Assessment of Strain Patterns in the Brain from Real-World Acceleration Data from Collegiate Football Players

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ABSTRACT

The purpose of this study was to determine how linear and rotational accelerations affected the location of high strain areas in the brain. Acceleration pulses obtained from instrumented football helmets were used as the input to the SIMon finite element model (FEM) of the brain. The acceleration pulses used for this study were isolated from a dataset that included 1,712 impacts recorded during one season of a NCAA football team. A multiple step process was used to select pulses of interest for modeling. First, the pulses were sorted by severity along each axis in linear and rotational acceleration. Next, each pulse was evaluated to determine the contribution of off-axis acceleration. The inclusion criterion for this study was that the off-axis acceleration could not be more than 30 percent of the acceleration of interest. The maximum acceleration of the pulses ranged from 27.2 to 56.3 g's in linear acceleration and 580 to 4180 rad/s^2 in rotational acceleration. These selected acceleration pulses were used as the input to the SIMon FEM (v4.0). The location of high strain and the Cumulative Strain Damage Measure (CSDM) were compared for all simulations. A quantitative measure of the different location of high strain elements was evaluated with three dimensional metrics. Relative strength of linear and angular peak acceleration and total velocity change as predictors of CSDM was also evaluated. The rotational acceleration and velocity were more predictive of CSDM with p-values of 0.06 and 0.002 for a second degree polynomial regression. In comparison, the p-values for the linear acceleration and velocity were 0.71 and 0.38. This study demonstrates that the strain patterns in the brain are more related to rotational acceleration than linear acceleration. Additionally, the spatial distribution of the high strain elements varies with the direction of acceleration in a potentially important way.

INTRODUCTION

Traumatic brain injury (TBI) is a prevalent injury in the United States with an estimated 1.7 million TBIs annually (Faul et al., 2010). Of these injuries, approximately 75% can be classified as concussions or Mild Traumatic Brain Injury (MTBI) (Faul et al., 2010). Many MTBIs go un-reported because the injured person does not consider the injury severe enough to seek medical help. In one study of self-reported head injury in collegiate football players, only 23.4% realized the symptoms they experienced were indicative of a concussion (Delaney et al., 2002). The injury mechanisms for this injury are still not well understood. To better prevent this injury in the future, advances in injury prediction are necessary. Collegiate and high school football presents a unique opportunity to study this injury because of the large number of impacts resulting in head acceleration and concussion. With instrumented helmets, researchers can study injurious and non-injurious impacts to these players (Guskiewicz et al., 2007; Mihalik et al., 2007; Duma and Rowson, 2009).

Currently, there are several head injury Finite Element Models (FEMs) available to simulate the response of the brain to a variety of impact conditions (Zhang et al., 2001; Takhounts et al., 2008). These models can evaluate strain in each element of the brain tissue; therefore, they facilitate the study of the strain response of the brain given complex acceleration data. The synthesis of the football acceleration data and the strain data from FEMs can elucidate the response of the brain given a wide range of impact conditions.

METHODS

The purpose of this study was to evaluate real world football impacts with a FEM and evaluate the resulting strain values in the model elements. To complete this modeling, the recorded impact data was sorted so there was one pulse for each primary direction with minimal off-axis velocity changes (X, Y, and Z linear and angular velocity). The model results were then compared to evaluate potential relationships between impact parameters and strain response.

Model Information

SIMon version 4.0 (Takhounts et al., 2008) was selected since it is has improved anatomical accuracy over earlier versions of the model. All simulations were run on a Linux cluster with LSDYNA version 971 (LSTC, Livermore, CA). The simulation time was 10 milliseconds with a maximum time step of 0.145 μs. The input data to the model were velocity boundary condition curves implemented with linear and rotational time vs. amplitude vectors for all three axes.

Acceleration data

The SIMon model was selected to analyze head acceleration data from one football season of a NCAA university. There were 1712 total impacts recorded and no diagnosed concussions for this dataset. The data acquisition system used was the Head Impact Telemetry System (HITS) (Simbex, Lebanon, NH) with a six degree of freedom accelerometer package placed inside the player's helmet. The sensors were coupled to the head with foam padding to record head acceleration instead of helmet acceleration. Data were recorded for all practices and games. The threshold for an event was defined as an acceleration of 10g or greater. When this criterion was met, 10 milliseconds of data were wirelessly downloaded to a side-line computer. All instrumented players gave written informed consent with Institutional Review Board approval from Virginia Tech and the Edward Via College of Osteopathic Medicine. Further description of the data acquisition technique is described by Duma et al (Duma et al., 2005).

The purpose of this study was to determine changes in the model response due to the impact configuration. To accomplish this, the database of impacts was reduced to six pulses that targeted largest velocity changes along each axis of interest. Velocity was selected, instead of acceleration, because when it is calculated by integrating acceleration it is dependent on the peak and the duration of acceleration. There was one pulse for each axis of linear and rotational velocity. To select these pulses, an in-house Matlab (Mathworks, Natick, MA) program was written that sorted the pulse by severity along each axis. Next, the program incrementally stepped through the pulses, from highest to lowest delta-V, and calculated the contribution of off-axis velocity change. The pulse was discarded if these off-axis components exceeded 30% of the delta-V of the pulse of interest. As an example, for the linear X-pulse selected, the maximum delta-V for the X-axis was approximately -4 m/s. The Y-axis and Z-axis were -1 m/s and 1.2 m/s, respectively. This pulse had the seventh highest maximum X-velocity in the dataset. This threshold was selected because smaller off-axis thresholds forced the primary pulse to be a relatively low severity impact. Figure 1 is a histogram of all of the resultant linear velocities for the dataset with the range of selected linear hits shown with a red box and the angular hits shown with a green box. Figure 2 is a histogram of the rotational velocities with the same annotations for the range of selected linear and angular hits. These plots illustrate that for the selected pulses, the linear component of the angular impacts fell in the lower range of the values; however, the rotational component of the linear impacts was higher than the majority of the other events.

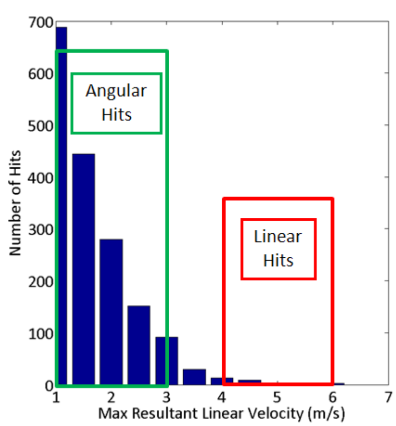

Figure 1: Distribution of maximum resultant linear velocity and the range of selected pulses.

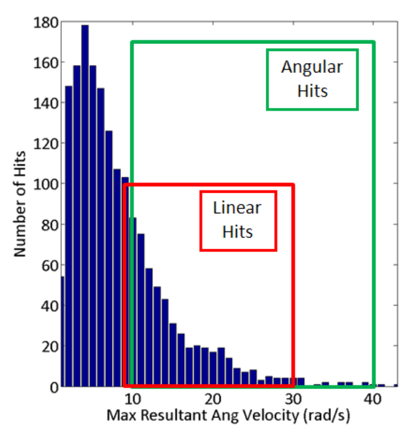

Figure 2: Distribution of maximum resultant angular velocity and the range of selected pulses.

Analysis of Results

Once the simulations were complete, there were two methods of assessing the differences in the model response. The first was a qualitative and quantitative assessment of the distribution of high strain elements. To assess the strain levels, plastic strain was plotted using the fringe plot menu in LS-PREPOST (LSTC, Livermore, CA). Plastic strain was selected because this is the metric used to calculate one of the SIMon injury metrics, the Cumulative Strain Damage Measure (CSDM). The locations of these highest strain elements were compared using varying strain values. To quantitatively assess the location of high strain elements, the centroid of the high strain elements was calculated and compared across simulations. The other method for determining differences in model response was calculating the CSDM for each impact. CSDM as a function of acceleration and velocity was investigated.

RESULTS

The results of the simulations demonstrated spatial differences in the location of high strain values with changes in impact direction. Figures 3 and 4 illustrate these differences in location. The percent strain highlighted was varied to best show high strain locations in each simulation. Varying strain thresholds were used as opposed to the same strain threshold because pulse magnitude variation strongly influences the volume of tissue exceeding a given strain value. The figures are designed to show similar volumes of tissue to compare spatial distribution rather than strain thresholds. Dichotomous selection of elements based on similar thresholds for these pulses shows substantial volume variation which obscures spatial distribution.

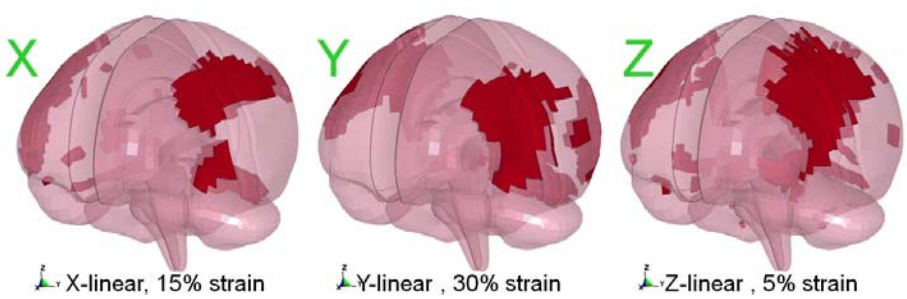

Figure 3: Distribution of high strain elements for isolated linear pulses.

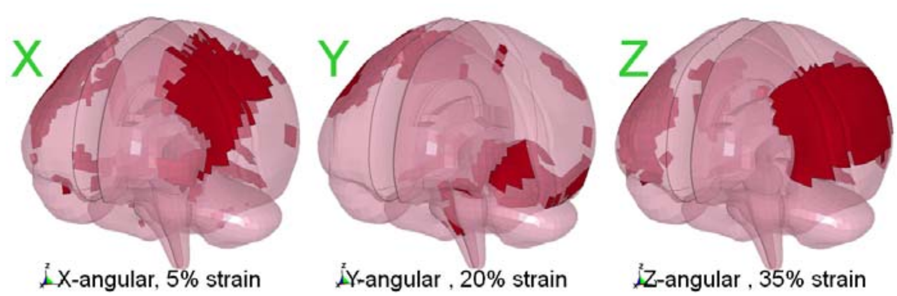

Figure 4: Distribution of high strain elements for isolated angular pulses.

The centroid location of the elements with high strain changed with the different pulses. Figure 5 includes 3 plots illustrating these locations for each scenario. These are plotted at the strain rates for the least severe pulse (5% strain); therefore the cases that experienced more severe pulses had a majority of elements highlighted which forced the centroid closer to the origin. These spatial differences could potentially indicate different functional areas of the brain potentially injured after impact.

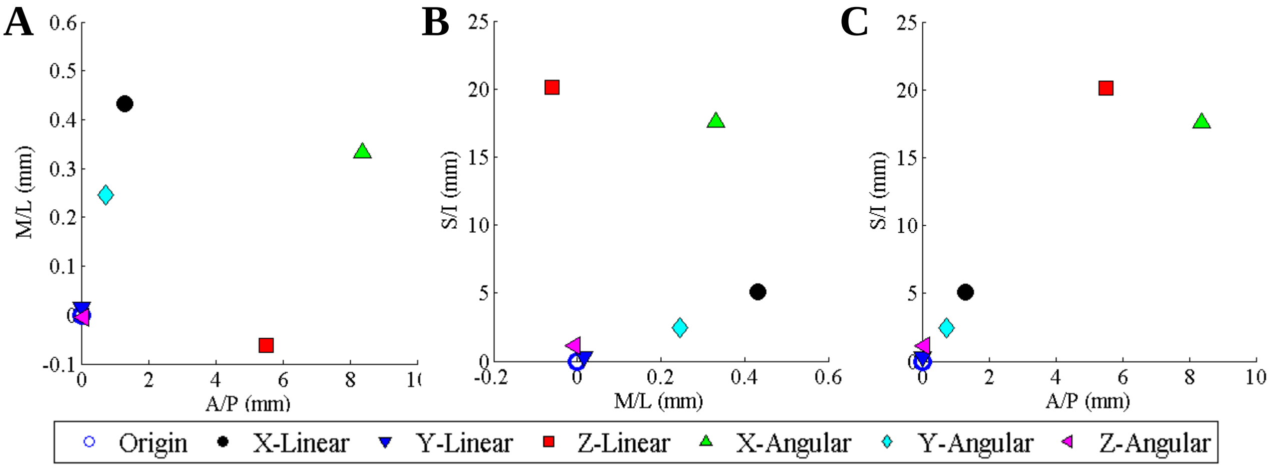

Figure 5: Changes in centroid location for all the brain elements with different pulse directions. A) X-Y plane or A/P by M/L plane. B) Y-Z plane or M/L by S/I plane. C) X-Z place or A/P by S/I plane.

To evaluate possible trends in the data, the resulting CSDM was plotted as a function of velocity and acceleration for both linear and angular values. Figure 6 and Figure 7 are plots of the CSDM as a function of linear and angular acceleration and Figure 8 and Figure 9 are the same plots for velocity instead of acceleration. To compare the results of these plots, p-values for second degree polynomial fits were calculated. There was minimal association between linear acceleration or velocity and CSDM with p-values of 0.71 and 0.38. In contrast, the angular acceleration and velocity plots suggest that as these variables increase the resulting CSDM increases with p-values of 0.06 and 0.002.

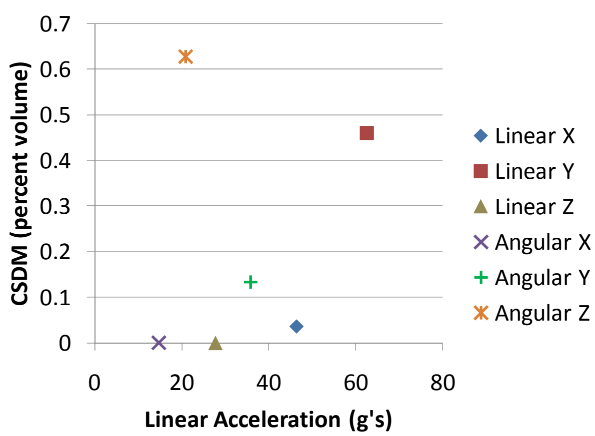

Figure 6: CSDM as a function of linear acceleration.

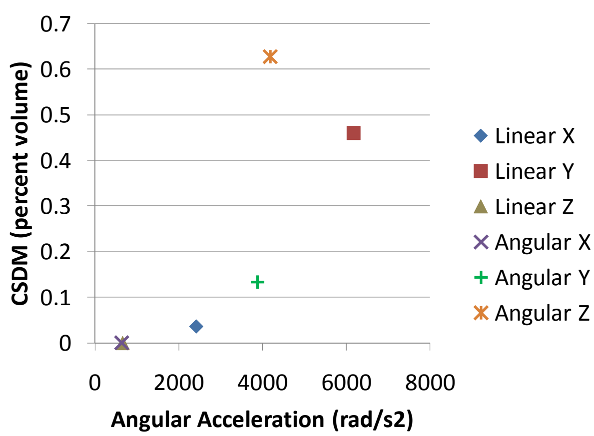

Figure 7: CSDM as a function of angular acceleration.

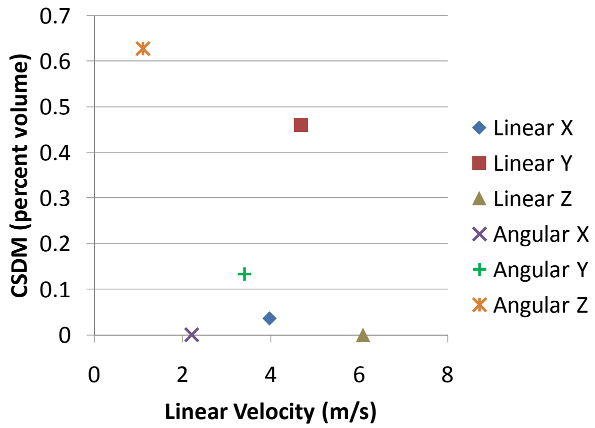

Figure 8: CSDM as a function of linear velocity.

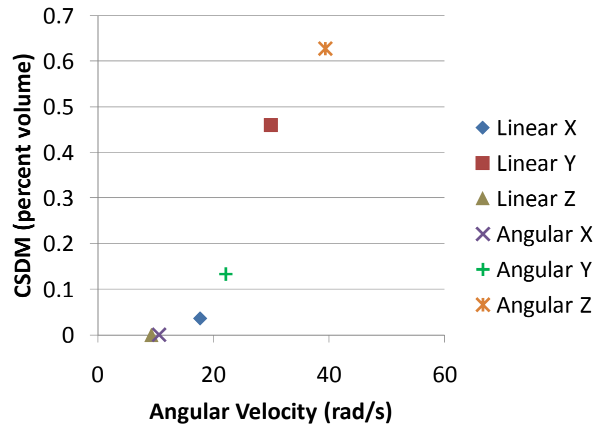

Figure 9: CSDM as a function of angular velocity.

Discussion

Future work will include selecting additional pulses that further decouple the source of acceleration. Specifically, angular velocity was not minimized in the linear acceleration pulses. The current study results suggest that angular velocity had the most influence on head injury metrics; therefore, the linear velocity pulses should have small angular velocity inputs. Due to the small number of pulses in the study, another one of the tasks for future research is to evaluate all of the pulses with SIMon. This larger dataset would identify possible trends in the data. Finally, the boundary conditions of a FEM are critical to proper model response. To evaluate the effect of the current SIMon boundary conditions, future analysis may include spatial distribution of high strain elements without consideration of the outer layer of brain elements.

CONCLUSIONS

Changes in the primary axis of impact resulted in different locations of high strain elements as observed with both qualitative and quantitative measures. The model also shows that angular inputs dominate strain values. This strain in the brain, as measured with CSDM, better correlated with angular velocity as seen by increasing CSDM values with increasing angular delta-V. Additionally, angular delta-V was a better predictor of CSDM than angular acceleration because of the duration component of acceleration captured by this metric. Real world impacts with high single-axis velocity change demonstrated changes in model response, even with confounding off-axis components.

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