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Final summary Overview
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Purpose

The purpose of this project was to develop a rigorous and validated computational toolset for predicting skull fracture patterns in infants. Our future goal is to be able to use the toolset to identify skull fracture patterns from common low height accident falls in infants, and to evaluate the effect of head impact direction, impact energy, and skull thickness on skull fracture patterns. The specific objectives for achieving this goal can be divided into experimental objectives (Objectives 1 and 2) and computational objectives (Objectives 3 and 4):

1. Quantify the mechanical properties of infant and toddler cranial bone and suture at rates similar to head impact
2. Characterize the fracture propagation properties of infant and toddler cranial bone
3. Develop a high-fidelity computational model for predicting crack propagation and resulting fracture patterns in infant cranial bone.
4. Test the validity of the model by simulating experimental fracture pattern studies, existing cadaver studies, and well-witnessed accidental falls in infants.

Project Design and Methods

Experimental Objectives 1 and 2:

All protocols were reviewed and approved by the Institutional Review Boards at the University of Utah and Primary Children's Medical Center. Human cranial bone specimens were collected from autopsy for high-rate anisotropic material property characterization (Objective 1) or fracture property testing (Objective 2). Criteria for specimens were donors < 3 years of age and no prior history of skull fracture, skull malformations, human immunodeficiency virus, or hepatitis. Two cranial specimens, occipital and parietal were collected from each donor and frozen upon removal. Specimens were thawed to room temperature in phosphate buffered saline (PBS) on the day of testing. Stereomicroscopic images of each specimen were taken to make width and thickness calculations, and to quantify interdigitation of the suture to the bone.

High-Rate Material Property Characterization. We previously published data evaluating the high-rate material properties of infant cranial bone tested *perpendicular* to the trabecular fibers.[1] In this project, we use the same techniques to characterize the high-rate material properties *parallel* to the trabeculae fibers. The device used previously was refurbished with a new 25-lb load cell, high-resolution laser displacement sensor, and dual dampers with hard stops. Due to these changes, the device was revalidated against standardized tensile testing using synthetic specimens with similar stiffness to human cranial bone. Specimens were impacted with drops from 20.5 cm and resulted in an average impact rate of 1.68 m/s to match our previous study. Displacement and force data were collected at 10,000 Hz. A 4^{th} order Butterworth low pass filter with a cutoff frequency of 800 Hz was used on each dataset.

Fracture Mechanics Testing. Specimens were tested using a single edge bend test. Specimen dimensions were between $1 < W/B < 4$ and maintained a support span of at least $4W$, where W and B are the specimen width and thickness, respectively. A Psylotech micro load frame (μTS, Psylotech, Evanston, IL) provided displacement control at a rate of 2 μm/s while a 222 N load cell (WMC-50, Interface, Scottsdale, Arizona) measured the resulting force. Data were sampled at 1000 Hz and low pass filtered with a cut-off frequency of 5 Hz. Crack growth was tracked using a confocal microscope (VHX-5000, Keyence, Itasca, IL) and testing paused at each 200 μm increment of crack displacement. At this increment, a stitched image of the total crack was created and crack length and kink angle, if applicable, measured. Testing for each specimen ended when the crack ratio (total crack length/specimen width) was greater than 0.7

Computational Objectives 3 and 4:

The experimental data in Objectives 1 and 2 were used to develop a high-fidelity fracture mechanics-based computational framework for predicting skull fracture patterns (Objective 3) which was validated against cadaver studies (Objective 4). From these validation studies, we discovered that impact angle had a large effect on fracture patterns. Impact angles are rarely observed or quantified in real-world head impacts. This precluded our ability to validate the model against real-world cases, so we examined the effect of small variations in impact angle and impact height on fracture patterns in infants to better understand the potential variability in the real-world.

Model Development. The infant head fracture model and simulation framework centered around a previously developed infant skull model by our team.[2] This infant skull model was created from CT images of a 1.5-month-old infant male patient. The model consists of cranial bones, suture, brain, and a rigid impact plate. Modifications to the original mesh were made due to incompatibility between the continuous shell elements (SC8R) used in the original model and those allowed by the fracture analysis software FRANC3D. The cranial bones were meshed in MSC Patran with tetrahedral 4-noded elements (C3D4). Two-dimensional three-noded and four-noded membrane elements were used to mesh the suture. A convergence study was performed with the new mesh to verify independence of stress prediction with the new mesh. Surface-to-surface contact was represented with a tangential friction coefficient of 0.2 for skull-to-brain, skull-to-plate, skull-to-suture, and skull-to-skull interactions. These contact interactions allow separation between surfaces and are similar to the original definitions by Coats et al.[2] The cranial material properties were modeled as transversely isotropic linear elastic using the anisotropy high-rate characterization data from Objective 1. The skull fracture toughness was assumed to be transversely isotropic with fracture toughness along the fiber direction assumed to be 5 MPa-mm$^{0.5}$, which is 21% of fracture toughness of adult skull.[3] Fracture toughness across the fibers was assumed to be ten times larger than fracture toughness along the fiber direction, based on our experimental data.

Skull fracture prediction. Skull fracture prediction consisted of two procedures: prediction of skull fracture initiation and resulting skull fracture pattern. The prediction of skull fracture initiation was based on the maximum principal stress criterion developed by Coats et al.[2] and further validated by Hajiaghamemar et al.[4] This criterion states that there is a 50% probability of skull fracture in infant parietal bone when the maximum principal stress averaged over a 1.75x1.75mm region is over 25 MPa. In this study, the initiation site of the fracture was determined at the location where the maximum principal stress exceeded 25 MPa for two consecutive time steps. The resulting skull fracture pattern was predicted by the fracture framework. The algorithms used in the framework and further detail about the computational framework are reported in He et al.[5]

Validation of fracture pattern predictions. To evaluate the predictability of the framework, three simulations were performed. In the first simulation, we simulated a 14.7 cm drop test of a 5-month-old infant head specimen as described by Loyd.[6] No information on impact angles was provided. Using trial-and-error, we found that angle set [30°,0°,14.7 cm] gives a crack-initiation location similar to that observed in the experiment. The mass of the model was scaled uniformly to match that reported by Loyd. In the second and third simulations, we simulated two 82 cm drop experiments reported by Weber.[7] Specifically, these simulations were of a 2.3-month-old infant (case A1) and a 4-month-old (case A3) infant. The mass of the model was scaled to match that reported in each experiment. Similar to Loyd, no impact angle data were provided. By trial-and-error, we found that angle sets [-50°,0°,82 cm] and [-40°,-50°,82 cm] gave reasonable initiation locations for cases A1 and A3, respectively, based on the hand-drawn figures of fracture patterns provided.

Evaluation of fracture pattern sensitivity to impact angle and fall height. In case evaluation, errors are commonly found in the estimation of fall height in a child. Further, impact angles are rarely observed or noticed. Therefore, accurate forensic evaluation of skull fracture patterns will require a better understanding of the potential variability due to inaccuracies in fall height or impact angle descriptions. Using the above computational framework, we simulated a right parietal impact from three heights (0.3, 0.6, and 0.9 m) at nine impact angles, resulting in twenty-seven impact cases. The impact angle was defined as a vector using a spherical coordinate system having the origin at the skull's center of mass. For each fall height, the impact angle was systematically varied by adjusting the polar angle to either 15 deg or 30 deg in four azimuthal directions (0 deg, 90 deg, 180 deg, and 270 deg) away from the parietal bone's center of mass. To represent the fall heights, h, an initial velocity was applied to the head model as calculated using the square root of 2*gravity*height. The initial velocity vector was parallel to the vector of the impact angle, perpendicular to the rigid plate position. ABAQUS/EXPLICIT (Dassault Systemes, Waltham, MA) was used for all impact simulations.

Data Analysis

High-Rate Characterization. Specimen testing and analysis was based on ASTM standard D790[8] and was similar to our previous study.[1] The span length to thickness ratio was kept to at least 14:1, so the depth of the cranial bone can be assumed small and the Bernoulli-Euler Equation (1) could be used to calculate elastic modulus, E. In-plane stress (σ_{xx}) was calculated using Timoshenko's corrected version of the beam theory Equation (2) which accounts for radial tensile forces within the beam as a result of an applied concentrated load to the center of the beam.[9] The ultimate stress was calculated by using the peak force in (2). The flexural strain (ε_f) was calculated from Equation (3).

$$E = \left(\frac{P}{\delta}\right)\frac{L^3}{48I} \tag{1}$$

$$\sigma_{xx} = \frac{3PL}{wt^3}y - 0.133\frac{2P}{wt} \tag{2}$$

$$\varepsilon_f = \frac{6t\delta}{L^2} \tag{3}$$

A correlation analysis was used to identify significant increases in E, with age. A paired Student's t-test identified significant differences in E, σ_{xx}, and ε_f with region (parietal/occipital). One-way ANOVAs were used to identify significant differences in age, orientation (parallel versus perpendicular), and region (parietal/occipital) when combined with data from our previous study. Due to limited sample size, interaction effects could not be explored. For all tests, significant differences were defined as $p<0.05$.

Fracture Mechanics Testing. Mode I fracture toughness (K_I, Eq. 4) was calculated at 100 μm increments of stable crack growth. In Equation 4, P_i is the load at crack growth [N], S is the span [mm], B is specimen thickness [mm], W is specimen width [mm], and a_i is crack length immediately before crack growth [mm]. To check the validity of the shape factor function (Eq. 5) used in Equation 4, a load-line compliance calibration was done with the load displacement data of each test and graphically compared to Equation 5.

$$K_{Ii} = \left[\frac{P_i S}{BW^{\frac{3}{2}}}\right] f\left(\frac{a_i}{W}\right) \qquad (4)$$

$$f\left(\frac{a_i}{W}\right) = \left(\frac{3\left(\frac{a_i}{W}\right)^{\frac{1}{2}}\left[1.99-\left(\frac{a_i}{W}\right)\left(1-\frac{a_i}{W}\right)\left(2.15-3.93\left(\frac{a_i}{W}\right)+2.7\left(\frac{a_i}{W}\right)^2\right)\right]}{2\left(1+2\frac{a_i}{W}\right)\left(1-\frac{a_i}{W}\right)^{\frac{3}{2}}}\right) \qquad (5)$$

An initiation K_{IC} with lower and upper bound were calculated for each specimen. The initiation K_{IC} corresponds to the first instance of crack growth in each test. The lower bound will use the load at the end of the linear region and the upper bound will be at the peak of force for the crack growth. The propagation values of K_{IC} are at each instance of growth recorded after the initial growth. The reason for collecting these bounds was due to the nonlinearity of the crack growth likely caused by microcrack formation prior to crack growth.

Due to limited availability of human specimens, we decided to focus on the anisotropic properties of fracture rather than the regional properties of fracture because anisotropy plays a critical role in prediction of fracture patterns. Regional specimens were still collected for early tests on porcine cranial bone. To evaluate the effect of region (parietal/occipital) and/or fiber orientation (along/across) on fracture toughness in human and porcine specimens, K_I was extracted from each R-curve at 2 mm and statistically compared using a one-way ANOVA for fiber orientation effects in human specimens or a two-way ANOVA for fiber orientation and region effects in porcine specimens. A p-value < 0.05 will be considered significant. A linear or exponential curve will be fit to each combination of human, porcine, fiber orientation and region to identify a representative R-curve for each group.

Validation of fracture pattern predictions. The comparison of fracture patterns between the simulation and the cadaver experiments was first performed visually and then quantitatively. The quantitative metrics were (1) the length of the fracture and (2) the orientation relative to the suture lines. No statistics could be performed due to the limited human data set.

Evaluation of fracture pattern sensitivity to impact angle and fall height. For each impact case, the initiation site of the fracture, distance between the impact site and fracture initiation site, final fracture length, linearity of the fracture, and orientation of the fracture were quantified. To quantify the initiation site of fracture, an 8 x 8 grid consisting of 10 x 10 mm squares was overlaid onto the right parietal bone. An array with two elements indicating the grid's column and row index defined the initiation site. The 8 x 8 grid was selected as it was observed to adequately distinguish similar initiation sites (<6mm apart) and dissimilar initiation sites (>17mm apart). The distance from the exact initial impact location and fracture initiation site was calculated by subtracting the three-dimensional simulation coordinates of the initial contact point of the rigid plate to the skull (impact site) and the three-dimensional simulation coordinates of the fracture initiation site. The final fracture length was the true length of the fracture path measured on the outer surface of the right parietal bone. To calculate the linearity and orientation of the fracture, a local 2D coordinate system was defined for each fracture pattern. The origin of the coordinate system was placed at the initiation site of the fracture. The linearity of the fracture pattern was calculated as the true length divided by the Euclidean distance, d, between the endpoints of the fracture. If the fracture initiation site was in the middle of the bone, the Euclidean distance of each segment (d1, d2) was added together. A linearity score of 1 indicates the fracture was perfectly linear.

One-way analysis of variance was performed between environmental variables (impact angle and fall height) and fracture pattern characteristics (distance between impact location and initiation site, fracture length, linearity, and orientation). A Pearson Chi-square test was used to determine the effect of impact angle and fall height on fracture initiation site. All statistical analysis were performed in JMP (SAS Institute, Cary, NC) and significance was defined as $p<0.05$.

Findings

High-Rate Material Property Characterization.[10] Fifteen human pediatric cranial bone specimens were collected from nine infant donors ranging from 32 weeks gestation to 10 months of age. Bending modulus, E, significantly increased with donor age ($p=0.008$). Ultimate stress, σ_{ult}, also increased with age, but variation was large, and it did not reach significance ($p=0.067$). Ultimate strain and modulus of toughness were not significantly correlated with age. Modulus and ultimate stress were higher in the parietal bone (E: 4807 ± 2976 MPa; σ_{ult}: 108.5 ± 42.28 MPa) compared to the occipital bone (E: 3884 ± 3016 MPa; σ_{ult}: 84.54 ± 36.09 MPa), but this was not significant in paired *t*-test analysis ($p>0.08$). Ultimate strain and modulus of toughness were not significantly different between parietal and occipital bone. When data were combined with our previous studies to look at the effect of fiber orientation on material properties, we found that E and σ_{ult} were significantly greater when tested parallel to the trabecular fibers. This effect was significantly greater in infants less than 1 month of age compared to older infants. Trabecular fiber orientation had no effect on strain.

Fracture Mechanics Testing.[11] The fracture toughness of *porcine* cranial bone was 2 times greater across the fibers than along the fibers, but there does not appear to be any region dependence (occipital vs parietal). Based on a small number of specimens that were collected from occipital and parietal bone in *human* infants, the fracture toughness of occipital bone is 1.7 times greater than the parietal bone. We confirmed that anisotropy of human infant cranial bone is critical to skull fracture pattern predictions. Cracks veered sharply when tested across trabecular fibers and ran parallel to the fibers. Further, fracture toughness across trabecular fibers was 10 times greater than along the trabecular fibers. Compared to adults, infant fracture toughness is at least 5 times smaller than adults, highlighting the ability of a fracture in an infant skull to propagate more readily than in an adult.

Validation of fracture pattern predictions.[5] In our simulation of the Loyd cadaveric study, the fracture initiated at the anterior edge of the right parietal bone and reached a final crack length of approximately 37 mm (measured by tracing each crack increment along the mesh outer surface). The crack extended nominally along the direction of trabecular fibers and eventually arrested near the ossification center. The overall kink angle, as measured from the end of the crack to the coronal suture line was 82.6°. Loyd observed a similar linear fracture nominally along the fiber direction with a reported length of 40 mm and overall kink angle of about 83.18°, resulting in excellent correspondence to our prediction. In our simulation of case A1 from Weber, the fracture initiated on the posterior edge of the right parietal bone. It curved slightly and continued to extend nominally along direction of the trabecular fibers. This was very similar to the crack In Weber's study, which curved as it extended away from the posterior edge. The experimentally observed crack from Weber covered about 80% of the right parietal bone length which was slightly longer than our simulation prediction which covered about 70% of the right parietal bone length. A small crack branching was observed in the experiment, but the current framework does not allow for the representation of a branched crack so it was not captured in the simulation. The final simulation of case A3 from Weber was met with similar success. Our algorithm identified two fracture initiation sites; one on the posterior edge of the right parietal bone and the other near the ossification center. In our simulation, these two cracks came close to coalescing but were manually arrested since crack coalescence is currently not possible in the framework. Similar nearly coalescing cracks were observed in the same locations in the experimental drawings. In summary, all fracture predictions resulting in strong similarities to the experimental data, providing strong confidence in the ability of the computational framework to capture real-world fracture patterns.

Evaluation of fracture pattern sensitivity to impact angle and fall height.[12] The initiation site of the fracture was typically near the impact site, but did not often overlap with it. The distance between impact location and fracture initiation site was 8.06±5.9 mm, and was significantly affected by impact angle ($p<0.0001$), but not by fall height. Fracture was initiated on the suture-bone boundary when impact location was within 10mm of the suture line. Fracture was initiated within 5mm of the ossification center when the impact location was within 16mm of the ossification center. The fracture initiation site was significantly affected by the impact angle ($p<0.0001$), but not fall height. The final fracture length significantly increased with increasing fall height, and was not significantly affected by the impact angle. Under the same impact angle, increasing the fall height by 0.3m increased the final fracture length by 21.39±34.26 mm. The largest increase was between 0.3 and 0.9m at an impact angle of [15 deg, 0 deg], which increased the final fracture length by 131mm (4mm to 135 mm). The nonlinearity of the fracture pattern (deviation of the fracture from a straight line) increased with increasing fall heights, but there was substantial overlap between heights and no significant differences were observed in the one-way analysis of variance. Increasing the fall height by 0.3m increased the nonlinearity of the fracture pattern by 24% on average. The orientation of the fracture pattern was significantly affected by the impact angle ($p<0.0001$), but not by the fall height. Generally, the orientation of the fracture was either vertical or horizontal, and could be predicted by the impact location of the bone relative to the ossification center.

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Products

<u>Publications</u>

1. He J, Yan J, Margulies S, **Coats B** and Spear A. An adaptive-remeshing framework to predict impact-induced skull fracture in infants. Biomechanics and Modeling in Mechanobiology. 19:1595-1605 (2020)

2. Metcalf RM, Comstock J, and **Coats B**. High-rate anisotropic properties in human infant parietal and occipital bone. Journal of Biomechanical Engineering. 143(6): 061010 (2021).

3. Yan J, He J, Spear A and **Coats B**. The effect of impact angle and fall height on skull fracture patterns in infants. Journal of Biomechanical Engineering. 143(7): 071004 (2021)

4. Metcalf RM, Comstock J, and Coats B. Anisotropic and region-dependent fracture toughness in human infant cranial bone. *In preparation.*

5. Metcalf RM and Coats B. Validation of a fracture mechanics-based computational framework for predictions of skull fracture patterns using a porcine surrogate. *In preparation.*

<u>Conference Presentations</u>

1. Coats B. Using fundamental fracture mechanics to predict infant skull fracture patterns. NIJ R&D Symposium. Baltimore, MD. February 2019

2. Metcalf RM, Comstock JM, **Coats B.** High-rate anisotropic and region-dependent properties in human infant cranial bone. Summer Biomechanics, Bioengineering and Biotransport Conference. Seven Springs, PA June 2019

3. Coats B. Using linear elastic fracture mechanics to predict infant skull fracture patterns. Computer Methods in Biomechanics and Biomedical Engineering. Columbia, NY. August 2019

4. He J, Yan J, **Coats B** and Spear AD. An adaptive-remeshing framework for prediction of impact induced skull fracture in infants. International Mechanical Engineering Congress and Exposition. Salt Lake City, UT. November 2019

5. Yan J, He J, Spear AD, and **Coats B.** The effect of impact angle and height on skull fracture patterns in infants. International Mechanical Engineering Congress and Exposition. Salt Lake City, UT. November 2019

6. Metcalf RM, Comstock JM, and **Coats B**. Anisotropic and region-dependent fracture toughness in porcine and human infant cranial bone. Summer Biomechanics, Bioengineering, and Biotransport. Virtual, June 2020

<u>Community Presentations</u>

1. *General biomechanics of child abuse.* U.S. JAG Corp, US Army, Virtual. March 2021

2. *A biomechanical approach to skull fracture.* Safe and Healthy Families. University of Utah. December 2019

3. *Just Science Podcast: Just Fundamental Mechanics of Infant Skull Fracture* (2019) https://soundcloud.com/just_science/just-fundamental-mechanics-and-infant-skull-fractures_2019-nij-rd_100

<u>Data Sets/Tools</u>

1. Material property data on human infant and toddler cranial bone

2. Fracture mechanics data on human infant and toddler cranial bone

3. A computational framework for predicting skull fracture patterns on infants

Implications for criminal justice policy and practice

Child abuse cases can be some of the most challenging cases for prosecutors, law enforcement professionals and child protection advocates. Typically, no one other than the accused is present to witness the event, and children (especially infants) are too young to communicate what events led to their injuries. To complicate matters, there are very few scientifically-based validated tools or datasets available to help distinguish between accidental and abusive head trauma in infants. This proposal measured the fracture mechanics properties of human infant bone and developed a computational framework to predict skull fracture patterns from head trauma in infants. This computational framework is the first skull fracture model to incorporate rigorous fracture mechanics theories of crack growth and propagation and lays the foundation to answer questions from medical and legal personnel related to skull fracture patterns in infants.

The availability of a computational toolset, such as this, can be used to evaluate specific cases of skull fracture, can alleviate confusion and uncertainty, and increase overall judicial accuracy. Future studies with the model can answer long-standing forensic questions such as what traumatic scenarios lead to biparietal skull fracture, how much impact energy is required for a fracture line to cross a cranial suture or create a complex skull fracture. The versatility of the model is such that it can continually evolve or be adapted to answer these questions, ultimately resulting in a sustained data stream to help identify the truthful cause of head trauma in infants and young children. Our future work with the model is focused in understanding variability in skull fracture patterns with natural variability in infant skull thickness and to begin to adapt the high-fidelity model into a tool that is readily available to the child abuse medicolegal community.

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