An experimental modal analysis of clavicle bending modes

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Abstract
Clavicle fractures are common medical emergencies; their prevention through design of protection systems depends upon understanding injury mechanisms. This work analyzes the bending natural frequencies and mode shapes of the human cadaver clavicles in situ. The method applied includes experimental modal analysis (EMA) techniques on cadaver clavicles and correlates results with previous analyses. The clavicle response to shock depends on mechanical energy transmission between load and bone and requires an understanding of modal characteristics of the clavicle as well as the frequency range of the shock. The loads acting upon the clavicle may be represented by hard impacts (i.e. sport-related hits) or loads with short durations which can excite a wide frequency spectrum. Modal analyses of clavicles have been reported in literature, but those studies were performed on the clavicles isolated from the body. As a result, those analyses found mode shapes dependent upon different boundary conditions than those found in nature. In our study, EMA employed triaxial accelerometers and a force hammer, a testing procedure was developed, and data was analyzed. The EMA was performed with the clavicle supported in situ, and results include the coronal and axial plane first bending modes. Modal parameters obtained serve to design shock mitigation systems.

Keywords
Modal analysis, biomechanics, sensors, data, modal parameter estimation

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Introduction
Researchers have investigated the effects of vibrations on humans¹,²–10 which negatively impacted their ability to function in activities ranging from construction work to space travel. Das et al.¹ presents overview of whole-body vibration (WBV), its health effects, relevant remedies for WBV, role and future trend of biomaterials. Fischer et al.¹¹ discusses the potential benefits of WBD in rehabilitation from orthopedic injury through a meta-analysis study. References present development of spring-mass-damper multi-degree of freedom (MDOF) models³–¹⁰,¹² to approximate the behavior of the human body subjected to shock and vibration. These models were developed after operational deflection shape and experimental modal analysis studies. Thus, the human body or one of its subparts was shown to have a set of natural frequencies, mode shapes, and damping factors which could be excited at resonance by the applied vibration or shock loading. Previous work has also shown that internal body cavities have their own resonances.¹³

Impact loading generally provides broadband excitation which, through the Fourier transform, can be decomposed into an infinite sum of harmonic components. Theoretically, a structure also features an

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unlimited number of natural frequencies and corresponding mode shapes. It is possible then that the harmonic excitation components could be equal to one of the structure’s natural frequencies and a resonance condition could occur. Only the first few structural frequencies are usually important for estimating the operational deflection shape (time response) of the structure since their corresponding modal participation factors decrease in magnitude with increasing mode number. It is worthwhile to note that the structural response to shock can be written as a sum of mode shapes multiplied by their respective modal participation factors. Most of the operational deflection shape can be numerically approximated using a limited, as opposite to an infinite, amount of mode shapes.

Operational Deflection Shape (ODS) as opposed to EMA gives actual displacements of the structure under tests. The modal content of a structure can be found using Experimental Modal Analysis (EMA). In EMA a structure is excited mechanically (shaker or force hammer) or acoustically (loudspeaker) and the response (displacement, velocity, or acceleration) structure is recorded. The various type of measurements can be conducted using corresponding devices such as displacement sensors, laser Doppler velocimetry, and accelerometers. Responses and excitations are acquired using set time record length, sampling frequency, and windowing functions. The time waveforms are converted from the time domain to the frequency domain via the Fast Fourier Transform algorithm. Further processing obtains the frequency response functions as ratios of response vs input at discrete frequency points. Natural frequencies are then found using software-based curve-fitting algorithms (i.e. polynomial based and least-square).

The main rationale behind this work is the elevated number of clavicle fractures caused by sport activities or moving vehicles accidents (MVA) demands better protection systems and implicitly a deeper knowledge of clavicle response to transient loading. Clavicle injuries lead to undesirable outcomes such as increased healthcare expenditures, increased difficulty in performing daily activities and inability to work. The focus of the current study is to determine experimentally the first two bending modes of a human cadaver clavicle and correlate the findings with analytical data previously found through finite element analysis (FEA). The logical first step and a future benefit of the work is the development of a procedure for conducting in situ experimental modal analyses for other bones. The procedure will address sensor type and selection, sensor fixation methods, selection of excitation sources matched to the frequency range of interest. The research outcomes may be employed to define new material parameters, such as damping ratios that could augment existing numerical clavicle models. It is hypothesized that the first two bending frequencies of the clavicle lie in a frequency range between 900 and 1500 Hz in line with FEA results. There is an expectation that additional clavicle-body coupled body modes may exist in a lower frequency range. This research expands the preliminary EMA clavicle work reported by the Tsuchikane et al., and has not been reported elsewhere in the surveyed literature databases such as Medline of EI Engineering Village.

**Experimental approach**

**Preparation**

The research was performed in the Anatomy Laboratory of the University of Central Florida College of Medicine, which is supervised by the second author. He received permission to employ a male donor body and has followed the ethical guidelines in performing this work. The donor body had a height of approximately 1.80 m and had a mass of approximately 90 kg.

The testing was performed on the right clavicle of the donor body. The clavicle was exposed by a longitudinal cut through the skin and platysma muscle, which were then reflected. The clavicle was measured to be approximately 14.2 cm in length and is shown in Figure 1. The clavicle was then debrided of any connective tissue to offer a clean surface for solid accelerometer attachment; the bone surface was not treated either chemically or mechanically post debridement. Triaxial accelerometers were then glued to the cleaned clavicle at each measurement location. The researchers noticed that wax attachment of accelerometers to bone was not possible. The wax attachment also may not have met the rigidity needed to obtain reliable response information around the 1000 Hz frequency range. The rigid glue attachment

![Exposed clavicle with triaxial accelerometer.](image)
served to yield better data in the higher frequency range without added artificial damping characteristic of wax.

**Experimental procedure**

The response to excitation was acquired at five (5) measurement locations distributed at equal lengths along the clavicle between the first location, situated 1-cm medial to the acromioclavicular (AMC) joint, and location 5 placed 1-cm laterally from the sternoclavicular (SCJ); as a result, the measurement points were thus spaced approximately 3-cm apart. This measurement spatial resolution was partially limited by the size of the accelerometers themselves, which were cubic in shape and had a 14-mm side length. Since the clavicle does not have regular surfaces, the spatial orientations of the accelerometers were noted in relation to a Cartesian coordinate system. The Cartesian global coordinate system defined for the measurements featured an origin “O” at the point closest to the sternoclavicular joint (point 5), the OZ axis pointed vertically up and defined a posterior to anterior direction as defined by the human body axes). The OX axis lied in a plane parallel to the cadaver table and pointed medially. The fixed hammer impact location was point 5 (Figure 2). The hammer impact was performed in the negative vertical direction Z. Two PCB Piezotronics Y356A16 triaxial accelerometers were chosen to acquire the response data in three mutually perpendicular directions) caused by the excitation. The accelerometers featured a nominal acceleration sensitivity of 100 mV/g (1 g = 9.81 m/s²) per measurement axis. Excitation was provided by a medium-sized, modally tuned PCB Piezotronics 086C01 hammer with a nominal sensitivity of 11.2 mV/N (50 mV/lbf). The hammer featured a steel tip to strongly-excite modes in the 0.9–1.5 kHz frequency range of interest.

A National Instruments CDAQ 9174 with analog input, 12-bit A/D resolution and NI 9234 modules harvested the excitation and response signals. Data was acquired using software triggers using a trigger level of 5% of the 5V acquisition data range. Data record length was 1 s and the acquisition rate was 5120 samples per second. The ensuing frequency resolution was 1 Hz. Acquired data was analyzed using National Instruments™ Modal View™ software. Data visualization was performed in real time to inspect and potentially eliminate double-hammer hits and erratic response signals. Figure 3 displays a data acquisition setup and results.

The practice of EMA dictated that an average of at least five hammer hits were needed per measurement location to obtain high-quality data. Potential source of errors in an impact measurement are double hammering and inadvertent changing of hammer impact direction. A “soft” hit whose spectrum did not have the necessary energy to excite the modes beyond a certain frequency range was also a potential pitfall. Double hammering usually occurs because of improper hitting technique and produces two hammer impulses instead of one within the acquisition window. This results in distortions in the response spectrum. Responses
acquired after a double-hammering event are eliminated upon visual data inspection. A short-duration hammer impulse results in a large excitation frequency range (desirable) but may lead to an increased possibility of double hammering (undesirable). Signals that, upon visual, inspection seemed to be cut off by the ±5V input range of the acquisition card were also eliminated. Acceleration responses were multiplied within the software by an exponential window function\(^\text{13}\) (default was 50% duty cycle and 10% final value for the force/exponential windows) and the hammer signals were multiplied by a force window function.\(^\text{13}\) The testing yielded three axes accelerations at five locations, which yielded fifteen (15) sets of data and upon processing, 15 frequency response functions (FRF). Each data set having one excitation force and three accelerometers digital sets of voltage data. Figure 2 presents several windows which show the hammer impulse signal (top left), accelerometer response middle upper window, data acquisition parameters (upper right window), one sample of the FRF (magnitude and coherence) of the acceleration response versus excitation force in the frequency domain (lower left windows) and a table of frequency response functions (FRF) showing the response point and direction versus excitation point and direction.

The time \((t)\) dependent input (force) and output (acceleration) is \(x (t)\) and \(y (t)\), respectively. The frequency response function is usually denoted by \(H_{XY} (f)\) and is a function of frequency \(f\). It is defined as the ratio between the Fourier transform (denoted by “F” in equation (1)) of the response \(Y (f)\) and excitation \(X (f)\).

\[
H_{XY} (f) = \frac{Y (f)}{X (f)} = \frac{F(y(t))}{F(x(t))} \quad (1)
\]

The cross spectrum of input vs output is \(G_{XY}\) (equation (2)) and the auto-power spectra of input is \(G_{xx}\) (equation (3)) and \(G_{yy}\) (equation (4)), respectively. \(X^*\) and \(Y^*\) are the conjugate quantities of \(X (f)\) and \(Y (f)\), respectively.

\[
G_{XY} (f) = X(f)Y^*(f) \quad (2)
\]

\[
G_{xx} (f) = X(f)X^*(f) \quad (3)
\]

\[
G_{yy} (f) = Y(f)Y^*(f) \quad (4)
\]

The coherence versus frequency plot approaches 1 (one) over the frequency range of interest (Figure 3) which indicated that output data was free of extraneous noise. The coherence function indicates what part of the accelerometer response is caused by hammer excitation: a value of zero indicate that the accelerometer response was caused solely by extraneous noise while a value of one indicates that the response was solely caused by the hammer impact. Coherence generally shows the quality of data in the possible presence of noise. Coherence \(\gamma^2\) is a function of frequency \(f\) and is defined as the ratio of the product of the cross spectrum \(G_{XY}\) and the conjugate power spectrum \(G^*_{XY}\) defined between input \(X (f)\) and output \(Y (f)\), and the product of the power spectra of the output \(G_{YY}\) and input \(G_{XX}\) which are also functions of frequency (equation (5)).

\[
\gamma^2(f) = \frac{G_{XY}(f)G^*_{XY}(f)}{G_{XX}(f)G_{YY}(f)} \quad (5)
\]
The software allowed for sensor sensitivity input, so that raw voltage data from sensors could be presented in acceleration and force units on the software interface. The data was stored and further analyzed as described in the specific method section.

**Specific methods**

The experimental modal analysis yielded fifteen (15) frequency response functions (three FRF per measurement point) for the fixed hammer impact which were then analyzed. These FRF have values at each discrete measurement frequency and generate a frequency response matrix \( H(f) \) at that frequency. The modal method used to identify the modes was the Complex Mode Indicator Function (CMIF). The Complex Mode Indicator Function curve versus frequency range was displayed within the stabilization diagram. The CMIF method is especially suitable since it can identify natural frequencies in highly damped systems, such as the human body where viscous damping is caused by high water content. At the heart of the CMIF algorithm is the singular value decomposition (SVD) method which is an eigenvalue extraction method applied to the frequency response matrix.\(^{13,14}\) The algorithm obtains the natural frequencies from FRF data and is highly effective in discriminating local minima or maxima at the damped natural frequencies of real-valued or complex-valued modes of vibration in the stability diagram. The stability diagram identified the stable (structural) poles and the associated frequencies and damping ratios. Poles are found through a Least Square Complex Function (LCSF) curve-fitting method. The stable poles change little in frequency with increasing curve-fitting polynomial (modal) order; the number of modes that the software uses to curve-fit the CMIF plot.

**Results**

The first two natural bending frequencies of the clavicle were sought in a frequency range between 0.9 and 1.5 kHz based in part on previously reported FEA results\(^9\) and initial measurements.\(^15\) The most prominent modes found in that frequency range, are shown in Table 1.

The Modal Assurance Criterion\(^{16}\) (MAC) was used to verify the independence of the modes (Figure 4). Figure 4 shows a three-dimensional MAC plot showing the relationship among the modes identified. A MAC value of 1 (the red color in the 3D MAC plot) signifies that two mode shapes are orthogonal (at least to the degree of spatial resolution of the measurement locations). A MAC value of 0 (blue color in the checkered MAC plot) usually signifies that the two mode shapes \( \varphi_i, \varphi_j \), are completely different in shape.

Equation (6) presents the formula for the MAC calculation and the symbols “\(*\)” signifies vector complex conjugate and “\( T \)” transposed vector, respectively.

\[
MAC(\varphi_i, \varphi_j) = \frac{\varphi_i^* T \ast \varphi_j^2}{|\varphi_i^* T \ast \varphi_i| \| \varphi_j^* T \ast \varphi_j |} \quad (6)
\]

The mode shapes were displayed graphically and animated within the computer software.\(^{17}\) The mode shape identified visually as the first bending mode of the clavicle occurring mostly in the coronal plane of the cadaver is shown in Figure 5 and is found at a natural frequency of 1014 Hz. The plot shows the line defined by the five measurement points in its initial versus displaced position. The point displacements are indicated by arrows. Figure 6 shows the first bending mode occurring mostly in the axial plane at 1108 Hz. The two axial and coronal

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**Table 1. Mode printout.**

| Index | Frequency | Damping (%) |
|-------|-----------|-------------|
| 0     | 1014.01   | 4.18524     |
| 1     | 1108.04   | 8.032       |

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**Figure 4.** MAC 3D plot.

**Figure 5.** Clavicle first bending mode in axial plane.
plane bending modes are closely spaced together in the frequency domain and their shapes appear similar in a stationary plot (Figures 5 and 6). The modal assurance criterion color map in Figure 4 shows that the modes are independent.

Figure 7 shows the curve-fitted (synthesized) FRF over the frequency range of interest created by using the two modes. Excellent correlation between both amplitude and phase synthesized FRF versus experimental FRF were observed.

**Conclusion**

The experimental modal analysis performed on a single exposed clavicle of a cadaver revealed the first clavicle bending mode in the coronal and axial plane at 1014 and 1108 Hz, respectively, as hypothesized. These two modal frequencies are in the range of the finite element model (FEM) analytical results\(^ {12} \) which predicted these frequencies to be between 900 Hz to approximately 1500 Hz, depending upon boundary conditions. Stiffer (such a rigidly constrained or fixed) clavicle boundary conditions lead generally to higher natural frequencies. Thus, the analytical range for the clavicle bending natural frequencies was defined by the clavicle FEM geometric boundary conditions, which were either specified to be either free or rigidly fixed at both ends in the previously reported studies.\(^ {12} \)

The modal damping ratios corresponding to the two bending frequencies were found to be 4% and 8%, respectively. The modal damping ratio $\zeta$ is the ratio between the damping constant $c$ and the critical damping $c_c$ of that mode, which is a function of its associated natural frequency and modal mass. The identified damping ratios are useful for introducing damping in any analytical modeling of the clavicle since the values cannot be analytically found.

$$\zeta = \frac{c}{c_c}$$

(7)

The benefit of the current study is not only the experimental validation of analytical modal analysis results, but the identification of damping ratios, which could be used as direct inputs into FEA codes.

Vibrational and modal studies of humans are easier when performed for whole body dynamics as accelerometer placement is straightforward. It is true that during measurements acquisition of acceleration data, for example, can be corrupted (in living humans) by the body’s inherent physiological tetanic movements or muscle twitching; a living human cannot stay perfectly still for longer acquisition durations.\(^ {18} \) For inner anatomical structures such as skeletal bones, any experimental vibrational studies could be hindered by access to the structure, especially in living humans. EMA results...
will be influenced by variations in body size. Material property variations also exist between a living body and a cadaver. The material characteristics of cadaver bones are influenced by body processing such as embalming. The geometric boundary conditions of the cadaver clavicle presented in the study were slightly modified from those of a living clavicle since accelerometer attachment considerations dictated that platysma muscles be cut. However, it is the opinion of the authors that the geometric boundary conditions present during the study replicated with high fidelity those found in a living human and allowed better estimation of clavicle modal data than similar studies performed on clavicles removed from the body.

The current study serves multiple purposes. One is the definition of modal data (natural frequencies, mode shapes and damping factors) that may be used to calibrate existing clavicle FEM. Another benefit of the study is to design shock protection systems that filter out load frequency components in the range where the first two bending modes of the clavicle were found, with applicability in the sports and military equipment industry.

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