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The mechanical properties of cranial bone:
The effect of loading rate and cranial sampling position

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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 and help aid the design of energy absorbing head protection systems and safety helmets. Cranial bone is a complex material comprising of a three-layered structure: external layers consist of compact, high-density cortical bone and the central layer consists of a low-density, irregularly porous bone structure.

In this study, cranial bone specimens were extracted from 8 fresh-frozen cadavers (F=4, M=4; 81±11 yrs old). 63 specimens were obtained from the parietal and frontal cranial bones. Prior to testing, all specimens were scanned using a μCT scanner at a resolution of 56.9μm. The specimens were tested in a three-point bend set-up at different dynamic speeds (0.5, 1 and 2.5 m/s). The associated mechanical properties that were calculated for each specimen include the 2nd moment of inertia, the sectional elastic modulus, the maximum force at failure, the energy absorbed until failure and the maximum bending stress. Additionally, the morphological parameters of each specimen and their correlation with the resulting mechanical parameters were examined.

It was found that testing speed, strain rate, cranial sampling position and intercranial variation all have a significant effect on some or all of the computed mechanical parameters. A modest correlation was also found between percent bone volume and both the elastic modulus and the maximum bending stress.

Nomenclature

A	=	specimen cross-sectional area (at mid-point) (m^2)
c	=	half-depth of beam
d_{total}	=	total specimen displacement (m)
d_{shear}	=	displacement due to shear (m)
$d_{bending}$	=	displacement due to bending (m)
E	=	elastic modulus (GPa)
F	=	load applied at mid-span (N)
G	=	shear modulus (GPa)
I	=	2^{nd} moment of inertia (m^4)
K	=	shear correction factor
L	=	span length (m)
P	=	maximum force per unit width of beam (N/m)
V	=	volume of span length of specimen (calculated by summation of bone pixels in μCT scan set).
$\int F\, d\delta$	=	integral of force-displacement curve (N.m) (see equation (3))
$U_{failure}$	=	energy absorbed until failure ($kN.m/m^3$)
σ_{rupt}	=	maximum bending stress or rupture modulus

Introduction

The mechanical properties of neural brain tissue have been extensively characterised (Miller and Chinzel, 1997; Hardy et al., 2001; Prange and Margulies, 2002; Gefen and Margulies, 2004). However, corresponding studies into human cranial bone, which has 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, 2007; Jans, 2003; Coats and Margulies, 2006) vary significantly. The majority of these studies concentrate on fetal cranial bone at quasi-static testing speeds. However, fetal and adult cranial bone are vastly different materials. Fetal cranial bone is a thin, non-homogeneous cortical bone layer which displays a highly directional fibre orientation (McPherson and Kriewall, 1980). When the cranium begins to mature in early life, the bones of the cranium differentiate structurally into a three-layered composite structure and its central trabecular layer, the diploë, develops a homogeneous grain structure. Mature adult cranial bone, analogous to engineering sandwich structures, has stiff outer strata consisting of cortical bone and an inner energy absorbing lightweight core. With this structural development, the mechanical properties of the cranial bones change: the elastic modulus increases (Margulies and Thibault, 2000) and the diploë acts to increase the thickness of adult cranial bone, thus increasing its bending strength. Furthermore, the diploë is an efficient energy absorbing layer and stiffens the whole sandwich structure.

In the literature, it can be noted that there is large variation in the recorded mechanical properties of cranial bone. This can be attributed to several factors, the most significant of which is the morphological variation between subjects. This affects the possible size and shape of tested specimens. Other factors relate to chosen testing parameters: how tissue is preserved prior to testing (fresh/embalmed), the type of loading used (bending/compression/tension) and the testing speed (quasi-static/dynamic). This present study aims to test, in a three-point bend set-up at dynamic testing speeds, a significantly large set of cranial bone specimens taken consistently from three of the cranial bones, in order to draw statistical conclusions as to the effect of cranial position, testing speed, intercranial variation and morphological parameters on various mechanical properties of cranial bone including elastic modulus.

Materials and Methods

The ethics committee of University Hospital, Leuven, approved this research.

Specimen Preparation:

Adult cranial bones were obtained from 8 fresh-frozen cadavers (F=4, M=4; 81±11 years old, max=97 years old, min=62 years old). All material was thoroughly examined to ensure there was no deterioration due to pathological conditions. All material was refrigerated at -20˚C until the time of specimen preparation and testing, thus minimising the number of freeze-thaw cycles to avoid the risk of bone damage. 63 specimens (6cmx1cm) were obtained from the 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. Furthermore, the orientation of specimens was kept as uniform as feasible to allow for realistic future comparisons (Figure 1).

Specimen Digitisation:

All specimens were scanned using a µCT scanner (MTM Tomohawk System, 72kV, 0.49mA) at a resolution of 56.9µm along with a solid calibration phantom to map greyscale µCT values to bone mineral density. Prior to testing, each specimen was instrumented with support structures, Figure 2(B), made from rigid two-part epoxy resin, to eliminate any erroneous slippage during testing and to give a flat surface on which to rest the specimens. This reduced the effective span length for tests to 3cm (the distance between the two epoxy supports) but still allowed specimen rotation in bending.

Mechanical Testing:

Dynamic three-point bend tests were carried out at three testing speeds (0.5, 1 and 2.5m/s); Table 1 shows the matrix of tests completed. All specimens were allowed to thaw prior to testing.

The test set-up, Figure 2(A), consisted of a static upper fixture with two metal support pins (radius 4mm) 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 6mm) was displacement controlled using a servo-hydraulic Instron 8502 testing bench. The displacement was measured during each test using a high accuracy laser displacement sensor (Type M5L/20, GEPA, Munich) attached to the Instron actuator head. A data acquisition system was used independently of the Instron and all data was sampled at 33kHz. The force-displacement curve for each test was recorded and each test was also captured using a high speed imaging camera (Phantom V5.1 monochrome camera, Photo-Sonics, England) at between 20,000 and 25,000 frames-per-second.

Data Analysis:

Of the 63 specimens tested, 3 were excluded from further analysis due to their lateral slippage during testing, which was visible from the high speed videos and evident from sudden changes in force observed in the force-displacement curves. For the remaining specimens, the 2nd moment of inertia, the sectional elastic modulus, the maximum force at failure, the energy absorbed until failure, the maximum bending stress, and the morphological parameters were calculated.

Euler-Bernoulli beam theory is applicable (Roarke and Young, 1986) when the ratio of span length to thickness is at least 8. The span length to thickness ratios of the tested specimens were found to be, on average, below 8. Therefore, Timoshenko beam theory was used to account for shear deformation (d_{shear}) when calculating the elastic modulus of each specimen as follows (Appendix A):

$$E = \frac{F\,L^3}{48\,I_{eq}} \cdot \frac{1}{\left(d_{total} - d_{shear}\right)} \qquad (1)$$

where:

$$d_{shear} = \frac{F\,L}{4\,G\,A\,K} \qquad (2)$$

The values of shear modulus (G) used were average adult values taken from two studies by Peterson and Dechow (2002, 2003) (parietal shear modulus: 6.8GPa; frontal shear modulus: 6.4GPa). The cross-section of each specimen was assumed to be rectangular and thus a shear correction factor (K) of 5/6 was used (Kaneko, 1975; Timoshenko et al., 1975; Caprino et al., 2009).

For each specimen the energy absorbed to failure (Margulies and Thibault, 2000) was calculated from the force-displacement curves:

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

The maximum bending stress or rupture stress, σ_{rupt}, was calculated using simple beam theory according to equation (4) (Timoshenko and Goodier, 1970; Margulies and Thibault, 2000). It was assumed that each specimen had a regular rectangular cross section. This calculation of bending stress applies solely to the point on the tensile surface of the beam directly under the applied load. In these bending tests, fracture occurred almost exclusively at this point and, therefore, this assumption holds for these tests. For this equation, P denotes the maximum force per unit width of the specimen.

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

Finally, the morphological parameters, namely percent bone volume and percent porosity, associated with each 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 i.e. any binarised solid object 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 that is fully surrounded on all sides in 3D by solid (black) pixels (Skyscan, 2008). 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 data is presented as mean±SD. The general linear model (GLM) 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

An example of the measured force-displacement curves is shown in Figure 3. The various mechanical parameters were calculated for 60 specimens extracted from 8 calvaria. The results are tabulated in Table 2 and the variation of the elastic modulus values are illustrated in Figure 4.

Loading Rates:

From the statistical analysis it was found that loading rates had a significant effect on the mechanical properties of the tested specimens. 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. According to the Tukey-Kramer post-hoc test, it was found 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). However, in the case of the elastic modulus, a significant difference was only found between the lowest and the highest speeds (0.5m/s Vs 2.5m/s) while the middle group (1m/s) was not significantly different from the others. A general trend to note is that the stiffness of cranial bone increases with average impact speed, Figure 5.

The relationships between strain rate and the mechanical parameters were also examined. 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). The correlation between strain rate and elastic modulus was found to be not significant.

Cranial Position:

Of the mechanical properties calculated in this study, 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). On further examination, it was found that the frontal group required the highest average forces at failure (1143.36±478.11N compared to 894.38±460.27N) and also absorbed the most energy prior to failure (173.19±66.35kN.m/m^3 compared to 126.23±85.66kN.m/m^3). This may be partially explained by the fact that the frontal bone had a significantly higher average 2nd moment of inertia ($3.37e^{-10}$±$1.78e^{-10}m^4$) than the parietal bones ($2.48e^{-10}$±$1.78e^{-10}m^4$). The multi-factorial analysis was repeated with the parietal group divided into two groups: left and right. This had no effect on the resulting statistically significant findings other than to slightly alter the p-values obtained.

The interaction between loading rate and cranial sampling position was also examined and was found not to be significant for any of the calculated mechanical parameters.

Intercranial Variation:

Given the variable nature of biological tissues, it was not unexpected that the intercranial variation between the eight calvaria had a significant effect on all mechanical parameters: the moment of inertia (p<0.0001), elastic modulus (p=0.0004), maximum force at failure (p<0.0001), energy absorbed until failure (p<0.0001) and maximum bending stress (p<0.0001).

Morphological Parameters:

The average percent porosity was 10.24±3.72% (frontal) and 13.94±4.02% (parietal) and the average percent bone volume was 74.94±7.17% (frontal) and 69.59±10.62% (parietal). No correlation was found between percent porosity and any of the computed mechanical properties. There was, however, a modest correlation 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). The average thickness and span to thickness ratio for each skull is reported in Table 3. It was found that the average thickness for the frontal bones (6.89±1.43mm) was greater than that for the parietal bones (6.30±1.68mm).

Discussion

The calculations of this study make some assumptions about the cranial bone specimens. While three-point bend tests allow for the use of simply shaped rectangular specimens, the associated Timoshenko beam theory makes assumptions about the material's structure. In general, beams are assumed to be composed of homogeneous, isotropic material and have 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 of the diploë and the variation in thickness. All specimens were harvested in a consistent manner with respect to location and orientation across all subjects in order to minimise the effects of anisotropy and inhomogeneity. While each of the specimens had initial curvature, care was taken to minimise this by selecting the straightest specimens possible. Additionally, it has been shown that the error caused by a small initial curvature (a midspan curvature of less than or equal to one tenth the span length) is negligible (error less than 1%) (McPherson and Kriewall, 1980).

In comparing the present results to the literature, it is important to note any differentiating factors which may cause inter-study variations. Few studies have reported bending properties of adult cranial bone (Hubbard, 1971; Delille et al., 2007) while studies examining fetal or infant cranial bone are more common (McPherson and Kriewall, 1980; Margulies and Thibault, 2000; Jans, 2003; Coats and Margulies, 2006). Adult and fetal cranial bones are physically different, as noted earlier and these differences should be regarded when comparing other studies against this work.

The values for the elastic modulus calculated herein are consistent with those in the literature (Figure 6). As expected, they are higher than those for fetal cranial bone. A study by McPherson and Kriewall (1980) reported, as a result of quasi-static bending tests, that the average elastic modulus of fetal cranial bone (gestational age 40 weeks) was 3.88GPa (fibres parallel to specimen long axis) and 0.95GPa (fibres perpendicular to specimen long axis). The same study reported the elastic modulus of a single calvarium taken from a 6 year old subject as 7.38GPa (parallel fibres) and 5.86GPa (perpendicular fibres). In a later study by Margulies and Thibault (2000), specimens of fetal cranial bone, taken solely from the parietal bones, were tested in a bending set-up. In their case the reported elastic modulus ranged from 0.071GPa (gestational age: 25 weeks) up to 3.58GPa (6 months term). In a more recent study, Coats and Margulies (2006) tested fetal-1 year old cranial bone taken from both the parietal and occipital bones at dynamic testing speeds (1.58 and 2.81m/sec). They reported average elastic moduli of 0.461GPa (parietal) and 0.329GPa (occipital). Hubbard (Hubbard, 1971) similarly tested adult cranial bone using a three-point bending set-up and, in contrast to the present study, used sandwich structure theory to calculate an average elastic modulus of adult cranial bone of 9.5GPa. In Figure 4 the total spread of the calculated elastic modulus values for this study are illustrated. Some outliers exist across the majority of cranial groups (1-8). However, these outliers do not tend to belong to either the same testing speed group or the same cranial position group (parietal/frontal).

Few studies, with the exception of (Margulies and Thibault, 2000; Coats and Margulies, 2006), have examined the maximum bending stress or the maximum energy for failure of three-point bending tests of cranial bone. In the case of the maximum bending stress, Margulies and Thibault report average values ranging from 3.1MPa (30 weeks gestation) up to 44.6MPa (6 months term) at a testing speed of 2.54mm/min and from 4.0MPa (30 weeks gestation) up to 71.6MPa (6 months term) at a testing speed of 2540mm/min. More recently, Coats and Margulies (2006) reported quite low average maximum bending stresses for fetal-1 year old cranial bone of 30.23MPa (parietal) and 14.7MPa (occipital), tested at dynamic speeds of 1.58 and 2.81m/sec. In this study, the average value of the maximum bending stress of adult cranial bone is 85.11±23.55MPa (0.5m/s), 86.44±27.08MPa (1.0m/s) and 127.84±46.88MPa (2.5m/s). This is consistent with the previous studies, as it would be expected that as cranial bone matures the maximum bending stress for fracture would be higher due to its structural superiority over fetal cranial bone. This is also the case when the maximum energy until failure is considered. As cranial bone matures, the development of the diploë layer acts to increase the energy absorbing capabilities of the bone. Margulies and Thibault report average ranges of the energy absorbed to failure as 31.2kN.m/m^3 (25 weeks gestation) to 183.4kN.m/m^3 (6 months term) at a testing speed of 2.54mm/min and 57.5kN.m/m^3 (30 weeks gestation) to 436.1kN.m/m^3 (6 months term) at a testing speed of 2540mm/min. In the present study, the average energy absorbed until failure was 123.15±56.77kN.m/m^3 (0.5m/s), 115.87±72.97kN.m/m^3 (1.0m/s) and 167.82±104.19kN.m/m^3 (2.5m/s). While the upper values reported by Margulies and Thibault for the maximum bending stress and the maximum energy for failure at the higher speed appear to be relatively large compared to those reported here, it should be noted that only one parietal specimen was tested under those test parameters in their study; that specimen may have been an outlier from the general trend, especially when it is noted that it gave values far in excess of the rest reported in their study.

It is evident from the results reported in this study that impact speed plays an important role in the fracture of adult cranial bone. The maximum force to failure, the elastic modulus and the maximum bending stress were all significantly higher at the higher average testing speed. In addition, significant modest correlations were found between strain rate and maximum force to failure, maximum bending stress and the energy absorbed until failure. Thus, the viscoelastic nature of cranial bone is evident from these results. 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.

Another important factor to consider is the morphological differences that occur between individual cranial vaults and between bone sites on a single cranium. The most notable variables are the porosity, overall bone thickness, the thickness of each of three cranial bone layers and initial radius of curvature (Figure 7). In this study, the percent porosity, calculated from each μCT scan on a 2D slice-by-slice basis, ranged from 3.59% to 21.21% and the percent bone volume, similarly calculated, ranged from 49.90% to 92.62%. This shows the degree of variation that can exist between cranial bones of different subjects. 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.

A high degree of inter-individual variation exists between the results for each of the skulls, Figure 4. It must be considered that there may be some parameter that is representative for an individual's skull which might be able to explain why some skulls are stiffer and stronger than others. Should such a parameter exist, it would then be possible to infer more case-specific mechanical properties for modelling purposes (Horgan, 2003; Doorly, 2006; Motherway, 2009) using that parameter to adjust the average skull bone mechanical properties reported here.

Conclusions

This study has examined the effects of cranial position, dynamic testing speed, intercranial variation and morphological parameters on the calculated mechanical properties of adult cranial bone and a number of important conclusions can be drawn:

1) The average elastic modulus of cranial bone, tested in a three-point bend set-up, was calculated to be 7.46±5.39GPa (0.5m/s), 10.77±9.38GPa (1.0m/s) and 15.54±10.29GPa (2.5m/s).
2) Different dynamic loading rates had a significant effect on the resulting values for the maximum force to failure ($p=0.0007$), elastic modulus ($p=0.0243$) and maximum bending stress ($p=0.0012$). Cranial bone proved significantly stiffer with increased dynamic testing speeds. Additionally, significant modest correlations were found between strain rate and maximum force to failure, maximum bending stress and the energy absorbed until failure.

3) Cranial sampling position (parietal or frontal) was a significant factor in the resulting values for the maximum force to failure ($p=0.0159$), 2nd moment of inertia ($p=0.0117$) and energy absorbed until failure ($p=0.0131$). Furthermore, 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.
4) Intercranial variation had a significant effect on all calculated mechanical parameters.
5) The average porosity of the tested specimens was 13.08±4.23% and the average percent bone volume (BV/TV) was 70.84±10.13%. A modest correlation between percent bone volume and both the elastic modulus ($r^2=0.1963$; $p=0.0004$) and maximum bending stress ($r^2=0.2708$; $p< 0.0001$) was found.

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Conflict of Interest Statement

None of the authors has any conflict of interest associated with this manuscript.

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