Analysis of loading curve characteristics on the production of brain deformation metrics

Andrew Post[1], Evan S. Walsh[1], T. Blaine Hoshizaki[1], Michael Gilchrist[2,1]

[1]Neurotrauma Impact Science Laboratory, University of Ottawa, Canada

[2]School of Mechanical & Materials Engineering, University College Dublin, Ireland

Abstract

Traumatic and mild traumatic brain injuries are incurred as a result of the complex motions of the head after an impact. These motions can be quantified in linear and rotational accelerations which cause the injurious levels of brain deformation. Currently it is unclear what aspects of the linear and rotational acceleration loading curves influence injurious brain deformation. This research uses the University College Dublin Brain Trauma Model (UCDBTM) to analyze the loading curve shapes from a series of centric and non-centric impacts to a Hybrid III head form fitted with different hockey helmets. The results found that peak resultant linear acceleration did not predict brain deformation measures. The results also indicated that due to the complex nature of the interaction between loading curve characteristic and tissue parameters, there was no commonality in curve shape which produced large magnitudes of brain deformation. However the discriminant function did show that angular acceleration loading curve characteristics were more commonly used to predict brain deformation than linear acceleration loading curves.

Keywords

Finite element modelling, Traumatic brain injury, Reconstruction, Injury biomechanics, Concussion

Corresponding author: Andrew Post, University of Ottawa, 200 Lees Ave. A-106, Ottawa, K1S 5S9, Canada. Email: apost@uottawa.ca

Introduction

There is a high economic, emotional, physical and medical cost associated with brain injury in everyday life. In car crashes brain injury often results in severe neurological defects or in the worst cases, death. In sports, helmet use has largely eliminated traumatic brain injuries, but concussions have remained prevalent, and may have undetermined long term effects on neural tissue[1]. In an effort to prevent these injuries scientists undertake research to better understand the injury mechanism in an effort to predict and prevent them from occurring. While this research has led to several prominent scientists proposing that any impact to the head can be measured and quantified by the linear and rotational acceleration curves, the injury itself is more closely linked to brain deformation[2]. This brain deformation is the result of the influence of the complex three dimensional loading curves and their interaction with the brain tissue[3,4].

Brain injuries typically fall into two categories, diffuse and focal. Diffuse injuries such as concussion are highly dependent on the amount of rotation an impact delivers to the brain along with the linear component[2]. Focal injuries such as subdural hematoma are thought to be more associated with linear acceleration[5]. However how these linear and rotational acceleration time histories interact with brain tissue may prove more interesting. The linear and rotational accelerations produced from a head impact causes compression, tension and shearing in different directions and regions of the brain[6]. Researchers have found that different brain tissue regions respond differently to various rates of loading, which would influence how easily the area would be affected by the rate of application of these loading curves[7,8]. This rate dependence has been of particular importance in regions such as the brainstem. The complex nature of the brain tissue and how it interacts with the loading curves from an impact are a possible reason why current kinematically based brain injury metrics such as peak linear acceleration and the Gadd severity index (GSI) have been unsuccessful at predicting brain injury[9]. This lack of correlation between peak kinematic variables and injury criterions such as the GSI and injury would suggest that there are aspects of the complete loading curve which are highly influential upon the creation of injurious levels of brain deformation.

Finite element modelling provides an opportunity to undertake research in this area. Finite element models of the human head are created to represent the material properties and complex interactions of neural tissue. These models allow for the simulation of impact conditions and how the linear and rotational loading curves influence the simulated brain tissues. Many researchers to date have estimated the amount of deformation which may be associated with various risks of injury using a finite element model[4]. Some have used complete computer based simulations[10,11,12]; while others have used physical models to generate the linear and angular acceleration loading curves that define the results of a potentially injurious impact[4]. From this reconstructive research it has been found that maximum principal strain and von Mises stress are brain deformation metrics which have some correlation to brain injury. However these researchers were attempting to discover the most pertinent variable to predict brain injury, and as a result did not examine how the loading curves contributed to the production of the deformation of brain tissues.

While maximum principal strain and von Mises stress, amongst others, have been identified by past research it remains unclear how the linear and angular acceleration time histories contribute to brain deformation. Yoganandan et al[12] generated some artificial acceleration/deceleration loading curves to examine how the duration of acceleration/deceleration influenced brain strains. They did not however examine how the characteristics of these curves contributed to the generation of maximum principal strain. Post and co-workers[14] examined three basic loading curves with identical areas and duration but different time-to-peak They found that their model produced different magnitudes of brain deformation. This research supported the notion that simply examining the peak resultant linear and rotational acceleration values may not be descriptive enough to predict the magnitude of the resulting brain deformation. This conclusion was further validated by recent animal and FE modelling of brain injury conducted by Lamy et al[15] who found that a combined stress-time variable may be more able to predict concussion severity rather than peak magnitudes.

The purpose of this research was to examine how the linear and rotational acceleration loading curve characteristics from a series of centric and non-centric impacts to ice hockey helmets influenced peak maximum principal strain and Von Mises stress using a discriminant statistical analysis.

Methods

A pneumatic linear impactor was used to impact the ice hockey helmets. The linear impactor consisted of a frame housing the impacting arm and a sliding table to which a 50th

percentile Hybrid III head and neck form were attached. The sliding table (12.8 ± 0.1 kg) allowed for motion of the Hybrid III after the initial impact. The mass of the impacting arm was 16.6 ± 0.1 kg and was propelled into the helmeted Hybrid III head form at 4.5 ± 0.05 m/s to match the velocity of the CSA ice hockey standard[16] (figure 1). The tip of the impacting arm was capped with a hemispherical nylon pad covering a MEP disc. The Hybrid III head form was instrumented with a 3-2-2-2 accelerometer (Endevco 7264C-2KTZ-2-300) array to measure linear and rotational accelerations in all three axes[17]. The accelerometers were sampled at 20 kHz with a 15 ms data collection which would begin when the loading curve passed 3 g. The data was collected using a TDAS Pro Lab system (Diversified Technical Systems) and was filtered using a 1000 Hz low pass Butterworth filter as per the SAE J211 convention. The x-axis is defined as facing forward from the head CG, the y-axis to the left of the head and the z-axis vertically upwards.

Twenty four individual helmets were impacted, of which 12 had vinyl nitrile liners and 12 had expanded polypropylene liners. The helmets were impacted at five sites (table 1; figure 2) designed to create different linear and rotational loading curve responses[18]. The results of the impacts are shown in table 2. These linear and rotational loading curves in x, y and z axes were then used as input for the model in a similar method to that used by

Willinger and Baumgartner[19]. The loading curves were applied to the model at the centre of gravity and the resulting deformation of the brain from the impact motion measured.

Table 1. Testing impact locations.

	Location on headform	Impact angle
Site 1	Ant intersection of the mid-sagittal and absolute transverse planes	15° elevation in the mid-sagittal plane towards the impactor
Site 2	Right intersection of the coronal and absolute transverse planes	No vertical or horizontal rotation was applied to the vector
Site 3	Midpoint between the ant mid-sagittal and right coronal planes in absolute transverse plane	45° rotation in the transverse plane
Site 4	Midpoint between the posterior mid-sagittal and right coronal planes in absolute transverse plane	-45° rotation in the transverse plane
Site 5	Posterior intersection of the mid-sagittal and absolute transverse planes	-45° rotation in the transverse plane

Table 2. Linear and angular acceleration responses from the helmeted hybrid III impacts used as input for the University College Dublin Brain Trauma Model (UCDBTM).

		Peak Acceleration	
Site	Material	Linear (g)	Angular (rad/s^2)
1	VN	113.6 (12.9)	5614 (863)
	EPP	105.1 (20.2)	5910 (1233)
2	VN	100.3 (7.3)	7843 (493)
	EPP	100.9 (7.9)	9179 (598)
3	VN	116.8 (7.4)	7888 (569)
	EPP	128.2 (19.7)	9745 (804)
4	VN	70.3 (6.7)	8890 (973)
	EPP	78.5 (3.4)	10410 (890)
5	VN	68.2 (5.6)	4949 (801)
	EPP	67.2 (6.3)	6077 (492)

The finite element model of the human brain used in this research was developed in Dublin, Ireland and is known as the University College Dublin Brain Trauma Model (UCDBTM)[20,21]. The geometry of the model was taken from medical imaging of a male cadaver. The head model was comprised of scalp, a three-layered skull (cortical, trabecular and cortical bone), pia, falx, tentorium, CSF, cerebrum, cerebellum and brain stem. The neural tissue differentiated between grey, white and ventricular matter. The scalp was modelled using shell elements, cortical and trabecular bone with brick elements, the dura with membrane elements, CSF with brick elements, pia with membrane elements, falx and tentorium with shell elements and the cerebrum cerebellum and brain stem with brick elements[20]. The model was validated against cadaveric experiments that measured intracranial pressure[22] and brain motion[23]. Overall, the model was comprised of 10,192 hexahedral elements. The type of reduced integration element used in this version of the UCDBTM was C3D8R with hourglass control.

The material properties for the model (tables 3 and 4) were taken from previous cadaveric research. A linearly viscoelastic material model with large deformation theory was chosen to model the brain tissue. The behaviour of brain tissue was described as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus[20]:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is a decay factor[20]. The parameters used to describe the characteristics of the grey and white matter of the brain tissue were taken from work of Zhang et al[26]. For the brain skull interaction, the model had a sliding boundary between the pia and CSF layers[26]. The contact algorithm allowed no separation between the pia and CSF to properly represent the absence of any gaps at the CSF-cerebrum interface. The CSF was modelled using solid elements with the bulk modulus of water and a low shear modulus. The sliding interfaces had a friction coefficient of 0.2[27].

Table 3. Finite element model material properties

Material	Young's modulus (MPa)	Poisson's ratio	Density (kg/m³)
Scalp	16.7	0.42	1000
Cortical Bone	15 000	0.22	2000
Trabecular Bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx	31.5	0.45	1140
Tentorium	31.5	0.45	1140
CSF	Water	0.5	1000
Grey Matter	30	0.49	1060
White Matter	37.5	0.49	1060

Table 4. Brain tissue material parameters for the UCDBTM

	Shear Modulus (kPa)		Decay Constant	Bulk Modulus
	G_0	G_∞	(s^{-1})	GPa
Grey Matter	10	2	80	2.19
White Matter	12.5	2.5	80	2.19
Brain Stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

For curve analysis the linear and rotational x, y, z and resultant acceleration loading curves were broken down into five measurable characteristics: a) time to peak, b) duration of impact as defined by when the resultant acceleration loading curve changes from a negative slope to a zero slope, c) slope to peak, d) peak magnitude, and e) integral of the entire loading curve (figure 3; figure 4). A discriminant analysis was used to identify how much variance each of the chosen loading curve characteristics in x, y and z components and the resultant linear and rotational acceleration accounts for the resulting brain deformation metric (as depicted in figure 5). The discriminant analysis was run by SPSS statistical analysis software. For this discriminant analysis a backward stepwise method was used. This method includes all the possible variables in its discriminant function used to predict the groups of maximum principal strain (MPS) and Von Mises stress (VMS). It then removes the loading curve characteristics from the discriminant function which do not add to it successfully predicting the resulting brain deformation target value using a Wilk's Lambda score of 3.84. For this type of analysis to work, the target brain deformation needs to be in discrete groups, therefore in this study the maximum principal strain and Von Mises stress responses were broken down into increments of 5% strain and 1,000 Pa to establish comparable categories. The results indicated which loading curve characteristics contributed to the discriminant function that was most successful in predicting the resulting brain deformation magnitudes and the percentage that the function was correct in predicting the respective magnitudes.

Results

The results of the discriminant analysis are presented in tables 5 and 6. Table 5 depicts the variables accounting for the most discrimination between groups and the chance of correctly predicting the maximum principal strain and table 6 shows the variables and chance of correctness for predicting Von Mises stress. The percent chance of correctness shows the ability of the discriminant function to use the chosen variables to correctly predict the resulting magnitude of MPS and VMS.

Table 5. Analysis results showing the variables found to discriminate the most between groups in maximum principal strain and percent correct prediction for the discriminant function which uses all the variables listed below each brain part

	Maximum principal Strain			
	Grey matter	White matter	Cerebellum	Brain stem
	resultant angular intergral z linear slope y angular integral	y angular time to peak z linear time to peak resultant angular integral	y linear integral x linear integral resultant linear integral x angular slope y angular time to peak y angular integral angular duration	z linear integral z linear peak y angular time to peak resultant angular peak z angular time to peak linear duration
% correct	43	59	87	86

Table 6. Analysis results showing the variables found to discriminate the most between groups in Von Mises stress and percent correct prediction for the discriminant function which uses all the variables listed below each brain part

	Von Mises stress			
	Grey matter	White matter	Cerebellum	Brain stem
	resultant angular integral z linear integral	-no variables qualified-	x linear peak y angular integral x linear integral x angular slope resultant angular time to peak	x angular slope z linear integral
% correct	27	n/a	62	48

The results indicate that for maximum principal strain and Von Mises stress there were no distinct commonalities amongst the five curve characteristics chosen to predict the brain deformation metrics. There were also no commonalities in curve shape characteristic between the variables included in the analysis between maximum principal strain and Von Mises stress. This means that there was not a single loading curve characteristics which were used in all discriminant functions used to predict the resulting magnitudes of brain deformation.

Overall, the ability of the variables chosen for the analysis to correctly predict the resulting maximum principal strain was low for the grey and white matter, at 43 and 59%. The values were higher for the cerebrum and the brain stem (87% and 86%), however the values for these two groups were assigned to 3 or 4 target ranges and thus increases the predictive likelihood of success. The ability of the variables to predict the Von Mises stress were also low (table 6), and notably no variables accounted for the variance in the white matter.

Discussion

Anatomically, the results confirm findings reported in the literature. The brain is a complex system, and researchers have shown different brain tissues can have different responses to strain and the rate of application of that strain[20]. Also, certain areas may have lower thresholds to injury, and therefore be more susceptible to certain types of loading than others[7,8]. In addition, brain tissue itself has been shown to be anisotropic in nature. The complexity of brain tissue and how it interacts with impact loading curves was the foundation for analysing all the curve characteristics in this study.

Brain injury is the result of motion of the brain and this study attempted to use characteristics of the linear and angular acceleration loading curves to predict the stresses and strains which are associated with brain injury. When examining the ability of the discriminant function to predict the resulting maximum principal strain and Von Mises stress in the brain interesting results were found. For the functions, the angular components of the loading curves were more commonly used in the equation to predict the MPS or VMS than the linear components. This supports previous literature on concussive impacts indicating that angular acceleration is more influential in the creation of damaging brain deformation than linear acceleration[18,28,29,30]. Even though this relationship is indicated, the discriminant functions included aspects of the linear acceleration loading curve, which supports a combined linear and angular loading theory to concussive brain injury.

When examining the ability of the discriminant function to predict the specific brain injury metrics used in this study it was found that the equation predicted magnitudes of maximum principal strain more successfully than Von Mises stress. This may be a result of the differences in the way the formulation of MPS and VMS is created. Although the ability to correctly predict either brain deformation metric was low for the grey matter and white matter of the brain tissue (below 60% correct). The higher prediction for MPS in the cerebrum and brain stem is a result of a tight grouping of a high number of values in a specific target category, which would improve the chance of a correct prediction.

Of the five curve characteristics that were used in this analysis, no single curve characteristic was commonly used for the prediction of maximum principal strain and von Mises stress in the regions of the brain analysed. Interestingly, peak linear or peak angular resultant acceleration was not found to account for any variance for either MPS or VMS, but aspects of these loading curves were used to predict these brain deformation metrics although with a low success rate. The use of multiple different loading curve characteristics for each part of the brain suggests that different aspects of the acceleration loading curves can influence different areas of the brain. These results also indicate that due to the wide variety of curve characteristic combinations used to account for deformation metric variance, one variable as a predictor of brain deformation may be inadequate.

The limitations inherent to this research lie in the assumptions surrounding the modelling techniques and the method used to break down the loading curves. The UCDBTM treats the brain tissue as homogeneous and may therefore limit the results. These results would suggest that perhaps analysing the response of a more refined model in terms of anatomical structures and characteristics may provide additional insight in terms of brain deformation response. The constitutive properties and material characteristics are derived from literature typically taken from cadavers and as such may not represent the true nature of the central nervous system in response to impact. Also, the use of the hyperelastic model for the brain tissue in shear is specific to this FE model and the results would change if the material model was different. The use of a linear viscoelastic model may not be appropriate for impacts of the magnitudes represented in this study. It is possible that using a non-linear model may be more appropriate to model concussive brain injury[31]. Also, this version of the UCDBTM used C3D8R elements with no hourglass control algorithm. The procedure used to generate the loading curves for these statistical analyses was to impact a Hybrid III headform fitted with an ice hockey helmet. The use of a Hybrid III headform has limitations in that it is not biofidelic and is typically only used for antero-posterior impacts. In this case the impacts were both centric and non-centric in nature which is not what this particular head and neck surrogate was designed for. As a result, the acceleration loading curves produced may not be representative of a brain injury.

While this study attempted to categorize the nature of the complex loading curves, the 5 parameters chosen here do not fully cover all possible curve characteristics and how they can influence brain deformation. Since finite element models use the entire dynamic response curve to calculate brain injury metrics it may be a more effective tool to predict risk of injury. The model is a simulation of the human system and, while validated, is an approximation of true physical response. The variables chosen in this study to represent the characteristics of the loading curve may not have been adequate in predicting brain deformation. The slope, for example was simplified into a straight line, when the shape of the curve to the peak is much more complex. In addition, this study only used two brain deformation metrics to represent brain tissue deformation. It is possible that if other brain deformation metrics were utilized such as strain rate or product of strain and strain rate more commonalities may be discovered with dynamic impact response measures.

Conclusion

This research presents a unique methodology to examine how characteristics of acceleration loading curves influence the creation of brain deformation metrics in the University College Dublin Brain Trauma Model. The results of the discriminant analysis indicate that angular acceleration loading curve characteristics are more commonly used as a predictor of MPS and VMS than linear acceleration loading curve characteristics. The discriminant analysis was better at correctly predicting the magnitude of maximum principal strain than Von Mises stress, which is a result of the method in which these metrics are calculated. It was also found that this method did not have high percentages of success in predicting either MPS or VMS, indicating that the curve characteristics chosen for this analysis may not represent enough of the unique characteristics of the loading curves to establish an adequate predictive function. Finally, angular acceleration loading curve characteristics were more influential than linear acceleration loading curve characteristics in the creation of the brain deformation metrics measured.

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