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Damping effects on the response of maxillary incisor subjected to a traumatic impact force: A nonlinear finite element analysis

Haw-Ming Huang[a,b], Chi-Yang Tsai[c], Hao-Fu Lee[c], Che-Tong Lin[a], Wei-Cheng Yao[b,d], Wen-Ta Chiu[e,f], Sheng-Yang Lee[c,*]

[a]*Graduate Institute of Oral Sciences, Taipei Medical University, Taipei, Taiwan, ROC*
[b]*Graduate Institute of Medical Sciences, Taipei Medical University, Taipei, Taiwan, ROC*
[c]*Institute of Oral Rehabilitation Sciences, School of Dentistry, Taipei Medical University, 250 Wu-Hsing Street, Taipei 110, Taiwan, ROC*
[d]*Department of Anesthesiology, Min-Sheng General Hospital, Taoyuan, Taiwan, ROC*
[e]*Department of Neurosurgery, Municipal Wan Fang Hospital, Taipei, Taiwan, ROC*
[f]*Institute of Injury Prevention & Control, Taipei Medical University, Taipei, Taiwan, ROC*

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Summary *Objectives*: The aim of this study was to evaluate the effects of damping on stress concentration in an impacted incisor.
Methods: Damping ratios of maxillary incisors were tested using an in vivo modal testing method. A finite element model of the upper central incisor was established for dental trauma analysis. To assess the effect of damping properties on induced stresses in the traumatized incisors, equivalent stresses in the finite element model with various damping ratios were calculated for comparison. The mechanisms of cushioning properties of the upper incisors on traumatic injuries were assessed by profiling the stress distributions in the incisor model sequentially with time.
Results: The measured damping ratio of maxillary incisors was 0.146 ± 0.037. When the incisor was subjected to an impact force, high stresses were concentrated at the labial and lingual incisor edges, cervical ridge, and the area around root apex. When the damping ratios of the incisor model were set at 10- and 50-fold of the measured values, the peak stresses induced near the impact site of the incisor model were reduced from 24.0 to 23.2 and 15.9 MPa, respectively. On the other hand, the peak stress lagged and the stress existence period increased when the damping properties were taken into consideration.

* Corresponding author. Tel.: +886 2 2736 1661x5123; fax: +886 2 2736 2295/2778 5383.
E-mail address: seanlee@tmu.edu.tw (S.-Y. Lee).

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Conclusions: Damping properties of teeth provide protection to the tooth during traumatic injury by decreasing the peak stress magnitude due to release of strain energy over a longer period.

Introduction

Damage to incisor teeth is a common injury due to accidents.[1] Although various biomechanical studies have focused on dental trauma, the mechanisms of such tooth injury are still not fully clear.[2,3] In fact, the experimental models of most biomechanical studies are simplified to a static force applied to a tooth with elastic material properties.[4-6] In such static analyses, the applied force was assumed to be constant during the impact period. However, in real situations, traumatic injuries to teeth typically result from dynamic forces. The magnitude of such dynamic forces alters with time. In biomechanics, when time is taken into consideration, the effects of damping properties of the impacted tooth cannot be ignored.

Many researches have found that the periodontal ligament (PDL), which connects the tooth root and the underlying bone, plays an important role in the mechanisms of tooth trauma and tooth mobility.[7,8] Recently, to understand the mobility of human dentition under the action of physiological short-term loading, nonlinear finite element (FE) analyses were performed by several authors.[9-11] After determining the damping properties of the PDL by means of a reversed calculation method, these nonlinear analyses could be used to overcome the inaccuracies of approximation due to the use of simplified models.

It is well known that damping materials in teeth act as shock absorbers to minimize external impact. Although the damping properties of PDL are the main contributors to the viscoelastic property of a tooth, the cushioning effect of other damping materials in a tooth, such as pulp, can affect the stress distribution in an impacted tooth. Therefore, any simulation that ignores the damping effect of a tooth will cause unexpected error.

To understand the viscoelastic properties of teeth, vibration analyses were performed by many scholars.[12-14] In these studies, after simplifying a natural tooth to a mechanical model, damping properties of the tooth were determined mathematically by curve-fitting the experimental data to the vibrational behavior of the mechanical model. However, such a reversed calculation is an indirect method of damping property determination. The results are somewhat selective because they are dependent on the mathematical or mechanical models the authors established. To directly measure damping properties of a tooth in vivo is a challenge for investigators.

In 1993, Dimarogonas et al. performed a modal testing experiment on a human tibia to determine its damping effect. After triggering the test tibia into vibration, the damping ratio of the test tibia was obtained directly by monitoring the decay rate of its vibration.[15] Their results showed that such a nondestructive method had advantages because it not only directly and accurately detected damping properties of a tissue, but was also easy to set up and use.

In this study, modal testing experiments were used to measure the damping ratios of maxillary central incisors. The effect of damping on stress distribution in the incisor subjected to a traumatic load was then evaluated using a series of dynamic FE analyses.

Materials and methods

For a vibrational object with viscous damping, the damping force F is proportional to the velocity v and can be expressed as

$$F = -Cv \tag{1}$$

where C is the damping constant. Mechanically, the damping corresponding to the limiting case between oscillatory and nonoscillary motion is called critical damping, C_c. For any damped system, damping can be expressed in terms of the critical damping using a nondimensional number ξ. This is called the damping ratio, and is defined as the ratio of the damping constant to the critical damping constant:

$$\xi = C/C_c \tag{2}$$

Damping ratio is an important parameter for developing an FE model with viscous property. However, due to the lack of quantitative scientific data, the damping ratio of the tooth was determined in this study by means of a series of in vivo modal testing experiments.

As shown in Fig. 1, when a maxillary central incisor is excited to vibration, its vibration amplitude decays along the vibration time line according

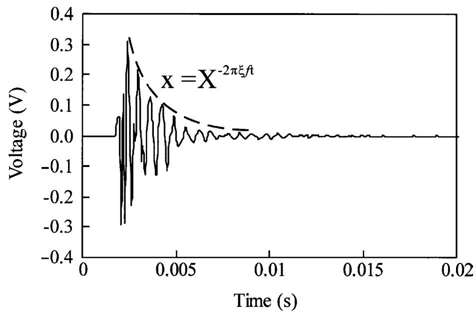

Figure 1 Vibrational response of the test maxillary central incisor. The formula describes the rate of decay of amplitude.

to the following relationship[16]

$$x = X\,e^{-2\xi\pi f t} \tag{3}$$

where X is the maximum vibration amplitude, t is the vibration time, f is the resonance frequency, ξ is the damping ratio. Therefore, the damping ratio of a tooth can be obtained by detecting the amplitude decay trend of a vibrating tooth subjected to a sudden nonperiodic excitation.

In vivo testing

Modal testing experiments were performed on 15 volunteer graduate students in the Dental Department of the Taipei Medical University, Taipei, Taiwan. The volunteers consisted of eight men and seven women aged between 20 and 22 years. To ensure the volunteers' periodontal health status, all the test teeth were examined by means of a probing depth examination (probing depth $\leq$3 mm) and X-ray imaging prior to the volunteers undergoing modal testing. A total of 30 maxillary central incisors were tested in vivo. Before undergoing testing, all the teeth were first dried using cotton. Fig. 2 demonstrates the apparatus and testing method used for this experiment. The test incisors were forced into vibration by the application of an impulse force hammer (Model GK291C80, PCB Piezotronics, Buffalo, NY, USA). The exciting force was directly applied on the surface of the test incisors in the lingual-labial direction. In addition, an acoustic microphone (FM-10B, FC Electronics, Taipei, Taiwan) was used as the transducer. The microphone was positioned 1 cm vertically above the striking surface to collect the vibrational signal. The vibration signal was then transferred to a frequency spectrum analyzer (Probel II, Prowave Engineering, Hsinchu, Taiwan) for resonance frequency and damping ratio display. Results for each test tooth were obtained by averaging the results from five triggering responses. For each tooth, the testing time was controlled to within 2 min.

Finite element analysis

To examine the effect of damping ratio on stress distribution in traumatic impact of an incisor, dynamic FE analysis was performed in this study. The FE analysis package, ANSYS® (Swanson Analysis Systems, Houston, PA), was used to perform the transient dynamic analysis on a personal computer. A 2D plane strain FE model of the human maxillary central incisor, containing enamel, dentine, pulp, periodontal membrane, alveolar bone, compact bone, and spongy bone was built (Fig. 3). The geometry and dimensions of the incisor, including the 23.5 mm long tooth and the 0.25 mm thickness of the periodontal membrane was selected as in a previous study.[17] The alveolar process was located 2 mm apically from the cementoenamel junction (CEJ). The material properties of the tooth (Table 1) were obtained from the literature.[17] The dynamic properties of the model, including damping ratio and resonance frequency, were input according to the in vivo tests in this study. The model had a total of 5310 2D quadrilateral elements.

As shown in Fig. 4, a sinusoidal force with a peak of 800 N at 2 ms,[3,18] and a total duration of 4 ms[19,20] was applied to the model in a direction of 45° labial

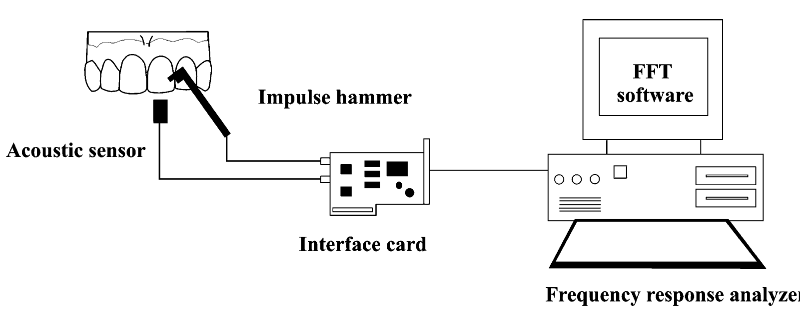

Figure 2 Instrumentation and testing methods used for this study.

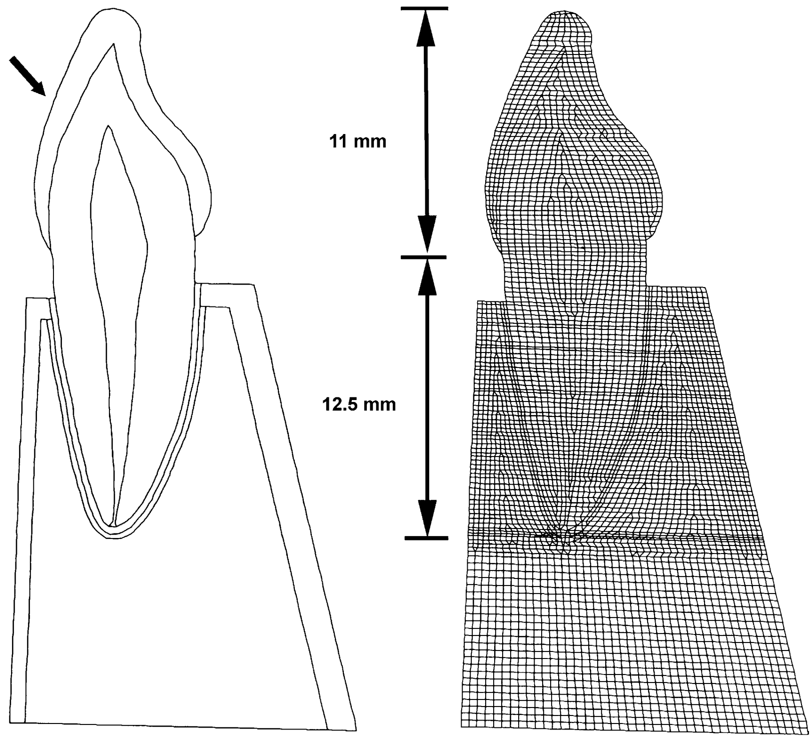

Figure 3 Two-dimensional solid model (a) and plane strain finite element model (b) of the human maxillary central incisor used in this study. Arrow demonstrates the position and direction of impact force applied.

to the incisal edge. Equivalent (EQV) stress contours within the FE models were calculated and were displayed for comparison. To assess the damping effects on stress concentrations, stresses at the labial incisal edge at a distance of 0.7 mm off the impact site were calculated when the damping ratio of the model was altered to 0.1-, 1-, 10- and 50-fold of the measured value. Furthermore, to compare the difference in stress concentration when the model was subjected to static and dynamic loads, a static force with a magnitude of 800 N was applied to the model. The applied position and direction of the static load were the same as that set in the dynamic analysis.

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
Cancellus bone	0.50	1.40	0.38

Results

The resonance frequencies of the maxillary central incisors tested in vivo ranged from 1167 to 1630 Hz, with an average of 1388 $\pm$ 148 Hz. The damping ratios of the tested teeth ranged from 0.091 to 0.240 with an average of 0.146 $\pm$ 0.037.

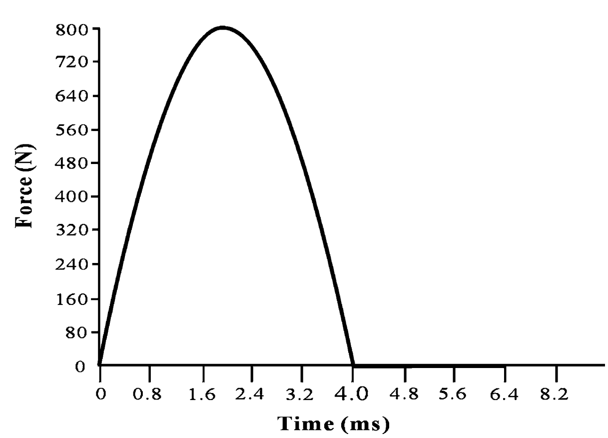

Figure 4 Force-time history applied to the incisor.

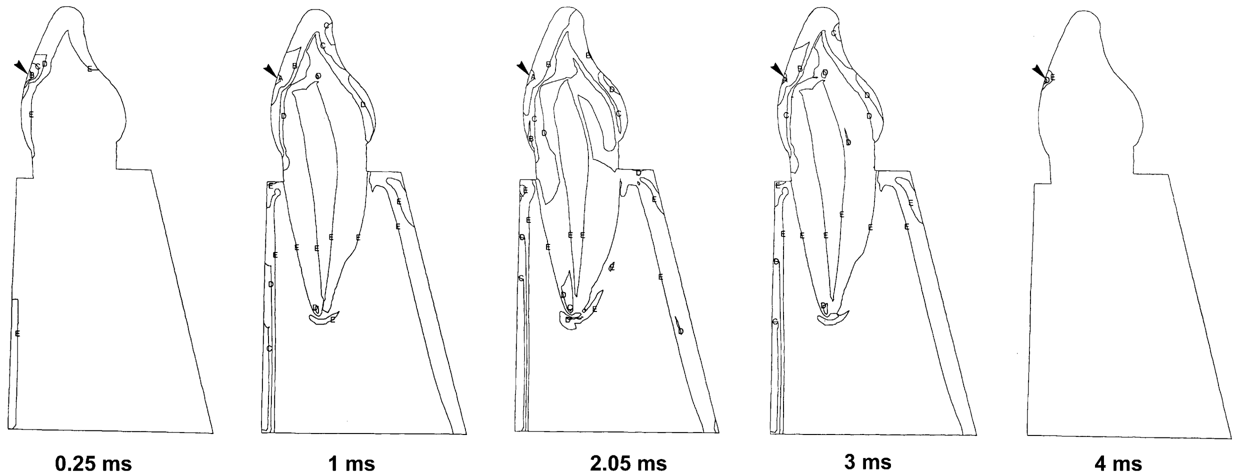

Figure 5 Equivalent stress distribution developed in the maxillary central incisor when the tooth was subjected to a dynamic force. (A) 50 MPa; (B) 10 MPa; (C) 5 MPa; (D) 3 MPa; (E) 1 MPa.

In the FE analysis, the sequential stress contours of the upper incisor, after it was subjected to an impact force, are demonstrated in Fig. 5. The figure, shows how the concentrated stress points, at various locations inside the tooth, alter with time. High stresses were concentrated at the labial and lingual incisal edges, cervical ridge, and the area around the root apex (Fig. 5). Maximum stress with a magnitude of 10 MPa first developed at the impact site at 0.25 ms then exceeded 50 MPa at 1 ms. At the same time, a stress over 5 MPa also occurred around the lingual incisal edge. At 2.05 ms, except for the above site, high stress concentrations were also found at the labial root apex, with a peak value higher than 5 MPa. After this time point, the stresses in the model decreased. At 3 ms, only the stress around the labial incisal edge remained at a high value exceeding 10 MPa. After 4 ms, no significant stress concentration could be found in the model, except for in the area around the impact site. When the model was subjected to a static force, the stress contour in the model (Fig. 6) was similar to the pattern of dynamic analysis at 2.05 ms.

Fig. 7 shows the equivalent stress at the labial incisal edge at a distance of 0.7 mm off the impact site that developed after the FE model was subjected to an impact force. The higher the damping ratio was in the model, the lower the peak stress conducted in the model. The maximum stress conducted in the models with damping ratios of 0.1-, 1-, 10- and 50-fold measured value are 24.6, 24.0, 23.2 and 15.9 MPa, respectively. These values are significantly lower than the corresponding results obtained from static analyses (26.7 MPa). Furthermore, results in Fig. 7 show that the damping properties, set in the model, influence the time point of maximum stress occurrence. Only when the damping ratio was set to 0.1-fold of the measured value, the maximum stress of the model occur at the same time that the input force has maximum value. While the damping ratio of the tooth increased, the time point of maximum stress conducted at the incisor edge lagged. The calculated maximum stresses occurred at 2.05, 2.36 and 3.01 ms when the damping ratio set in the model was varied to 1-, 10- and 50-fold of the measured data.

Discussion

An accelerometer is the most common transducer used for detecting a tooth's vibrational signals in modal testing experiments.[8,10] However, due to the relatively small size of human teeth in comparison to the accelerometer, using an accelerometer as a detector in this study might have caused unexpected errors. To eliminate the unnecessary mass effect of the accelerometer, an acoustic microphone has previously been used as a transducer to measure the resonance frequency of the tibia and natural teeth.[21-23] These investigations have suggested that using a microphone as a sensor has advantages when examining the vibrational behavior of hard tissues of living creatures. In our study, a noncontact acoustic microphone was used to collect the vibrational responses of tested incisors. Our results indicated that a microphone can also

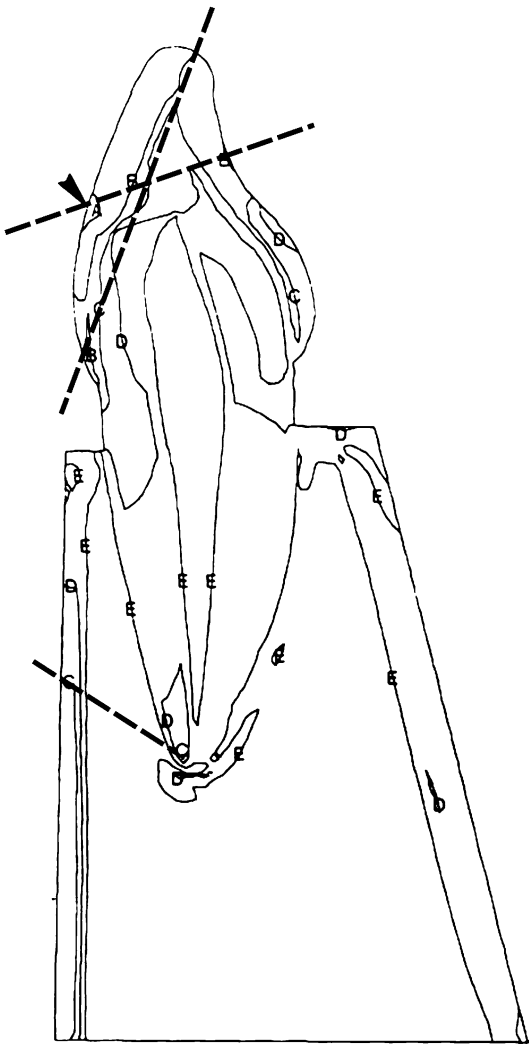

Figure 6 Equivalent stress distribution developed in the maxillary central incisor when the tooth was subjected to a static force. The dashed lines denote the possible fracture lines of the tooth. (A) 50 MPa; (B) 10 MPa; (C) 5 MPa; (D) 3 MPa; (E) 1 MPa.

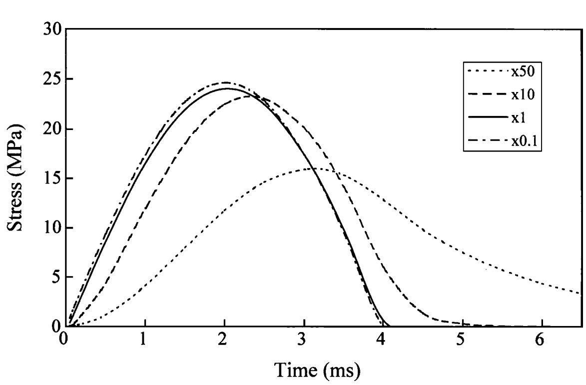

Figure 7 Equivalent stress at 0.7 mm off the impact site, calculated using the finite element model with various damping ratios.

serve as an effective tool for the testing of dynamic parameters of human incisors due to the absence of contact with the tooth being tested and, if selected appropriately, its highly-sensitive characteristics.

The vibration signals captured in our in vivo modal testing experiments came from the entire test incisors. All damping material in the incisor, including periodontium and pulp, contribute to the measured damping ratio. Therefore, the measured data is the damping ratio of the intact upper incisor including pulp and surrounding soft tissue. A vibrating tooth may encounter many different types of damping force, from internal molecular friction, sliding friction, and fluid resistance. Mathematical descriptions of teeth are complicated. Thus, simplified damping models are adequate in evaluating the system response.[24] To perform a dynamic FE simulation of head injury, Chu et al. (1994) simplified the damping properties of various tissues in their head FE model to a single value. Despite the simplification, the computed intracranial pressure and stress matched the experimental measurements reasonably.[19] In this study, the tooth's movement was not discussed in our FE analysis and so we simplified the FE model to a composite material with a single value of damping ratio.

The nondimensional number, the damping ratio (ξ), was used to express the degree of damping property of the tested incisors in this study. In formula (2), when a system with a ξ value equals zero (damping constant, $C=0$), it means the system is an undamped system. In contrast, as ξ exceeds unity, the system is under overdamping condition, the motion of the system is an exponentially decreasing function of time. For a structure under damping vibration, its damping ratio should be between 1 and 0. In this study, the measured damping ratio of the test incisors was 0.146 ± 0.037. This is much higher than the value of metal (with a damping ratio of 0.01)[25] because, except for enamel and dentine, a tooth contains various damped soft tissues and tissue fluid in it. In comparison, when Chu et al. (1994) reverse calculated the damping ratio of the human head, it was about 0.40.[17] This is because a head contains much more soft tissue and damped tissue fluid than does an incisor.

Material damping is defined as the fraction of strain energy lost in one full cycle of deformation.[13] Therefore, in Fig. 7, the larger that the damping ratio of the upper incisor model is, the lower the peak stress induced. On the other hand, the peak stress lagged and the stress existence period increased when the damping properties were taken into consideration. That is, an impacted incisor with a higher damping property can lower the concentrated stresses by dispersing the strain energy over a longer period. Such a cushioning effect is the main principle that allows a mouth guard to be used for protecting a tooth from severe impact injury.[26]

Moaveni (1999) suggested that experimental testing may be the best way to verify an FE model.[27] According to the descriptions of Andreasen and Andreasen (1994), the fracture lines in canines caused by frontal impacts fall into four categories: horizontal crown fractures, horizontal fractures at the neck of the tooth, oblique crown-root fractures, and oblique root fractures.[1] In Fig. 5, we also found that high stresses concentrated at some specific areas, which obeyed engineering principles. These areas are the labial and lingual incisor edges, cervical ridge and the area around the root apex. When connecting lines between these high stress areas, fracture lines similar to those described by Andreasen and Andreasen's were found.[1] The finding of similar qualitative results from clinical observation and the numerical simulation demonstrates that the FE model used in this study is a reliable model for dental trauma analysis. However, when tracing the stress growth process (Fig. 5), we found that high stresses in each concentrated area occurred independently. These phenomena explain why the fracture lines caused by impacts reveal specific directions and positions. After comparing the results for Figs. 5 and 6, we found that although the static analysis demonstrated that the stress contours were similar to the patterns of dynamic analysis at the transient time when maximum stress occurs, the stress values calculated in the static analysis are overestimated. This is because in static FE analysis, the model is always set to be elastic.

In conclusion, the damping tissues of the maxillary central incisors provide protective effects for teeth during traumatic injury by dispersing the concentrated stress over a longer period. The damping ratio of human maxillary central incisors measured in this study can be a useful reference for further related dynamic analysis.

Acknowledgements

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