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ORIGINAL ARTICLE

Vibrational analysis of mandible trauma: experimental and numerical approaches

Shu-Li Lin · Sheng-Yang Lee · Long-Yi Lee · Wen-Ta Chiu · Che-Tong Lin · Haw-Ming Huang

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Abstract The aim of this study was to evaluate the effectiveness of vibrational assessment of the mandible fracture patterns. Measurement of natural frequencies and associated vibrational mode shapes was performed to determine the relationship between the dynamic behavior of the human mandible and incidence of mandibular fractures using both in vitro modal testing and finite element analysis. Our results show that the natural frequencies of the human mandible in dry and wet conditions are 567 Hz and 501 Hz, respectively. The first vibrational mode of human mandible is a bending vibration with nodes located at the mandibular body where bone fracture is less likely to occur. By contrast, high vibration amplitudes were identified in the symphysis/parasymphysis and subcondyle regions where bone fractures tend occur. These findings indicate that the vibrational characteristics of the mandible are potential parameters for assessment of the mechanisms of injury.

Keywords Natural frequency · Mandible · Modal testing · Finite element analysis · Trauma

S.-L. Lin
Dental Department, Cathay General Hospital,
Taipei, Taiwan

S.-Y. Lee · L.-Y. Lee · C.-T. Lin
School of Dentistry, Taipei Medical University,
Taipei, Taiwan

S.-Y. Lee
Dental Department of Wan-Fang Hospital,
Taipei Medical University, Taipei, Taiwan

W.-T. Chiu
Department of Neurosurgery,
Municipal Wan Fang Hospital, Taipei, Taiwan

H.-M. Huang (✉)
Graduate Institute of Oral Sciences,
Taipei Medical University, No. 250, Wu-Hsing Street,
Taipei, Taiwan
e-mail: hhm@tmu.edu.tw

1 Introduction

Oralfacial trauma is the most frequently seen injury in industrial countries. This injury can be caused by vehicle accident, fall, sporting injury or assault. Clinical researches have revealed that mandible fractures account for over half of Oralfacial injuries [22, 26]. Almost two-thirds of mandible-fracture patients sustained multiple fractures of this bone [8, 19]. Overall, Symphysis/parasymphysis area and condyle are the two most common fracture sites within the mandible. By contrast, the frequency of mandible body fractures is relatively low [19, 32]. For many years, it was simply considered that the locations of mandible fracture lines were associated with the geometrical and anatomical characteristics of mandible.

It is generally concluded in traumatic injuries that the peak-applied load is not sufficient to predict bony fracture [18]. In the realm of traumatic biomechanical analysis, an important bone-tissue property is its resonant responses during an accident. This is because traumatic injuries typically result from impact with a hard object. Such impact forces are, in general, of short duration and typically give rise to a vibrational response, superimposed on a rigid body motion of the impacted tissue [18]. A number of investigators have performed vibrational experiments using hard human tissue involving, for example the natural frequencies and mode shapes of the femur [7, 18], tibia [11, 36], ulna [15], lumbar vertebra [16, 20], head [17, 35], and teeth [14, 23]. All of these have already been analyzed systematically.

The method used for the estimation of an object's natural frequency is non-invasive and non-destructive. Recently, several investigators employed resonant frequency as a parameter for rigidity assessment of maxillary and mandibular implants [3, 5, 12, 13] and mandibular, major removable partial-denture connectors [2]. Relevant published research pertaining to the human mandible in this regard is, however, obviously lacking. Given the fact that the exact mechanisms of mandibular fracture are still unclear, investigation of the structure's dynamic characteristics such as natural frequency and vibrational mode shape appears to be especially important and imperative for development of reliable predicators of injury patterns.

The internal responses of traumatized hard tissue are difficult to measure directly by experimental instruments due to problems associated with the covering of soft tissues [31]. Therefore, numerical approaches, such as the finite element method (FEM), are widely used to calculate dynamic responses in the hard tissues after traumatic impacts. For vibration behavior assessment, FEM has been used in orthopedics [11, 24, 38] and in the study of lumbar spine [16, 30] and head injury [21, 41], to enhance our understanding of the detailed mechanical response of human hard tissue under dynamic loads.

The development of a predicator of injury patterns will be a useful guide to accurate diagnosis and management of mandible fractures. According to the above-cited literature, the natural frequencies of hard tissue not only provide a meaningful clinical index as to its material properties, but also, these vibrational characteristics constitute an important parameter in traumatic-injury analysis. Therefore, to clarify the mechanisms of fracture patterns in traumatic mandible injuries in this study, a 3-D finite element (FE) model was used to calculate the natural frequencies and mode shapes of the human mandible. The model was first validated against a series of modal testing experiments. The relationships between mandibular fracture patterns and related vibrational behavior are subsequently discussed using the calculating results of the model application.

2 Materials and methods

2.1 In vitro modal testing

Resonant and natural frequencies represent the vibrational responses of a structure when damping properties are and are not taken into account, respectively. As the mass of soft tissue, with its large damping ratio, obviously lowers the vibration response of test bone [18], dry human cadaver mandibles were used in our in vitro study. Ten cadaver mandibles (labeled A–J) were obtained and tested at the Oral and Maxillofacial Physiology Laboratory, Taipei Medical University, Taipei, Taiwan. The apparatus and test method used for this experiment are depicted in Fig. 1. The tested mandibles were supported on sponge to approximate a free-support condition [18, 29], and then forced into vibration by an impulse force hammer (Model GK291C80; PCB Piezotronics, Buffalo, NY, USA). Applied by hand, the exciting force was directly applied to the surface along the lingual-labial or lingual-buccal direction (frontal or posterior position, respectively). Along the length of the mandible, seven anatomic locations (symphysis, parasymphysis, angle, ramus, subcondyle, and the body at premolar and molar areas) were selected for excitation (Fig. 2).

The response signal obtained as displacement were acquired through piezoelectric accelerometers (Model 352B22, PCB Piezotronics, Inc., NY, USA) connected to an analyzer through a multi-channel adapter. The

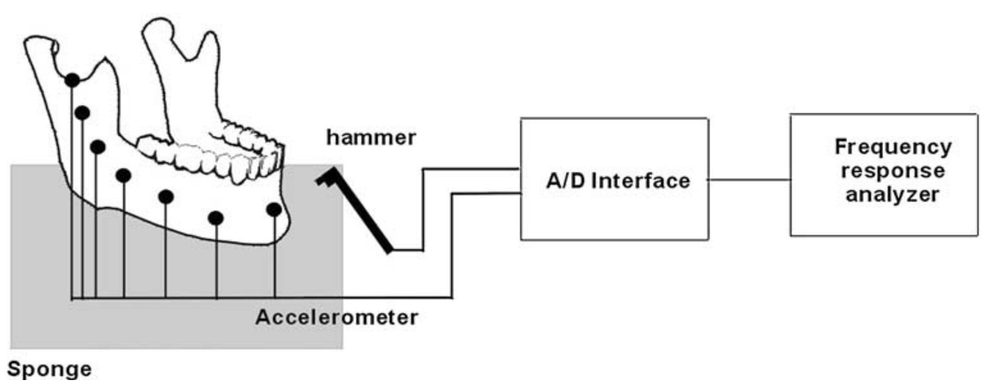

Fig. 1 Schematic diagram of the modal testing experiment

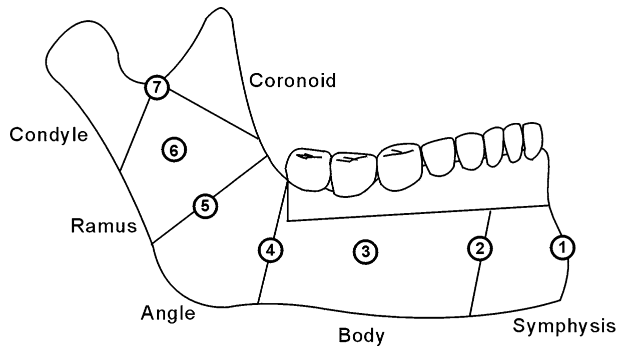

Fig. 2 Anatomic location of the seven testing positions (*closed circles*)

accelerometers were attached at all the selected anatomic locations as illustrated in Fig. 1. The force and acceleration signals were output to a personal computer through a dynamic signal analyzer interface card (AD102A; Prowave Engineering, Inc., Hsinchu, Taiwan) (Fig. 3a, b). Fast Fourier Transform software was used to convert the vibrational signal from time-domain (Fig. 3b) to frequency-domain formats (Fig. 3c). The natural frequency of the test mandible was assessed through evaluating of peak mobility amplitude as reflected by the frequency response spectrum. The impacts were repeated at all the seven locations. Impact was done up beside the accelerometer when it was set to strike on the same anatomical location as the accelerometer. One finial signal was obtained by averaging the results from every 3 independent triggerings. The testing for each situation was repeated five times, and the mean value and standard deviation (SD) were calculated for later analysis.

To obtain the vibrational mode shape of the test mandible, a nondimensional frequency response function (FRF) was defined as the ratio between the peak amplitudes of the vibrational response in frequency domain (Fig. 3c) and input force in time domain (Fig. 3a). The mode shape of a test mandible was determined by connecting the detected FRF for every location, as described earlier [11]. Signal detections were performed only on one side of the test mandibles because of the structure's symmetry. It has been demonstrated that the first natural frequency of a hard-tissue union is effective for tracking clinical problems [1, 23], hence, only these frequencies and their associated mode shapes were tested in this study.

2.2 Finite element modeling

In this study, a 3-D finite element model of the human mandible, containing compact bone, spongy bone and

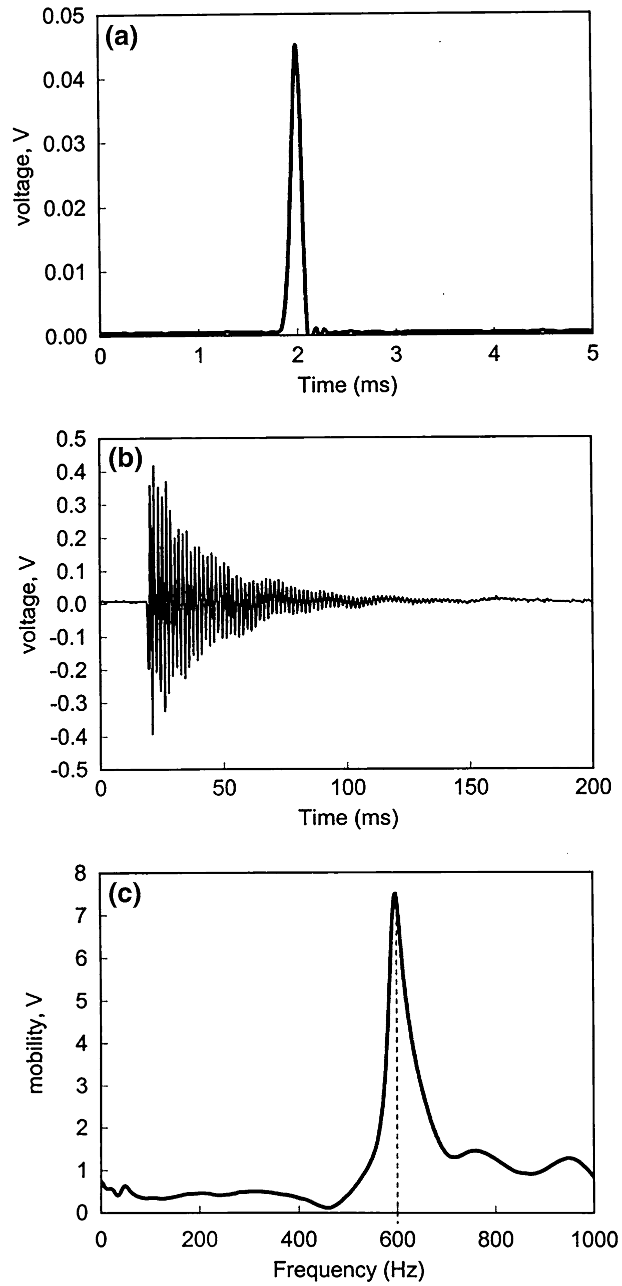

Fig. 3 Typical spectra of a modal test, **a** impulse force at location 5 and **b** the vibrational response of the test mandible received at location 7. **c** The natural frequency is 598 Hz which can be obtained through the frequency response spectrum with the highest amplitude of vibration (*dashed line*)

intact teeth, was built using a CT image-reconstruction technique [40]. The CT scans were performed on a single dry human mandible at a resolution of 5 mm along the mandibular angle-mental tubercle axis. Basic cephalometrics were acquired from the transverse CT images to ensure that the geometries and dimensions of the mandible model fell into the normal range [27]. Bone CT-section coordinates were digitized using image processing software (Image-pro Plus, Media Cybenetics Inc., Silver Spring, MD). Mandibular contour lines from each CT image were replotted at the same intervals and a three-dimensional solid model reconstructed and meshed using general-purpose finite element software package (ANSYS, Swanson Analysis System, Houston, PA, USA). The model consisted of a total of 2,724 nodes and 2,658 3-D solid elements, including a majority of eight-node hexahedral elements and a small percentage of four-node pyramids and five-node tetrahedral elements (Fig. 4). The material properties of the model were considered homogeneous, isotropic and linearly elastic. The specific values of the material properties are listed in Tables 1, [9, 10 and 34]. The simulations were performed with both dry and wet conditions assumed for the mandibular bone tissue. Based on previous works, it was assumed that the elastic modulus of the dry bone was 1.2-fold that of the wet analog [33]. The first natural frequency and associated mode shape of the model were derived using modal analysis under a free boundary condition.

It has been demonstrated that experimental testing may be the best way to verify an FE model [25, 40]. In the present study, therefore, the in vitro modal testing experiments described above were performed for validation of the FE analysis. A difference between the numerical and experimental frequencies of less than 5% was set as the indicator for a credible FE model judgment.

3 Results

In this study of the human mandible, both in vitro modal testing experiments and FE modeling were performed to monitor its dynamic characteristics. The measured natural frequencies for the ten dry cadaver mandibles ranged from 560 to 598 Hz, with a mean of 579 ± 12 Hz (mean ± SD). For each mandible, same natural frequency value was obtained regardless of the location of the excitation and the location of the signal receiving. Connecting of the FRF values allows plotting of the vibrational mode shapes of the mandible. A typical example from mandible A (which has a measured natural frequency of 580 Hz) is illustrated in Fig. 5. It is evident that the minimum and maximum FRF values are always found at locations 3 and 7, respectively, regardless of where the impact force was applied. Meanwhile, when the test mandible was excited by an impact force applied at position 3, the seven accelerometers detected a series of related, smaller FRF values compared to the other excitement conditions. In contrast, the highest responses were recorded by the accelerometers when the impact force was applied at location 7. Obviously, the vibrational mode shape of the structure is a bending mode with a vibrational node proximal to location 3. Figure 6a–c demonstrate the mode shapes of the other ten tested mandibles with impacts at the symphysis, body, and condyle, respectively. Although the vibration amplitudes varied with the impact location, the bending vibration shapes were similar for all ten mandibles.

When a dry mandible was simulated, the calculated natural frequency of the mandible FE model was

Table 1 Mechanical properties of the finite element model

	Young's modulus (GPa)	Density (g/cm^3)	Poisson's ratio
Enamel	133	2.9	0.3
Dentin	18.3	2.2	0.3
Cortical bone	15	1.40	0.3
Cancellus bone	1.37	1.40	0.3

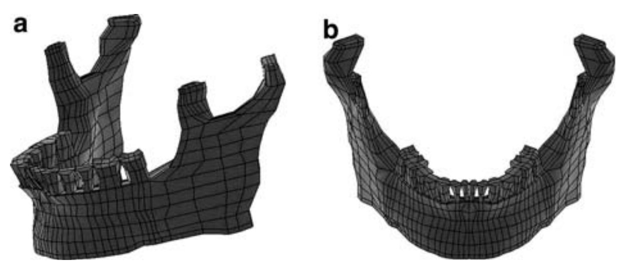

Fig. 4 Three-dimensional finite element model established in this study plotted from a side and **b** sagittal views

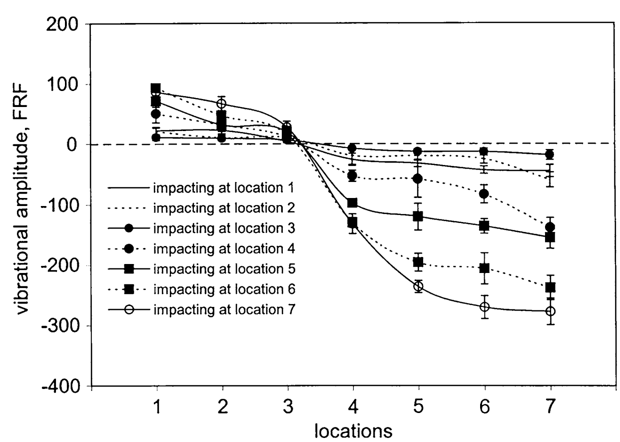

Fig. 5 Typical example of vibrational mode shapes (mandible A) recorded with impacts at various positions. The negative and positive scales on the longitudinal axis represented different vibrational directions. Data are presented as mean ± SD

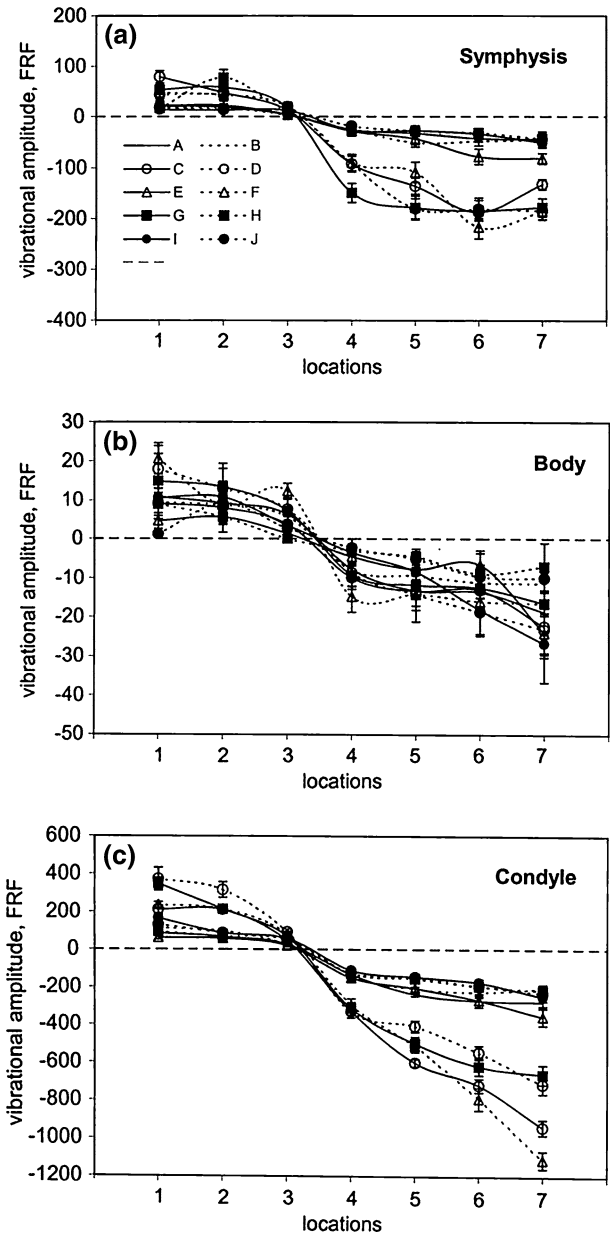

Fig. 6 Vibrational mode shapes of the ten tested mandibles with impacts at symphysis, body, and subcondyle are shown in (**a–c**), respectively. Data are presented as mean ± SD

567 Hz, which is only 1.9% lower than that derived from the in vitro tests. When the model was used to simulate a mandible with a wet bony property, the natural frequency of the model was 501 Hz. The associated mode shape of the vibrated mandible model is depicted in Fig. 7. As in the in vitro modal testing experiments, the mandible is under a bending vibration with a node closed to location 3 (Fig. 8). No obviously difference in mode shape was determined when comparing the results obtained using FE models with dry and wet bone properties, however.

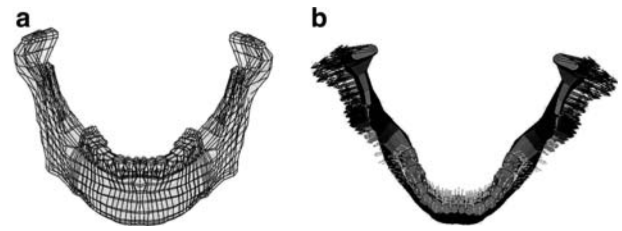

Fig. 7 Vibrational mode shapes calculated using the mandible FE model. **a**, **b** demonstrated the vibrated model and its directions of deformation, respectively

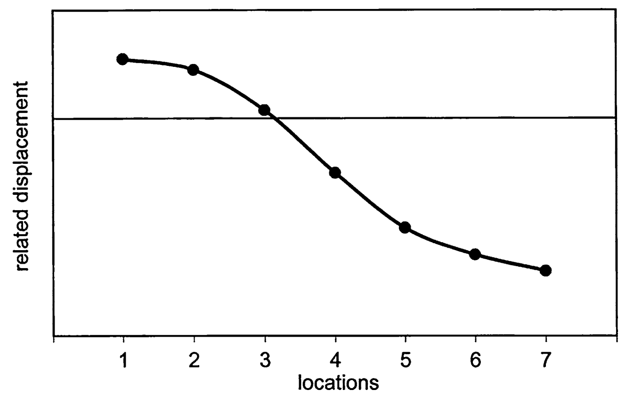

Fig. 8 Vibrational shape of FE model. The mandible is under a bending vibration with a node closed to location 3

4 Discussion

Dynamic responses of a hard tissue could be affected by many anatomical factors, such as shape, size, elastic modulus and density. In Fig. 6, the ten mandibles demonstrated different FRF values and vibrational amplitudes. These variations may be due to the anatomical differences between each sample.

The relentless advancement of computer technology has facilitated considerable attention to be focused on mathematical modeling (e.g., FE analysis) of the physical processes leading to dental trauma in recent years. Importantly, however, there are various sources of error, such as the number of simplifying assumptions made in the construction of the FE model, which can contribute to incorrect results in the numerical analysis. Experimental testing of the mathematical model may be the best way to check its reliability [25]. Vollmer et al. [40] investigated mandibular deformations under mechanical loads, and compared them with results derived from FE analysis. These works concluded that, given the high correlation between the results of the FE analysis and experiment findings, various data within the specimen could be effectively visualized using FE modeling. In the realm of traumatic biomechanical analysis, an important property for a credible model is that it must have a natural frequency response similar to the actual structure being tested. The first natural frequency of the dry mandible calculated using the present FE model was 567 Hz, which is only 1.9% lower than the results from the modal testing experiments. Thus, the results of the FE modeling, in terms of natural frequency, accorded well with our experimental data, validating the present model through vibrational analysis.

When comparing the natural frequencies calculated from FE models, a higher natural frequency is derived for dry bone relative to the wet material, with a ratio of 1.13 demonstrated in this study. This finding is in line with previous modal testing experiments [29], with these works also finding dry tibia bone has a higher natural frequency relative to the wet analog. This is probably due to the larger Young's modulus of the former [33]. From the literature concerning living tissue, the first natural frequency is, mostly, below 1,000 Hz. Reported values for natural human bone frequencies are: approximately 250 Hz for the femur [18], tibia, 260–430 Hz [6, 11, 38], third lumbar vertebra, approximate mean 30 Hz [15], and human head, 300–600 Hz [35]. In this study, the value derived for the natural frequency of the human mandible was 501 Hz, this figure being obviously larger than the corresponding values reported for other hard tissue.

Natural frequency has been widely investigated for the monitoring of fracture healing and osteoporosis, and for evaluation of implant stability in orthopaedics [1, 31]. Clinical studies demonstrated that for a fractured bone the natural frequency is often as low as half that of the intact one [31]. As the facture heals, changes in the resonance frequency indicate an increase in the strength of the bone. However, soft tissue coverage around the long bone restricts the clinical adaptation of this technique. On the contrary, natural frequency analysis is practicable in mandible because, the covered soft tissue at these areas is thinner than other place. However, there are no associated studies of the mandible. In the current investigation, our results may serve as an important reference for future mandible studies of fracture healing and bone-density problems.

The causes of mandible fracture can be divided into direct and indirect violence. Direct violence to one area may produce an indirect force in another part of the mandible. This concomitant indirect violence may be sufficient to cause a second or third fracture [4]. It has been suggested that the resultant indirect violence is closely related to mandible shape, force direction, and impact location. However, traumatic injuries typically result from an impact force of short duration, and this gives rise to the vibrational response.

Viano et al. [39] recorded the fracture force in cadaver mandibles, and measured duration of an impact of 1–2 ms. The natural frequency of human mandible obtained in this study was 501 Hz, which reflects a period of approximately 2 ms. As the waveform of an impact force is nominally half a sinusoidal wave, where the duration is around 1 ms, the impact would trigger vibration in the human mandible.

Cadaver experiments indicated that the fracture tolerance in the mandible body is lower than that in the symphysis area [28, 39]. In their review of 580 mandible fracture cases, however, Olson et al. [32] found that the proportion of body fractures (16.7% of total) was lower than those for symphysis/parasymphysis (19.7%), condylar and subcondylar (31.8%), and angle fractures (21.2%). A previous 3D FE simulation found that dynamic loading of the symphysis region produces direct and indicate violence in the symphyseal area and condylar process, respectively. On the other hand, impact to the symphysis region produced correspondingly lower stress in the mandible body [37]. A high relationship was demonstrated between mandible fracture patterns and vibrational mode shapes comparing the clinical observations and dynamic FE analyses of the above study and those of the present investigation. The coincident results of our numerical and experimental tests confirm that the first vibration mode of the human mandible is a bending vibration with nodes located in the body areas (Figs. 5, 8). Therefore, the mandible body exhibited the lowest vibration deformation regardless of the position of the applied impact force. Further, a direct impact to the body area produced less vibration deformation along the whole mandible compared to impacts at other positions.

Although high relationships were demonstrated between fracture patterns and vibration amplitude in most locations of mandible, the ramus exhibited correspondingly fewer fractures (1.7%) with large vibration deformation [32]. In the real world, a mandible does not exist in isolation. It is covered in, and constrained by, various soft tissues. Although the lack of analysis of soft-tissue effect is a limitation on the present study of mandible vibration, we would suggest that the reduced injury in the ramus area could reasonably be attributed to the stable effects of the medial pterygoid m. and masseter m. These two major muscles not only provide a fixation force, but also act as dampers to reduce the vibration amplitude.

In conclusion, although more-advanced study is needed in the future, our limited results demonstrate that vibration characteristics of the mandible can be a useful parameter for assessing the mechanisms of mandible injury. On the other hand, for assessing the causes of fracture patterns in an impacted mandible, FE vibration analysis provides greater insight into the issue.

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