Dental Traumatology 2008; 24: 454–457; doi: 10.1111/j.1600-9657.2008.00618.x

Biomechanical model of incisor avulsion: a preliminary report

Jiro Miura, Yoshinobu Maeda

Division for Interdisciplinary Dentistry, Osaka University School of Dentistry, Osaka, Japan

Correspondence to: Jiro Miura, DDS, PhD, Division for Interdisciplinary Dentistry, Osaka University School of Dentistry, 1-8 Yamadaoka, Suita, Osaka 565-0871, Japan
Tel.: +81 6 6879 2386
Fax: +81 6 6879 2387
e-mail: miura_j@dent.osaka-u.ac.jp

Accepted 16 April, 2007

Abstract – The aim of the current study was to produce a two-dimensional finite-element model (FEM) of a maxillary incisor being palatally displaced with sufficient force to result in avulsion. The viscosity and elastic coefficient of bone were chosen from previous studies to model the properties of the periodontal complex. A ramped impact of 100 N was applied over a period of 1.5 ms in the direction perpendicular to the labial surface of the incisor model. Changes in the periodontal ligament (PDL) length (elongation and compression) started immediately after the impact and progressed to strain developing in the alveolar bone. In accordance with these changes in the PDL length, the tooth started to rotate with the center of rotation located at the edge of the palatal alveolar bone, combining the stress concentrations at the labial and cervical edges of the bone. When the elongation reached the designated limit, PDL disconnection occurred, leading to tooth avulsion. The resulting two-dimensional FEM is capable of modeling changes in the PDL length and compression, rotation and avulsion.

Tooth avulsion possibly occurs following labial alveolar bone fracture or periodontal ligament (PDL) disconnection. However, the viscoelastic behavior of the PDL and the alveolar bone may also affect the mode of avulsion (1). There have been only a few studies (2, 3) that investigate either stress distribution following external impact suffered by teeth, PDLs and alveolar bones or mechanisms of tooth avulsion. Research studies have found that the PDL plays an important role in the mechanisms for tooth trauma. Recently, to understand the mobility of human teeth in impact loading, nonlinear finite-element (FE) analyses were performed. It is well known that damping materials of PDL act as shock absorbers to minimize external impact. Although the damping properties of the PDL are known to be the main defense for preventing tooth fractures, details such as real-time stress distributions in tooth avulsion remain unknown.

The purpose of the present study was to (i) produce a two-dimensional FEM of a maxillary incisor being displaced palatally with sufficient force to produce an avulsion; and (ii) observe its stress distribution in real time.

Methods

A two-dimensional FEM of a human maxillary central incisor, containing enamel, dentine, pulp, PDL, alveolar bone and compact bone, was constructed for a dry skull with 8864 quadrilateral elements from the cross-sectional surface of CT DICOM data (Fig. 1). The thickness of the PDL was set at 200 μm according to previous reports (4, 5). The corresponding elastic properties, such as Young's modulus (E) and Poisson's ratio, were determined from the literature and summarized in Table 1 (6–10). Material properties of enamel and dentin were established from previous reports (6, 7). In these reports, the indentation method with credible parameter values was used to evaluate material properties. Cortical and cancellous bones were established from previous reports (8, 9). The high Poisson's ratios assigned to the pulp and the gingiva were chosen as representative values, but their material properties were not affected by the stress distribution in teeth (Table 1). To incorporate the damping behavior of the PDL, a nonlinear analysis was performed using a FE analysis package (MARC2002; MSC, CA, USA) by setting both the inter-node relationship of the stress–strain diagram and the designated limit, which was 1.4 times the original length. This designated limit was determined from stress–strain curves found in Toms (2003) and Provatidis et al. (2000) (4, 10). When the elongation reached the designated limit (4), PDL disconnection occurred and the stress between the nodes became zero. A ramped impact of 100 N was applied in the 90° direction labial to the incisal edge (Fig. 2). Constraints were applied to the cutting surface of the maxillary bone. The von Mises stress and displacement were continuously analyzed as real time evolved under the defined conditions.

Results

The von Mises stress contours of the upper incisor and the surrounding bone were displayed in real time (Fig. 3). Changes in the PDL length (elongation and compression) started immediately after the impact. Meanwhile translation and rotation of the tooth in the alveolar socket occurred 0.5 ms after the impact with stress beginning to develop in the alveolar bone on both the palatal and labial sides. Subsequently, the tooth

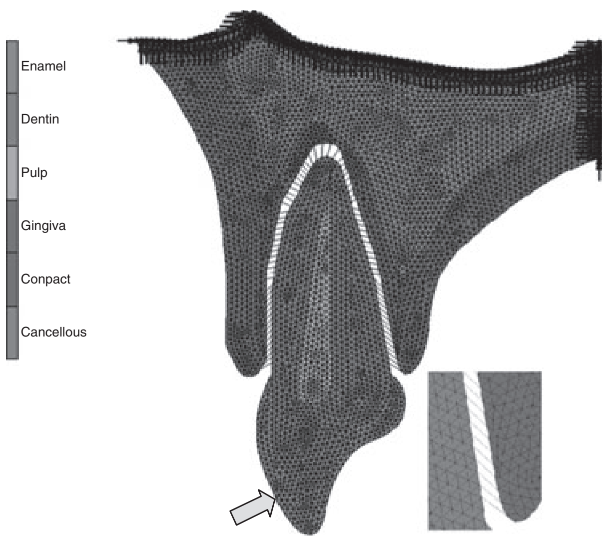

Fig. 1. Two-dimensional finite-element model of the tooth–bone complex with 8864 quadrilateral elements from the cross-sectional surface of CT DICOM data. An impact of 100 N is applied in the 90° direction labial to the incisal edge (arrow). Fixed nodes are set at the surface of the maxillary bone.

Table 1. Material properties of finite-element model

	Young's modulus (MPa)	Poisson's ratio
Enamel	50 000	0.3
Dentin	18 600	0.31
Cortical bone	11 500	0.33
Cancellous bone	431	0.3
Pulp	2	0.45
Gingiva	200	0.45

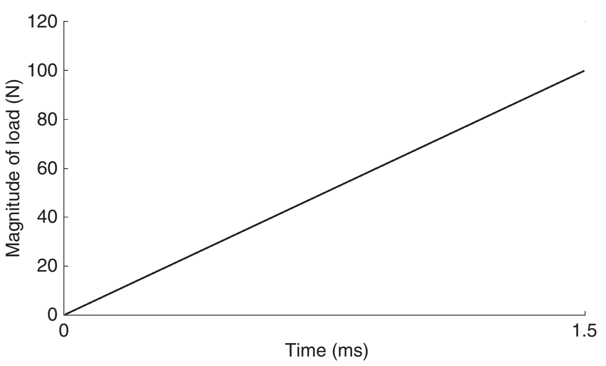

Fig. 2. Time–force history applied to the central incisor. A ramp load was applied to the surface of the incisor. The maximum load was set to 100 N.

center started to rotate at the edge of the palatal alveolar bone, combining the incremental stress increases from the labial and cervical edges of the bone that occur 0.5–1.0 ms after the impact. When the elongation reached the designated limit, the PDL disconnected, leading to tooth avulsion (1.0–1.5 ms). Just before avulsion, the maximum stress at the labial bone was 200 MPa, while that at the ridge crest area of the palatal bone was 400 MPa at 1.45 ms after the impact (Fig. 4).

Discussion

The viscoelastic behavior of the PDL and the alveolar bone may also affect the mode of avulsion. There have been only a few investigations (2, 3) regarding the modes of and force distributions for external impact on teeth, PDLs and alveolar bones. Nor is much understood about the mechanisms involved in tooth avulsion, even though computer simulation modeling remains an appropriate tool for this area of research. Most previous simulation studies of dental trauma have involved static analyses, whereas the present study models a dynamic analysis. Viscoelastic behavior of the PDL was included in the model to confine the inter-node relationship. The results of our simulation under the conditions of a traumatic injury reveal that viscoelastic material, used for confining the inter-node relationship, is suitable for inducing a mechanical response from the PDL. The structures of teeth and PDLs are very complicated, and we therefore used a simplified FEM for evaluating tooth avulsion. This method enabled us to predict the movements of the tooth at the time of the traumatic event. Regarding the generation mechanism for avulsion, modification through compression of the PDL occurred at the palatal side. When the limitation of the modification was approached, rotational movement occurred while centered on the palatal alveolar bone ridge, and the root of the tooth may have contact with the side wall of the dental socket. PDL disconnection evolves through its initial stage on the apical side far from the fulcrum of tooth movement and then gradually spreads upwards. The tooth starts to move after PDL disconnection, resulting in avulsion. It is well known that the incisor and the labial alveolar bone can easily break following frontal impact, often resulting in tooth avulsion or alveolar bone fracture. From this simulation, stress distributions under impact loading could be observed in real time.

Within 1.5 ms of the impact, the designated threshold was reached, leading to PDL disconnection and the rapid rotations of the tooth. After this disconnection of the PDL, the von Mises stress in the buccal alveolar bone decreased, whereas that in the palatal bone increased.

Although the von Mises stress in the palatal bone was higher than that in the buccal bone, the buccal alveolar bone was previously reported to show a statistically higher frequency of fracture (1). This cause is thought to originate from the thickness of the bone. Furthermore, as the von Mises stress in the buccal bone increased before the avulsion (1.0–1.5 ms), the risk of buccal bone fracture is considered to have increased at this stage. The results of our analysis lead us to conclude that the fracture of the alveolar bone will occur easily before PDL disconnection. On the other hand, tooth avulsion occurs easily after PDL disconnection.

The impact threshold by which bone fracture becomes tooth avulsion is therefore considered to be dependent on the magnitude and duration of loading.

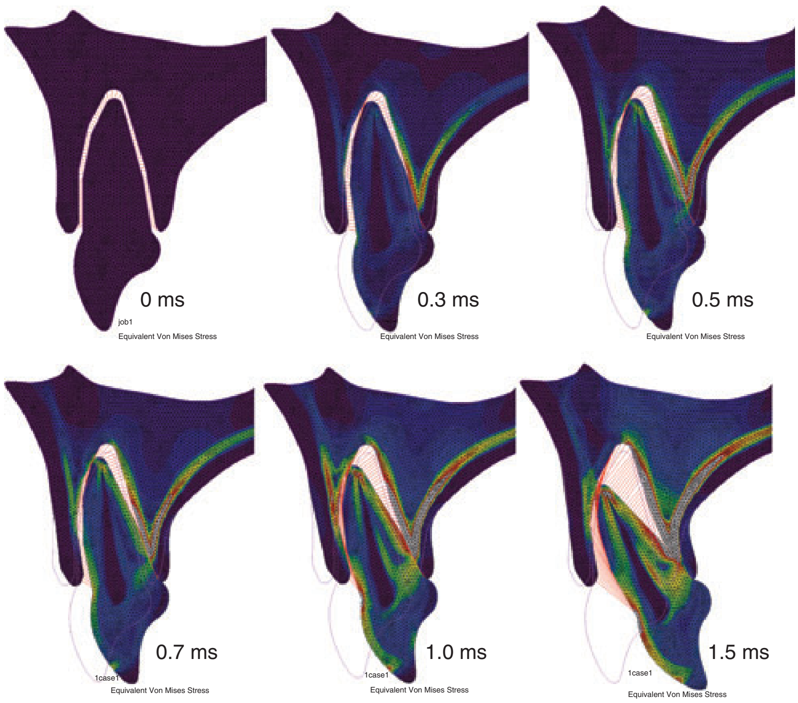

Fig. 3. Results of the tooth avulsion analysis. The model was subjected to a dynamic load with a magnitude of 100 N within 1.5 ms. A distribution of von Mises stress develops in the maxillary central incisor and alveolar bone.

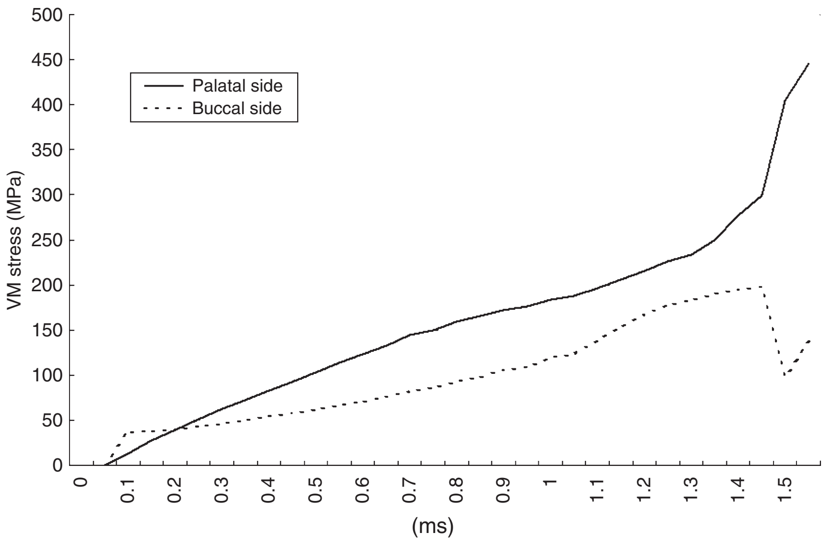

Fig. 4. Von Mises stress values in the lingual and palatal compact bone. At 0–0.3 ms, the tooth starts to move and deformation of the PDL begins. At 0.3–0.7 ms, the tooth rotates and the PDL becomes deformed. At 0.7–1.5 ms, the PDL starts to disconnect. At 1.5 ms and beyond, the tooth is completely dislocated.

Conclusions

The two-dimensional FEM described is capable of modeling the displacement and involved forces as well as dynamic avulsion events trauma. Moreover, the PDL of the maxillary central incisors provides protective mechanisms for teeth following trauma. The method of limiting parameters in the inter-node relationship to help describe PDL mobility can further improve the dynamic analysis of tooth avulsion.

References

1. Andreasen JO. Traumatic injuries of the teeth, 2nd edn. Copenhagen: Munksgaard; 1981. p. 34–55.
2. Huang HM, Tsai CY, Lee HF, Lin CT, Yao WC, Chiu WT et al. Damping effects on the response of maxillary incisor subjected to a traumatic impact force: a nonlinear finite element analysis. J Dent 2006;34:261–8.
3. Huang HM, Ou KL, Wang WN, Chiu WT, Lin CT, Lee SY. Dynamic finite element analysis of the human maxillary incisor under impact loading in various directions. J Endod 2005;31:737–7.
4. Toms SR, Dakin GJ, Lemons JE, Eberhardt AW. Quasi-linear viscoelastic behavior of the human periodontal ligament. J Biomech 2002;35:1411–5.
5. Pini M, Zysset Ph, Botsis J, Contro R. Tensile and compressive behavior of the bovine periodontal ligament. J Biomech 2004;37:111–9.
6. Cuy JL, Mann AB, Livi KJ, Teaford MF, Weihs TP. Nanoindentation mapping of the mechanical properties of human molar tooth enamel. Arch Oral Biol 2002;47:281–91.
7. Kinney JH, Marshall SJ, Marshall GW. The mechanical properties of human dentin: a critical review and re-evaluation of the dental literature. Crit Rev Oral Biol Med 2003;14:13–29.

8. Lerry S, Seedhom BB, Berry E, Cuppone M. Quality assessment of the cortical bone of the human mandible. Bone 2003;32:35–44.
9. Giesen EB, Ding M, Dalstra M, van Eiden TM. Mechanical properties of cancellous bone in the human mandible condyle are anisotropic. J Biomech 2001;34:799–803.
10. Provatidis CG. A comparative FEM-study of tooth mobility using isotropic and anisotropic models of the periodontal ligament. Med Eng Phys 2000;22:359–70.