THE MECHANICAL PROPERTIES CRANIAL BONE

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ABSTRACT

Linear and depressed skull fractures are frequent mechanisms of head injury and are often associated with traumatic brain injury. Accurate knowledge of the fracture of cranial bone can provide insight into the prevention of skull fracture injuries. A set of cranial bone specimens were extracted from 8 crania and, after μCT imaging, the specimens were tested in a three-point bend set-up at dynamic speeds. Important mechanical and morphological properties were calculated for each specimen. The correlations between those parameters were examined statistically. Testing speed, strain rate, cranial sampling position and intercranial variation were found to have a significant effect on some or all of the computed mechanical parameters.

Keywords: Biomechanics, Bones, Dynamics, Human Body

HUMAN CRANIAL BONE has previously been tested in compression, tension and bending (Evans and Lissner, 1957; McElhaney et al., 1970; Wood, 1971; Hubbard, 1971; McPherson and Kriewall, 1980; Margulies and Thibault, 2000; Delille et al., 2003 and Delille et al., 2007; Coats and Margulies, 2006). The majority of these studies concentrate on fetal cranial bone at quasi-static testing speeds. This study uses dynamic testing speeds, adult cranial bone and calculates a comprehensive selection of mechanical and morphological properties of adult cranial bone. Mature adult cranial bone has stiff outer cortical strata and an inner porous lightweight core, the diploë. The diploë acts to increase its thickness thereby increasing its bending strength and is an efficient energy absorbing layer.

MATERIALS AND METHODS

SPECIMEN PREPARATION: Adult cranial bones were obtained from 8 fresh-frozen (refrigeration at −20 °C) cadavers (F=4, M=4; 81±11 years old). 63 specimens (6 cm×1 cm) were prepared (parietal and frontal bones) using a vertical band saw and gently filing the cut faces to ensure accurate dimensions. The thickness and initial curvature of the specimens could not be controlled but care was taken to extract specimens with the least curvature and the orientation of specimens was kept as uniform as possible. All specimens were scanned using a μCT scanner (MTM Tomohawk System; resolution= 56.9 μm) with a solid calibration phantom to allow the bone densities to be calibrated. Before testing, each specimen was instrumented with support structures, made from rigid two-part epoxy resin, to eliminate slippage during testing and to give a flat surface on which to rest the specimens. This setup still allowed normal specimen rotation in bending (span length: 3cm).

MECHANICAL TESTING: Dynamic three-point bend tests were carried out at three testing speeds (0.5, 1 and 2.5 m/s). The test set-up consisted of a static upper fixture with two metal support pins (radius 4 mm) for holding the specimens. This was rigidly attached to a quartz load cell (Type 9331B, Kistler, Switzerland) for measuring the incident force. The impacting pin (radius 6 mm) was displacement controlled using a servo-hydraulic Instron 8502 testing bench. The displacement was measured using a high accuracy laser displacement sensor (Type M5L/20, GEPA, Munich) attached to the Instron actuator head. An independent data acquisition system was used (sampled at 33 kHz). The force–displacement curve for each test was recorded and each test was also captured using high speed video (Phantom V5.1 monochrome camera, Photo-Sonics, England) at between 20,000 and 25,000 frames-per-second.

DATA ANALYSIS: Three specimens were excluded from further analysis due to their lateral slippage during testing. For the remaining specimens, the following mechanical and morphological parameters were calculated (Further details can be found in Motherway et al. (2009)):

Elastic modulus (E): Timoshenko beam theory was used because, on average, the span length (L) to thickness ratio was less than 8:

$$(1)\ E = \frac{F\,L^3}{48\,I_{eq}} \cdot \frac{1}{\left(d_{total} - d_{shear}\right)} \qquad \text{where: (2)}\quad d_{shear} = \frac{F\,L}{4\,G\,A\,K}$$

The shear moduli (G) used were average adult values from Peterson and Dechow (2002, 2003) (parietal:6.8GPa; frontal:6.4GPa). The cross-section (assumed rectangular) required a shear correction factor (K) of 5/6 (Kaneko, 1975; Timoshenko et al., 1975; Caprino et al., 2009). I_{eq}= 2nd moment of inertia, d_{total}/d_{shear}= total/shear displacement, F= load at mid-span and A= cross-sectional area.

Energy absorbed to failure ($U_{failure}$): Calculated from the force–displacement curves and normalised by specimen volume (Margulies and Thibault, 2000):

$$(3)\ U_{failure} = \frac{\int F\,d\delta}{V}$$

$\int F\,d\delta$ = integral of force-displacement curve and V= volume of span length of specimen.

Maximum bending stress or rupture stress (σ_{rupt}): Calculated using simple beam theory (Timoshenko and Goodier, 1970; Margulies and Thibault, 2000). Cross-section assumed rectangular.

$$(4)\ \sigma_{rupt} = \frac{3P\left(\frac{L}{2}\right)}{4\,c^2} - 0.133\left(\frac{P}{c}\right)$$

P= maximum force per unit width of the specimen and c= half-depth of beam.

Morphological parameters: The percent bone volume and percent porosity of the cranial specimens were calculated from the µCT scans using CT Analyser (Version 1.6.1.1, Skyscan, Belgium). The percent bone volume is the proportion of the volume of interest (VOI) occupied by bone where bone is defined as any non-porous region within the VOI. The percent porosity is the volume of pores as a percent of the total solid plus pore volume within the volume of interest, where a pore is defined as a connected assemblage of space (white) voxels. The average specimen thicknesses were calculated from the µCT scan sets using custom code written in Matlab 7.3.0 (The MathsWorks Inc.).

STATISTICAL ANALYSIS: The general linear model statistical procedure and subsequent post-hoc multiple comparison test known as Tukey–Kramer were used to determine the effect of test parameters on the calculated mechanical data. This method was used as it can account for the unbalanced group sizes compared in this analysis. Correlations were measured using the Pearson coefficient and two-tailed p-values are reported. The significance level for all analyses was set as $p<0.05$ and all statistical analyses were performed using SAS 9.1 (SAS Institute Inc., USA).

RESULTS

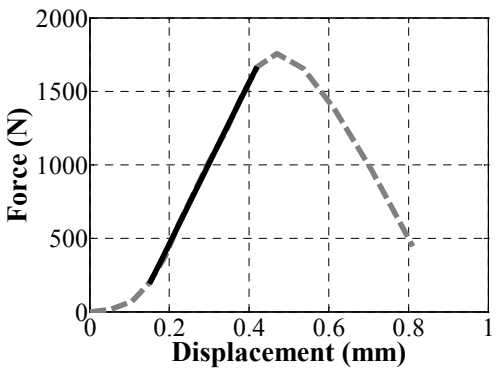

Fig. 1 - Measured force-displacement graph for a cranial specimen tested at 2.5m/s.

Position	Frontal	Parietal
2nd Moment of Inertia (m^4)	$3.37e^{-10}$ ($1.78e^{-10}$)	$2.48e^{-10}$ ($1.78e^{-10}$)
% Bone Volume	74.94 (7.17)	69.59 (10.62)
% Porosity	10.24 (3.72)	13.94 (4.02)
Average Thickness (mm)	6.89 (1.43)	6.30 (1.68)

Table 1. Average Calculated Morphological Properties of Human Cranial Bone. (Standard Deviation in Parentheses).

Table 2. Average Calculated Mechanical Properties of Human Cranial Bone in Three-point Bending. (Standard Deviation in Parentheses). *The result from one specimen was not recorded in this group as the upper portion of the force–displacement curve was off-scale. (F=Frontal, R/L P= right/left parietal)

Speed (m/s)	Position	N	Max Force (N)	E (GPa)	σ_{rupt} (MPa)	$U_{failure}$ (kN.m/m3)	Strain Rate (s^{-1})
0.5	RP	8	734.6 (323.7)	10.33 (7.04)	84.50 (27.37)	105.53 (47.41)	21.09 (4.89)
1.0	RP	9	793.7 (378.6)	9.44 (5.98)	82.98 (25.46)	99.45 (49.21)	30.83 (8.02)
2.5	RP	9	1161.9 (565.1)	12.80 (5.50)	123.12 (43.53)	153.61 (73.58)	109.43 (29.17)
0.5	LP	6	721.7 (379.3)	5.70 (1.73)	82.13 (26.33)	106.93 (52.79)	20.85 (10.16)
1.0	LP	6	584.3 (243.5)	17.69 (13.37)	78.15 (18.31)	76.06 (36.42)	26.00 (8.50)
2.5	LP	8	1228.6 (461.8)	18.12 (14.36)	133.61 (55.03)	198.37 (149.11)	106.70 (29.15)
0.5	F	4	1062.3 (275.6)	4.35 (1.71)	90.80 (13.34)	182.70 (47.84)	21.77 (5.69)
1.0	F	5	1035.7 (461.7)	4.87 (1.93)	102.60 (36.20)	193.19 (91.30)	26.28 (6.64)
2.5	F	5	1315.9 (643.8)	16.34 (10.18)	126.91* (48.76)	138.66* (42.76)	103.98 (18.85)

LOADING RATES: The maximum force to failure (p=0.0007), elastic modulus (p=0.0243) and maximum bending stress (p=0.0012) were significantly affected by the loading rates. The Tukey-Kramer post-hoc test showed that noticeably higher maximum forces were associated with the higher speed (2.5m/s) when compared to both of the lower speeds (0.5 and 1m/s). Similarly, the maximum bending stress was significantly larger at the highest speed (2.5m/s). For the elastic modulus, a significant difference was only found between the lowest and the highest speeds (0.5m/s Vs 2.5m/s). A general trend to note is that the stiffness of cranial bone increases with average impact speed. Significant modest correlations were found between strain rate and maximum force to failure (r^2=0.2861; p<0.0001), maximum bending stress (r^2=0.0844; p=0.0256) and the energy absorbed until failure (r^2=0.1063; p=0.0117).

CRANIAL POSITION: The cranial sampling position (parietal or frontal) was found to significantly affect the resulting values only for the maximum force to failure (p=0.0159), the 2nd moment of inertia (p=0.0117) and the energy absorbed until failure (p=0.0131). Furthermore, it was found that the frontal group required the highest average forces at failure and absorbed the most energy prior to failure. This may be partially explained by the fact that the frontal bone had a significantly higher average 2nd moment of inertia than the parietal bones.

MORPHOLOGICAL PARAMETERS: A modest correlation was found between percent bone volume and both the elastic modulus (r^2=0.1963; p=0.0004) and the maximum bending stress (r^2=0.2708; p<0.0001). It was found that the average thickness for the frontal bones was greater than that for the parietal bones.

DISCUSSION

The calculations of this study make some assumptions about the cranial bone specimens. Timoshenko beam theory makes assumptions about the material's structure: beams are assumed to be composed of homogeneous, isotropic material with a uniform cross-section along their length. Neither of these assumptions are perfect for the present cranial bone specimens. However, using information available from the µCT scans of each specimen, it was possible to account for structural variations along the length of each specimen, both in terms of the porosity and thickness. All specimens were

harvested in a consistent manner with respect to location and orientation across all subjects to minimise the effects of anisotropy and inhomogeneity. Specimens of the least possible curvature were selected and it has been shown that the error caused by a small initial curvature (a mid-span curvature of $\leq 1/10^{th}$ the span length) is negligible (error <1%) (McPherson and Kriewall, 1980). The results from this study compared favourably to those in the literature and a full discussion on this comparison can be found in Motherway et. al. (2009). This study tests adult cranial bone at dynamic testing speeds in comparison with the majority of the studies in the literature which concentrate on fetal cranial bone at quasi-static speeds. In addition, a comprehensive examination of the cranial bone mechanical and morphological parameters are reported.

The results show that impact speed plays an important role in the fracture of adult cranial bone and thus, the viscoelastic nature of cranial bone is evident. Cranial bone is naturally able to adapt with respect to its material response to protect the internal soft tissues of the cranium more effectively with increased loading rate. The advanced subject ages used within this study are not thought to have a significant effect because cranial bone does not tend to loose strength at a rate comparable to other bones due to the large proportion of cortical bone in the inner and outer tables (Yoganandan and Pintar, 2004). The morphological differences that occur between individual cranial vaults and between bone sites on a single cranium proved important. The most notable variables are the porosity, overall bone thickness, the thickness of each of three cranial bone layers and initial radius of curvature. It was found that an increase in the percentage bone volume results in a corresponding increase in elastic modulus and the maximum bending stress. It was also found that frontal bone tends to be thicker, less porous and have a higher percent bone volume than parietal bone and, thus, requires higher forces at fracture and absorbs more energy before fracture. Consequently, the cranial vault appears better able to resist frontal impacts as opposed to side impacts in a dynamic accident situation.

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