Comparison of Skull Deformations during Head Drop Tests from Experimental Accelerometers and Finite Element Simulations

Carolyn E. Hampton[1] and Karthik Somasundaram[2]
[1]DEVCOM Army Research Laboratory, Aberdeen Proving Ground MD 21005 USA
[2]Medical College of Wisconsin, Wauwatosa WI 53226 USA

ABSTRACT – There are many experimental sources for validating biomechanical models of the head, but relatively little has been done with impact-induced skull acceleration data. This study used data from drops tests in which PMHS heads were dropped 1-8 ft (0.3-2.44 m) onto rubber pads of 40 or 90 durometer. Acceleration data were double-integrated to obtain position, then the change in distance between anterior and posterior accelerometers defined the length deflection. Increases up to 5.2 mm were observed and the 3D position and orientation could be reconstructed. A finite element simulation estimated similar length deflection concurrent with up to 10.0 mm lateral compression. Simulated fractures initiated at 1.83 and 1.52 m for the 40 and 90 durometer pads respectively, similar to the 2.13 and 1.52 m from the experiments. The significance of the deflections has yet to be determined. These deflections and fracture thresholds will be used to further improve the simulation.

INTRODUCTION

Accurate biomechanical models of the head and skull are critical to understand injury mechanisms and improve protective systems. Finite element (FE) models are commonly used models that depend on rigorous validation against experimental data by using various metrics (e.g. strain, kinematics, force, acceleration). Skull fractures are severe injuries associated with dynamic events such as car collisions, industrial accidents, and military combat. Models of skull fracture often use stress or strain-based metrics derived from head accelerations to predict fracture initiation and propagation. However, small transient deflections during the impact may affect force calculations using accelerometer data (Pintar et al 2005).

There was little literature provided on the flexion of the skull under impact, which is difficult to measure due to the high strength of cranial bone and obscuration from the overlaying soft tissues. The most pertinent was the works of Messerer (1880) who quasistatically compressed PMHS (post-mortem human subject) heads after removing the exterior soft tissues. Linear regression of length deflection vs. lateral compression was 0.086 on average. Regression of vertical deflection vs. compression was larger at 0.111 on average.

This study quantified the dynamic deflection of a skull during impact by utilizing standardized drop tests, which offer control and reproducibility. Using data from tests conducted both below and above injurious levels provides further data on the deflection and impact conditions associated with fracture initiation.

Address correspondence to Carolyn E Hampton, Neurosurgery Research 151, 5000 W National Ave, Milwaukee WI 53295. Electronic mail: carolyn.e.hampton.civ@army.mil

METHODS

Experiments

Linear acceleration data (12.5 kHz, CFC 1000) from 70 left lateral head drops from 1-8 ft (0.3-2.44 m) (Yoganandan et al, 2003) were obtained from the public NHTSA Biomechanics Database (NHTSA, 2025). 13 unembalmed PMHS heads (average 62 years old, 3.75 ± 0.38 kg) were isolated at the level of the occipital condyles. Three-axis accelerometers were attached at shaved sites on the anterior and posterior skull surface. A nine-axis array (NAP) was also mounted at either the crown or right side. All initial positions and orientations relative to the center of gravity were documented in the original reports.

Each specimen was subject to 3-7 drops, each increasing height by 1 or 2 ft, onto a 2" (50 mm) thick 40 or 90 durometer Neoprene pad laid over a load cell (12.5 kHz, CFC 1000). Testing ended when a fracture was noted by CT scan or palpation or the force decreased or exceeded maximum load capacity.

The original reports indicate that 8 of the 13 specimens sustained a fracture, and describe most of these fractures as singular temporal and parietal fractures, with only one being described as comminuted.

The NAP was used to calculate the rotational acceleration, velocity, and position (Padgaonkar et al, 1975). However, the appropriate corrections, including those accounting for the initial accelerometer orientations, had already been applied to the linear accelerometers and no further modification was needed.

The linear accelerometer data was integrated to obtain the velocity, with the initial velocity calculated from the drop height. This was integrated again to obtain position, with the initial position of each drawn from the original reports. The change in the distance magnitude between the anterior and posterior accelerometers was the skull deflection. The position data was also used to make 3D plots of the head position and orientation over time.

Finite Element Model

Simulations were run in LS-DYNA explicit MPP solver v15.0.1. A tetrahedral finite element head (3.87 kg) was prepared to match the experiment average mass and size, replacing the single layer skull material with a user-defined material with 3 stress-based failure modes (Alexander & Weerasooriya, 2021). Accelerometer elements were added at the position and orientation described in the original reports.

Properties for the hexahedral mesh rubber pads were previously characterized (Sahoo et al, 2013). A surface-to-surface contact was added between the head and pad. The head has an initial velocity corresponding to each experiment drop height. The final FE model is shown in Figure 1.

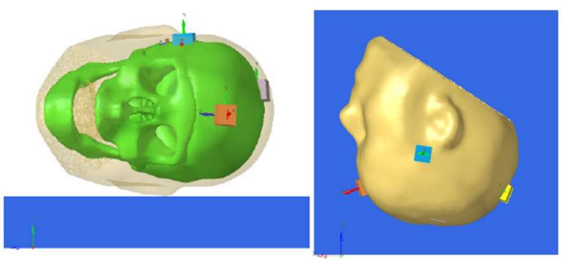

Figure 1: Two views of the FE model with anterior (red), right (blue), crown (purple) and posterior (yellow) accelerometer blocks.

RESULTS

Experiments

Drop tests between 2-8 ft onto the 40 durometer pad had peak forces of 4.1-8.4 kN and fractures most often initiated at 7 ft (2.13 m) which corresponded to an impact speed of 6.47 m/s.

Drop tests between 1-6 ft onto the 90 durometer pad had peak forces of 4.5-11.9 kN and fractures most often initiated at 5 ft (1.52 m), at which the impact speed would be 5.47 m/s.

Skull deflection could be calculated for 86% of the drop tests and increased along with drop height. Deflections at the lowest drop heights were small to imperceptible. The fracture-inducing 8 ft drop onto a 40 durometer pad had the highest deflection at 5.21 mm increase in anterior-posterior length. The highest deflection for the 90 durometer pad was 2.55 mm. In general, the deflection-time showed a similar shape and duration to the force-time.

Finite Element Model

Simulated drops between 1-10 ft onto the 40 durometer pad resulted in peak forces ranging from 6.1-14.1 kN and fractures on the left lateral surface appeared for drops from 6 ft or higher.

Simulated drops between 1-6 ft onto the 90 durometer pad had peak forces between 3.8-11.4 kN. Fractures initiated at 5 ft and looked identical to the simulated fracture from the 7ft, 40 durometer drop.

The simulated fracture was always a linear crack that initiated on the underside near the foramen magnum. With increasing impact severity, it propagated upward through the parietal and temporal bone.

Simulated skull deflections for 1-8 ft drops were 0.2-1.8 mm increase in anterior-posterior length (Figure 2). The deflection increased sharply for fracturing drops. The corresponding lateral compression, which was not available experimentally, was 2.2-10.0 mm. Increases in height were larger at 1.0-4.1 mm.

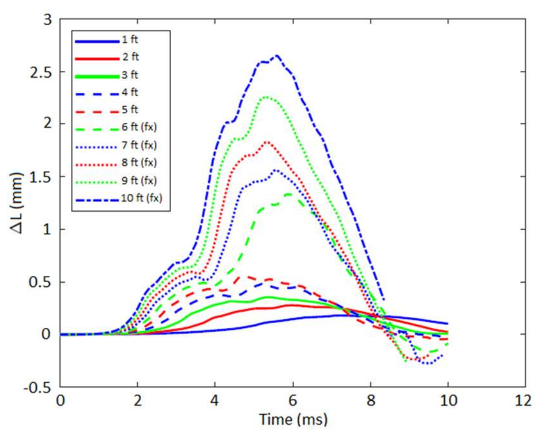

Figure 2: Simulation skull deflection-time by height, fracture-inducing drops marked with “fx”.

DISCUSSION

The experimental deflection for the fracturing drops (2.5-5.2 mm) may be high enough to introduce significant error, as Pintar et al (2005) suggested a limit of ±2 mm when working with accelerometer data to infer distal force. The simulations attributed the deflection to movement of the anterior accelerometer. The posterior accelerometer moved little, possibly because of the greater thickness of the posterior skull.

The experimental fracturing drop heights were 7 and 5 ft for the 40 and 90 durometer pads, respectively. The simulations were similar, predicting 6 and 5 ft.

However, the FE head used stress-based failure limits derived from quasistatic testing of excised cranial bone, which may be different from the in-situ, dynamic nature of the drop tests. The simulated fractures were suspected to be less extensive than the experimental fractures which had no provided images.

Limitations

The deflection could not be calculated for all of the experimental drops. This was attributed to biases on the accelerometer channels, magnified by integration and accumulating over time. It was rare for deflection measurements to return to and stay at zero. This also resulted in warping of 3D position plots.

The experiments did not include a left or bottom accelerometer (presumably due to impact interference and unfeasible to attach, respectively) which prevented lateral and vertical measurements. The simulation could provide these measurements, enabling a linear regression of the length:width (0.16) and height:width (0.47). These are larger than observed by Messerer (1880), but there were many differences in the loading conditions and load rate.

The FE head model itself relied on a linear elastic material model for the skull, when it is well known that bone has some rate-dependence (McElhaney et al, 1970). The skull was one element thick in the face and underside, but remeshing these areas for better flexural strain profiles would be difficult for such complex geometry. Similarly, capturing biovariability may be best with a scaling tool to avoid for individual meshes.

Future Work

The experimental data needs to be reanalyzed to be determine whether significant errors were introduced by the small skull deflections. This might be best done with an iterative approach due to interplay between the NAP calculations and integrations for deflections. The original CT scan data could be reanalyzed for more detail on the extent of the skull fractures.

For the simulation, the bone material model could be updated to include rate-sensitive effects, with these drop tests providing 5-10 m/s data to compliment the existing quasistatic data. A scaling tool would also be helpful to capture natural variations in shape, or different FE head models could be used.

CONCLUSION

Accelerations from lateral drop tests were used to calculate that the skull deflection can be up to 5.2 mm and fracture initiated at 5 or 7 ft drop heights. Future work will determine if this introduces significant error. The FE simulation reproduced the drop tests, but fracture occurred at 5 or 6 ft drop height. Future work will include the dynamic drop test data to develop a rate-sensitive model and improve the fracture initiation and propagation.

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REFERENCES

Alexander, S.L. and Weerasooriya, T. (2021) Implementation and Validation of Finite Element Model of Skull Deformation and Failure Response during Uniaxial Compression. Journal of the Mechanical Behavior of Biomedical Materials 115: Paper 104302.

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

Messerer, O. (1880) Elasticitat und Festigkeit der menschlichen Knochen. J.G. Cotta'schen Buchhandlung, Stuttgart Germany.

National Highway Traffic Safety Administration, NHTSA Biomechanics Database (accessed July 2025).

Padgaonkar, A.J., Krieger, K.W., and King, A.I. (1975) Measurement of Angular Acceleration of a Rigid Body using Linear Accelerometers. Journal of Applied Mechanics 42(3): 552-556.

Pintar, F.A., Yoganandan, N., and Baisden, J. (2005) Characterizing Occipital Condyle Loads under High-speed Head Rotation. Proc. 49th Stapp Car Crash Conference, pp. 33-47. Society of Automotive Engineers, Warrendale PA.

Sahoo, D. et al. (2013) Anisotropic Composite Human Skull Model and Skull Fracture Validation against Temporo-parietal Skull Fracture. Journal of the Mechanical Behavior of Biomedical Materials 28: 340-353.

Yoganandan, N. et al. (2003) Biomechanics of Lateral Skull Fracture. Proc. IRCOBI Conference, Lisbon Portugal.