CHARACTERIZATION OF PEDIATRIC PORCINE SKULL PROPERTIES DURING IMPACT

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ABSTRACT

Falls and inflicted impacts are leading causes of severe brain injury in children. We determined mechanical properties of pediatric porcine skull and suture at impact rates (2.16 and 3.67 m/s) to complement previous quasistatic data. As before, elastic modulus (E) and ultimate stress (σ_{ult}) of cranial bone were significantly higher and ultimate strain (ε_{ult}) was significantly lower than suture. At high rates, E and σ_{ult} had no statistically significant rate dependence. Compared to quasistatic conditions, skull and suture were significantly less stiff, with larger $\varepsilon_{ult,}$ indicating impact at higher rates may be associated with larger distortions of the braincase.

KEYWORDS: infant, viscoelasticity, mechanical properties, brain injury, skull fracture

HEAD INJURIES are the primary cause of mortality in young children, and falls and inflicted impacts are the leading causes of severe brain injury (CDC, 2000). Distinguishing between accidental and intentional injuries remains hindered by a lack of experimental and computational models specific to pediatric head injury. Our broad objective is to understand what types of unique brain injuries occur in children of specific ages who experience falls, vigorous shakes, and inflicted impacts. In this communication, our goal is to define the material properties of skull and suture at high rates, to be used in future computational models simulating the infant head during accidental and inflicted impact.

While there have been multiple measurements of the dynamic material properties of adult skull tissue (McElhaney 1970; Wood, 1971), very little has been reported for pediatric skull and suture material. McPherson and Kriewall (1980a) were the first to investigate the fetal skull material properties in hopes of modeling fetal head molding. They performed tensile and three-point bending tests at quasistatic rates of 8.3×10^{-6} m/s. More recently, Margulies and Thibault (2000) examined porcine and human pediatric skull in tension and three-point bending at modest rates of 4.23×10^{-5} m/s (2.54 mm/min) and 4.23×10^{-2} m/s (2540 mm/min) typical of a slow "crush" event. They found the elastic modulus of pediatric skull was 12 times less than adult cortical compact bone from the outer table of the skull. In addition, there was a significant rate-dependence in both cranial bone and suture. If there is a rate-dependence that extends to high rates, as Margulies and Thibault's lower rate data suggests, then the properties already obtained may not be suitable for a model attempting to simulate low height falls or inflicted impacts. Thus, in order to understand injury risk in infants during falls and inflicted impacts, it is important to obtain pediatric properties at more appropriate rates.

Due to the brain geometry, central nervous system (Pampiglione, 1971), cerebral blood flow (Buckley, 1986), and cerebral metabolism development (Corbett, 1990), porcine tissues are often used to gain insight in mechanisms of human head injury (Armstead, 1994; Hoehner, 1994; Meaney, 1995; Smith, 2000). As a first step toward determining cranial pediatric material properties at high rates, this study focuses on 3-5 day old porcine cranial

bone and suture. It is hypothesized that in addition to the cranial bone being stiffer than suture, the higher test rates will produce larger values of elastic modulus and ultimate stress for both the cranial bone and suture compared with historic data at low rates.

EXPERIMENTAL DESIGN

TESTING DEVICE: A drop test apparatus (Figure 1) was designed to test samples in three-point bending at desired rates. Validation of the drop test apparatus and experimental design was performed by measuring the material properties of a copolymer in three-point bending using the test apparatus and also in tension using a commercial materials testing device (Instron, Canton, MS). Data from both methods were statistically compared.

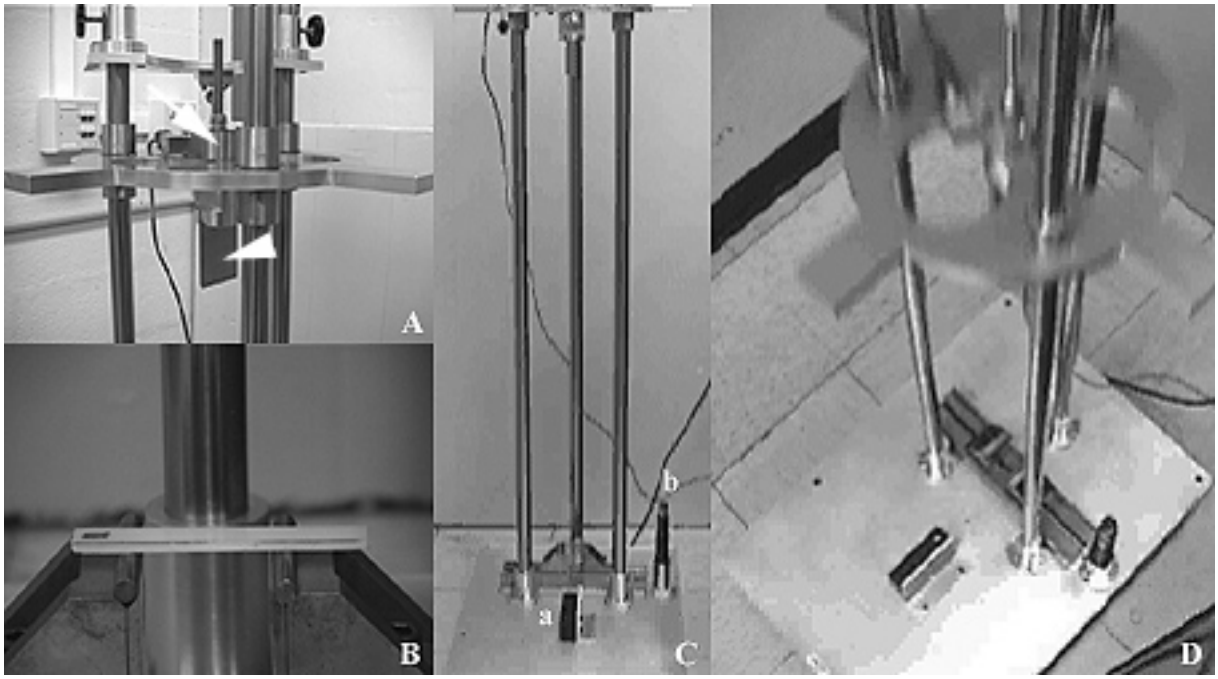

Fig. 1 – Images of the drop test apparatus. Components of the apparatus shown consist of a laser displacement sensor (a) and shock absorber (b) shown at the bottom of (C), a moving plate with attached load cell (arrow) and upper knife (arrowhead) (A), and lower supports (B). (D) A top view of the device after the moving plate is released.

PORCINE BONE AND SUTURE SAMPLE COLLECTION AND PREPERATION: Porcine skull and suture samples were collected by removing the calverium from 3-5 day old piglets within at least 30 minutes post-mortem. The protocol was approved by the University of Pennsylvania Animal Care and Use Committee. Briefly, a craniotomy was performed using a diamond cutting blade (Stoelting Co., Wood Dale, IL, USA) attached to a rotary tool (Dremel®, Racine, WI) under a drip of saline to maintain moisture and prevent overheating in the tissue. Once removed, the entire tissue was wrapped in gauze soaked with mock CSF (see references) solution, placed in a specimen jar, and immediately frozen in a -4°C freezer. Cranial tissue was kept frozen until day of testing. Preliminary tests found that samples frozen longer than 100 days had a strong correlation between storage time and elastic modulus ($p = 0.0002$). Thus, no samples used in this study were frozen longer than 100 days.

On the day of testing, frozen cranial samples were thawed to room temperature (25°C) for 3 hours. Using a diamond cutting blade and rotary tool in a fume hood, 30mm x 5mm rectangular samples were machined. The mock CSF solution was used to keep the samples moist and prevent overheating. Samples were taken from locations indicated in Figure 2 to

minimize the curvature of the beam-shaped sample. In this manner, 2 suture samples and 2 cranial bone (non-suture) samples were obtained from each porcine cranium.

A sanding band attachment to the Dremel® was used on the interior surface of the sample to smooth any bumps and create a more uniform sample thickness. Any alteration to the center of the sample was avoided. Thickness and width were measured and any malformations or peculiarities were noted. Samples were wrapped in gauze soaked with mock CSF for transport to the test device, and tested within 30 minutes after completion of machining. The entire preparation from end of thaw to completion of testing was approximately 2 hours.

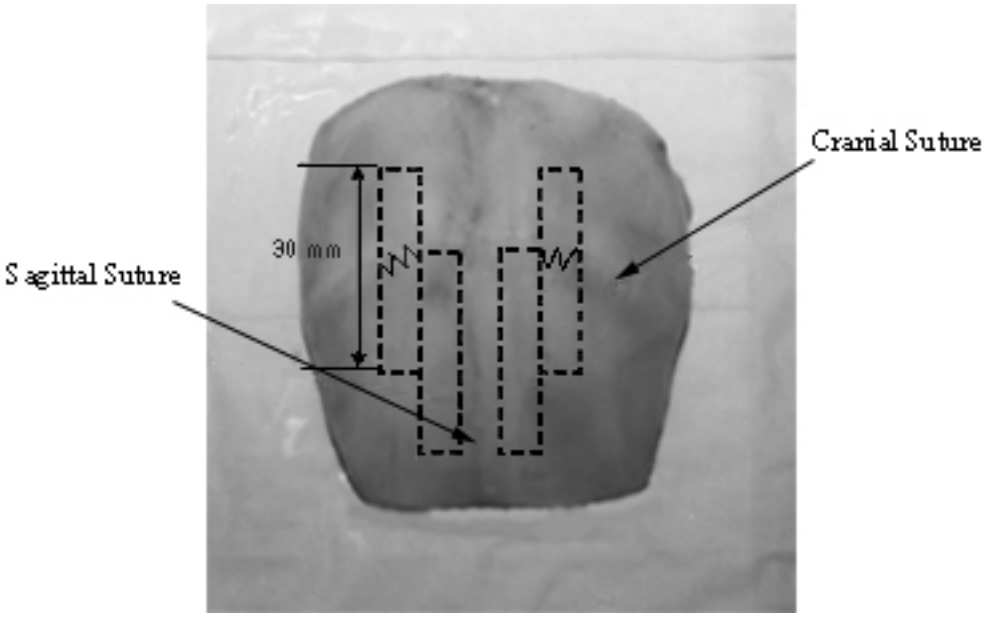

Fig. 2 – Image depicting excised 3-5 day old porcine cranium, overlaid with lines indicating locations of suture and sample removal.

No standard exists for performing three-point bending tests on pediatric skull or suture. Therefore, the testing protocol was loosely based on ASME standard D790, *Standard Test Methods for Flexural Properties of Unreinforced and Reinforced Plastics and Electrical Insulating Materials*. This standard suggests a minimum span length to thickness ratio of 14:1 for samples less than 1.6mm thick. The span length ranged from 16-18mm, thickness ranged from 0.8-1.3mm, and width ranged from 3-6 mm.

COPOLYMER SAMPLE COLLECTION AND PREPARATION: For tensile tests, copolymer polypropylene (Boston Brace International Inc., Avon, MA, USA) samples were machined to a standard dog bone shape. The average width was 10.54 mm and the gage length varied from 21.54-23.17 mm. The thickness of each sample was approximately 2.3mm. For drop tests, samples were machined to 80 x 12 x 2.3 mm rectangular beams as recommended by ASME standard D790. Span was 36 mm for all tests.

TESTING: Porcine and copolymer samples were placed on the lower supports so that the center of the sample or location of the suture lay in the middle of the support span. For cranial bone and suture samples, the moving plate was adjusted to either 0.305 or 0.914 m to produce rates that might occur during low height falls (2.45 m/s and 4.24 m/s, respectively; See Appendix for derivation). For copolymer samples, the moving plate was adjusted to 0.02 m to produce a rate similar to the tensile tests (0.66 m/s).

For the tensile tests, the dog-bone shaped copolymer samples were mounted with ridged rubber grips and pulled to failure using an Instron 8501 material testing device (Instron Corp,

Canton, MS, USA). Each sample was tested at the maximum rate allowable by the Instron, 0.58 m/sec. No sample slipping was observed.

Displacement and force data were collected using a computer data acquisition system (Labview 4.1, National Instruments, Austin, TX, USA) at 10,000 samples per second, and saved onto a laptop computer (Dell, Austin, TX, USA).

DATA ANALYSIS

Because the span length to thickness ratio was at least 14:1, the depth of the beam can be assumed small and the Bernoulli-Euler equation can be applied to calculate elastic modulus:

$$E = \left(\frac{F}{\delta}\right)\frac{L^3}{48I} \qquad \text{Equation [1]}$$

where F/δ is the force-displacement ratio during the linear elastic region of a trace (Figure 3), L is the span of the beam, and I is the moment of inertia of the rectangular cross section of the beam (Timoshenko, 1970).

Ultimate stress was calculated using Timoshenko's corrected version of the beam theory equation which accounts for radial tensile forces within the beam as a result of an applied concentrated load to the center of the beam.

$$\sigma_{xx} = \frac{3PL}{8c^3}y - 0.133\frac{P}{c} \qquad \text{Equation [2]}$$

$$\sigma_{yy} = 0 \qquad \text{Equation [3]}$$

where P is the measured force per unit width, L is the span, c is half of the thickness, and y is the location of interest along the y-axis (outer surfaces at $y = \pm c$) in the center of the beam. The ultimate stress is then calculated by using the maximum force ('A' in Figure 3) divided by the sample width for P in units of N/m.

The flexural strain from three-point bending was calculated from the relationship:

$$\varepsilon_f = \frac{6Dd}{L^2} \qquad \text{Equation [4]}$$

D is the maximum deflection in the center of the beam ('B' in Figure 3), d is the thickness of the sample, and L is the span (ASTM D790).

For the tensile tests, material properties (E, σ_{ult}, and ε_{ult}) were obtained from stress-strain data in a similar manner as the determination of F/δ, F_{ult}, and D_{ult} from the 3-point bending load-deflection data.

A Student's unpaired t-test with a Type I error of 5% was used to determine significant material property difference between rates or tissue type.

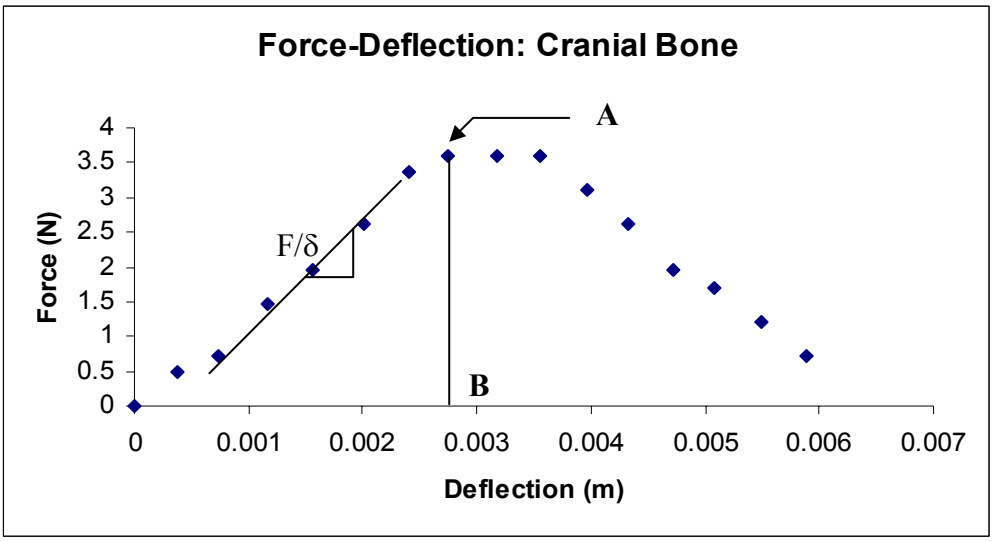

Fig. 3 – Typical trace recorded. **A** indicates the force value used to define P in Equation [2], **B** indicates the deflection value used in Equation [4], and F/δ is determined by a linear regression fit to the elastic region of the curve. A similar stress-strain curve is obtained from tensile tests. In those, **A** would indicate σ_{ult}, **B** would indicate ε_{ult}, and a linear regression fit to the elastic region would determine E.

RESULTS

The average drop test rate of the copolymer was 0.66 m/s. Results from this test were compared to those from the tensile tests performed at 0.58 m/s (Table 1). No statistically significant difference in elastic modulus or ultimate stress was found between tensile and the three-point bending. However, ultimate strain calculated from the drop test was significantly lower ($p<0.05$). These results indicate that the drop test apparatus and three-point bending analysis method are suitable to determine the elastic modulus and ultimate stress of a material, but will tend to underestimate the ultimate strain.

Properties	*Tensile (n=7)*	*3-Point (n=6)*
Flexural Modulus (MPa)	535.0 ± 138.8	652.2 ± 37.1
Ultimate Stress (MPa)	35.2 ± 2.6	33.2 ± 0.6
Ultimate Strain (m/m)	0.240 ± 0.175	0.063 ± 0.004*

Table 1 – Summary of copolymer polypropylene material properties determined from both the tensile and three-point bending tests. Values reported as a mean ± S.D. * $p<0.05$.

The drop test apparatus height was set to test cranial bone and suture from 0.305 m and 0.914 m heights. From Equation [5] in the Appendix, test rates of 2.45 and 4.24 m/sec were expected, respectively. However, due to frictional forces on the guide rods, the resulting average test rates were 2.16 and 3.67 m/s. Table 2 indicates the resulting material properties for each material at each specified height. Bone was found to have a statistically significantly higher ultimate stress and elastic modulus than suture at both the lower ($p<0.001$) and higher ($p<0.02$) drop heights. Conversely, suture was found to have a higher ultimate strain than bone for both drop heights, but was statistically significant only at the lower drop height ($p<0.001$).

Surprisingly, unpaired t-tests showed no statistically significant effect of rate for any of the material properties of pediatric cranial bone. Thus, over the impact rate range tested, pediatric cranial bone resembles an elastic material. For suture, only the elastic modulus at the higher rate was statistically significantly greater than the elastic modulus at the lower rate (p=.0487).

Material	*Elastic Modulus (MPa)*	*Ultimate Stress (MPa)*	*Ultimate Strain (m/m)*
0.305 m drop			
Bone (n=15)	158.9 ± 15.7	10. 8 ± 1.0	0.099 ± 0.010
Suture (n=16)	57.1 ± 8.2	6.0 ± 0.6	0.176 ± 0.018
0.914 m drop			
Bone (n=11)	165.1 ± 18.3	11.2 ± 1.2	0.104 ± 0.011
Suture (n=11)	83.8 ± 9.9	7.7 ± 0.8	0.141 ± 0.021

Table 2 – Summary of porcine cranium material properties found from drop tests. Values reported as the mean ± S.D.

To date, the only material properties of pediatric cranial tissue were reported by McPherson and Kriewall (1980a) and Margulies and Thibault (2000). Integrating these historic data for pediatric cranial bone at quasistatic rates with the data in the current study obtained at high rates (Figure 4) reveals the dramatic decrease in elastic modulus over a broad range of test rates (over 2 orders of magnitude). This finding was contrary to our hypothesis of higher modulus values at higher rates.

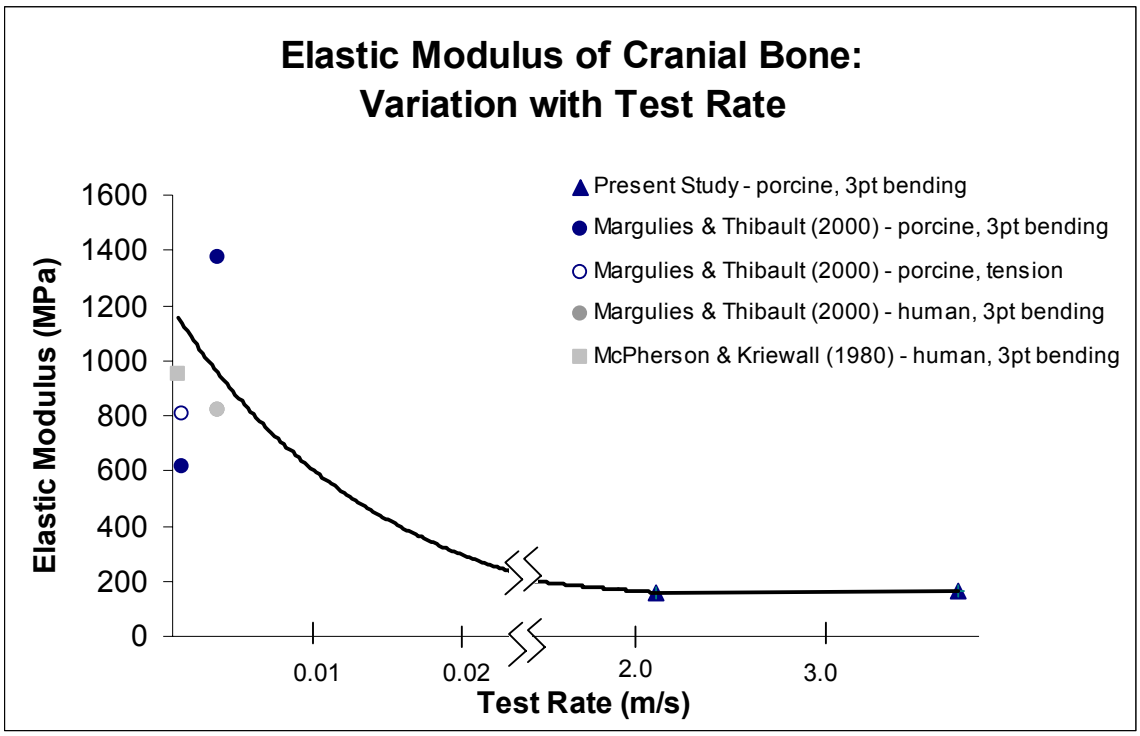

Fig. 4 – Graph of pediatric porcine and human cranial bone elastic modulus data to date. All human data was taken between the age range of 1-2 weeks post term, porcine data was taken between 3-5 days post term. Note scale change in x-axis.

DISCUSSION

There is a paucity of measurements of the material properties of pediatric cranial bone and suture at moderate to high test rates applicable to impact injuries. The device designed for this study has the benefit of not only simulating rates appropriate for impact following free fall, but doing so in a reproducible, controlled manner. Using a copolymer material to determine the accuracy of results compared to tensile tests provides an efficient assessment of the validity of the device. Ideally, validation would be best performed at rates similar to those used in the porcine tests; however, speed limitations of the commercial Instron material testing device prevented this experimental design.

For both of the rates tested, the material properties of the bone were significantly stiffer than suture, as previously reported (Margulies & Thibault, 2000). Thus, the braincase is likely to deform by flexing at the sutures prior to significant deformation of the cranial bone. These braincase distortions may result in brain deformation in the absence of skull fracture. In addition, the ultimate strains for suture were significantly larger than for cranial bone; therefore, significant deformations of the braincase may produce bone fracture before suture failure.

Porcine cranial bone was found to have no statistical significant difference of material properties when comparing test rates of 2.16 and 3.67 m/s, characteristic of an elastic material. Others have shown in the literature that adult cortical bone is a viscoelastic material (Sedlin & Hirsch, 1966; McElhaney, 1976), and Margulies and Thibault (2000) found a rate dependence for pediatric skull, however, these studies were all performed at very slow rates (8×10^{-6} to 4×10^{-2}m/s). No study has been performed on pediatric skull at the rates reported in this communication (2 to 4 m/s). The cranial suture ultimate stress and elastic modulus had a tendency to increase at the higher rate tested, similar to the response of a viscoelastic material. Generally, this increase did not achieve statistical significance except for the elastic modulus (which just barely reached statistical significance). Thus, a viscoelastic response appears to be attenuated at the highest rates.

Given the 100-fold rate range and the variation in species (porcine vs. human), caution should be used when interpolating between the quasistatic and high impact rate data in Figure 4. Further information regarding this transition region must be determined before a definitive statement can be made regarding the overall materials response to a load. However, elastic modulus is significantly attenuated at the high rate of deformation tested, indicating that the skull seems to be more malleable at higher rates associated with falls and impacts than lower rates typical of crush injuries.

Another interesting result was the large magnitude of ultimate strain for both pediatric suture and bone at the high rates tested. Wood (1971) measured adult skull at high test rates (4.23 m/s) and reported a range of ultimate strain from 0.006-0.008 m/m. The results presented in this communication show a 12.5 fold increase for cranial bone and a 22 fold increase for suture over adult values. Margulies and Thibault (2000) found tensile ultimate strain of porcine pediatric cranial bone to be $0.0341 \pm .0069$ mm/mm at the slow rate of 4.23×10^{-5} m/s. High impact rates appear to increase the ultimate strain five-fold. In addition, this increase could be an underestimation as illustrated in the validation tests. Collectively, this data further illustrates the deformability of the braincase at higher rates as well as the need to distinguish tissue responses with respect to age and rate.

Using the equation for energy absorbed to failure given in Margulies and Thibault (2000), further calculation indicates the absorption energy found for cranial bone in these high impact studies was approximately 0.06 $N \cdot mm/mm^3$ for both the 0.3 and 0.9 meter drop tests. This is 40-62% less than the energy absorption reported by Margulies and Thibault for porcine skull

at slower rates. It is understandable that such a decrease would occur given that the high impact allows little time for energy absorption.

The vast difference of material properties from quasistatic to high impact rates underscores the importance of using accurate properties in finite element models and anthropomorphic dummy studies, and also raises interesting challenges. For example, using a braincase material for an anthropomorphic dummy that mimics the elastic modulus, ultimate stress, and ultimate strain properties reported here would be impractical as the dummy's braincase would likely fail frequently and require replacement. It might be more appropriate to mimic the impact absorption energy of pediatric skull.

Several precautions were taken to improve the accuracy and reliability of the results of the present study. Sedlin and Hirsch (1966) evaluated factors affecting material properties of cortical bone. They found that while small changes in temperature (± 5°C) had only minor effects, exposure to dry room air for ten minutes increased the strength of cortical bone. In the present study, samples were consistently wrapped in mock CSF-soaked gauze when not in use as well as transported in a mock CSF filled container. Samples were removed from gauze immediately prior to testing and were exposed to the room air for no more than 1 minute. Another factor that contributes to an increased stiffness is the amount of time the tissue is frozen from collection date to test date. In the early stages of this study, freeze time was found to significantly increase elastic modulus and ultimate stress in samples frozen longer than 100 days. Preliminary tests of a cranial bone frozen for 241 days resulted in an elastic modulus of 801 MPa, almost five times more than the average value. To account for this variation, all samples frozen longer than 100 days were excluded from this study. As a result of these preparations, the variation was minimized. The coefficient of variation for elastic modulus of cranial bone was 11.1%, while previous studies (Margulies and Thibault, 2000) reported that the coefficient of variation for the elastic modulus of cranial bone was 49.2% (3-5 day porcine) and 27.9% (fetal and neonatal human). This improvement is a further illustration of the control and reproducibility of the current experimental approach.

However, even with these precautions, the experimental design is not ideal. It is more preferable to measure the material properties directly, as in tension, rather than estimate them from 3-point bending tests. However, the possibility of obtaining a tensile device able to instantaneously reach the desired rates was small. The three-point bending test utilized has the ability of not only achieving the desired rates, but also simulating the effect impact might have on the material properties. While the beam theory used in three-point bending analysis is often a source of uncertainty, the corrected form used in this paper has been shown to have good approximation if the length is much longer than the thickness of the beam (Timoshenko, 1970), which has been maintained with at least a 14:1 ratio.

CONCLUSION

The present study has shown the importance of testing material properties at rates similar to the applicable environment. In the situation of pediatric head injury following a modest fall or inflicted impact, rates as high as 4.24 m/s can be experienced. The data of this study reinforce previous findings that show cranial porcine bone is stiffer than suture and that the ultimate strain of pediatric cranial bone is many fold higher that of the adult. Importantly, comparison over the broad range of test rates indicates that high rates of impact may result in larger deformations of the braincase and thus result in an increased incidence of brain injury without skull fracture compared to the adult.

The present study provides insight into the role of rate-dependent material properties in understanding the mechanisms of head injury in the pediatric population.

ACKNOWLEDGEMENTS

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REFERENCES

Armstead, W. and C. Kurth (1994). "The role of opioids in newborn pig fluid percussion brain injury." Brain Research 660: 19-26.

Buckley, N. (1986). "Maturation of circulatory system in three mammalian models of human development." Comparative Biochemistry and Physiology 83A(1): 1-7.

CDC (2000). "Traumatic Brain Injury in the United States: Assessing Outcomes in Children. Appendix B,C." National Center for Injury Prevention and Control. http://www.cdc.gov/ncipc/tbi_report/tbi.htm

Corbett, R. (1990). "In vivo multinuclear magnetic resonance spectroscopy in investigations of cerebral development and metabolic encephalopathy using neonatal animal models." Seminars in Perinatology 14(3): 258-271.

CSF Mock cerebrospinal fluid (CSF): KCl 0.220g, MgCl 0.132g, CaCl2 0.221g, NaCl 7.71g, Urea 0.402g, Dextrose 0.665g, NaHCO3 2.066g, distilled water 1000mL.

Hoehner, P., J. Kirsch, et al. (1994). "Dihydropyridine ligand binding decreases earlier in adolescent that in infant swine after global cerebral ischemia." Stroke 25: 2060-2066.

Margulies, S. S. and K. L. Thibault (2000). "Infant skull and suture properties: measurements and implications for mechanisms of pediatric brain injury." Journal of Biomechanical Engineering 122(4): 364-71.

McElhaney, J. H., J. L. Fogle, et al. (1970). "Mechanical Properties of Cranial Bone." Journal of Biomechanics 3: 495-511.

McElhaney, J., V. Roberts, et al. (1976). Handbook of Human Tolerance. Tokyo, Japan Automobile Research Institute.

McPherson, G. and T. Kriewall (1980). "The elastic modulus of fetal cranial Bone: a first step toward understanding of the biomechanics of fetal head molding." Journal of Biomechanics 13: 9-16.

Meaney, D., D. Smith, et al. (1995). "Biomechanical analysis of experimental diffuse axonal injury." Journal of Neurotrauma 12(4): 311-322.

Pampiglione, G. (1971). "Some aspects of development of cerebral function in mammals." Proceeding of the Royal Society of Medicine 64: 429-435.

Pelker, R. R., G. E. Friedlaender, et al. (1984). "Effects of freezing and freeze-drying on the biomechanical properties of rat bone." Journal of Orthopaedic Research 1(4): 405-411.

Sedlin, E. D. and C. Hirsch (1966). "Factors affecting the determination of the physical properties of femoral cortical bone." Acta Orthop Scandinav 37: 29-48.

Smith, D. H., M. Nonaka, et al. (2000). "Immediate coma following inertial brain injury dependent on axonal damage in the brainstem." Journal of Neurosurgery 93(2): 315-22.

Timoshenko, S. and J. Goodier (1970). Theory of Elasticity. NY, McGraw Hill.

Wood, J. L. (1971). "Dynamic Response of Human Cranial Bane." Journal of Biomechanics 4: 1-12.

Appendix A
Drop Test Apparatus Design

A drop test apparatus was designed to test samples in three-point bending at desired rates. Based on a typical three-point bending test jig, the device has two lower rounded supports that the sample rests on and an upper rounded knife that makes contact with the middle of the sample after a free fall from a specified height (Figure 1A&B). Three 1.5 m stainless steel shafts run through linear ball bearings attached to the upper knife's support plate and provide guidance, with minimal amount of friction, as the upper knife falls (Figure 1D). A quick release lever is attached to an adjustable plate at the top of the system allowing for an instantaneous clean release of the upper knife blade. After making contact with the test sample, the upper knife trajectory is terminated by a hydraulic MC 3325-0 shock absorber (Ace Controls Inc., Farmington Hills, MI, USA) (Figure 1C-b). The shock absorber minimizes rebound of the moving plate and allows for 25.4 mm of travel before completely stopping the moving plate. Care was taken to ensure that the upper knife has contacted and completely fractured the sample before the moving plate came into contact with the shock absorber.

Porcine and copolymer samples were placed on the lower supports so that the center of the sample or location of the suture lay in the middle of the support span. The moving plate was adjusted to either 0.305 m or 0.914m to produce rates that might occur during low height falls (2.42 m/s and 4.24 m/s, respectively). Using the conservation of energy equation (Equation [5]) and neglecting air friction, an infant's head would be expected to hit the ground at an impact velocity of 2.42 m/sec and 4.24 m/sec, for falls from 0.305 m and 0.914 m, respectively.

$$mgh = \frac{1}{2}mv^2 \qquad \text{Equation 5]}$$

Once drop height had been established, the moving plate was released via the quick release lever. Attached to the base plate of the test apparatus was a laser displacement sensor (OptoNCDT 1605-100, Micro-Epsilon, Ortenburg, Germany) with a sensing range of ± 50 mm, a resolution of 30 μm, and a frequency response of 10 kHz (Figure 1C-a). The laser detected the displacement of the moving plate and allowed computation of the velocity of the upper knife as it came in contact with the sample. Determination of contact force with the sample was made from a load cell (Sensotec, Columbus, OH, USA) firmly attached to the moving plate and attached to the upper knife via a threaded rod attached to the sensing center of the load cell (Figure1A). Adapters were made to allow a variety of load cells (111.2, 222.4, and 444.8N) to be used with the moving plate, in order to maximize the sensitivity for a variety of sample materials.