208–10.
25. Dow PR. Thompson ML. Pulp management for the immature fractured anterior tooth. J Canad Dent Assoc 1960;26:5–9.
26. Hurmuzlu F, Kiremitci A, Serper A, Altundasar E, Herguner S. Fracture resistance of endodontically treated premolars restored with ormocer and packable composite. J Endod 2003;29:838–40.
27. Pontius O, Hutter JW. Survival rate and fracture strength of incisors restored with different post and core systems and endodontically treated incisors without corono-radicular reinforcement. J Endod 2002;28:710–5.
28. Ettinger RL, Qian F. Postprocedural problems in an overdenture population: a longitudinal study. J Endod 2004;30:310–4.
29. Lertchirakarn V, Palamara JE, Messer HH. Finite element analysis and strain-gauge studies of vertical root fracture. J Endod 2003;29:529–34.
30. Dominkovice T. Total reflection in tooth substance and diagnosis of cracks in teeth: a clinical study. Swed Dent J 1977;1:163–72.
31. Ravn JJ. Follow-up study of permanent incisors with enamel cracks as a result of an acute trauma. Scand J Dent Res 1981;89:117–23.
32. Andreasen JO. Etiology and pathogenesis of traumatic dental injuries. A clinical study of 1298 cases. Scand J Dent Res 1970;7:329–42.

Copyright © by the American Association of Endodontists. Unauthorized reproduction of this article is prohibited.