Finite element analysis of the effect of loading curve shape on brain injury predictors

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Original Article

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Prediction of traumatic and mild traumatic brain injury is an important factor in managing their prevention. Currently, the prediction of these injuries is limited to peak linear and angular acceleration loading curves derived from laboratory reconstructions. However it remains unclear as to what aspect of these loading curves contributes to brain tissue damage. This research will use the University College Dublin Brain Trauma Model (UCDBTM) to analyze three distinct loading curve shapes meant to represent different helmet loading scenarios. The loading curves were applied independently in each axis of linear and angular acceleration, and their effect on currently used predictors of TBI and mTBI. Loading curve shape A had a late time to peak, B an early time to peak and C had a consistent plateau. The areas for all three loading curve shapes were identical. The results indicate that loading curve A produced consistently higher maximum principal strains and Von Mises Stress than the other two loading curve types. Loading curve C consistently produced the lowest values, with loading curve B being lowest in only 2 cases. The areas of peak Von Mises Stress and Principal strain also varied depending on loading curve shape and acceleration input.

Keywords: Brain injury; concussion; finite element modelling

1. Introduction

In the past, researchers using animal models have linked mechanisms of injury to damaging levels of brain deformation caused by linear and angular loading curves (Viano et al., 1989; Kleiven, 2007). These loading curves reflect head motion and are represented by three dimensional linear and angular head kinematics resulting from either a blow or resulting from inertia (O'Donoghue, 1999). Research has supported this phenomenon showing that different injuries can be created by changing the characteristics of the linear and angular acceleration loading curve (Gennarelli et al., 1979; Adams et al., 1981; Kleven 2003). To reduce the risk of brain injuries, helmets work by changing the characteristics of the linear acceleration curve. Specifically, helmets lengthen the duration of the impact and lower the magnitude in an effort to prevent severe brain trauma.

To prevent these injuries from occurring, researchers attempt to establish a threshold whereby the damage would occur. Currently, the engineering parameter used to predict the extent and severity of these injuries is peak resultant linear acceleration. This is reflected in the use of peak linear acceleration for the certification of safety devices such as helmets (Hoshizaki and Brien 2004). While the use of peak resultant linear acceleration has been successful in reducing the prevalence of traumatic brain injury (TBI) in sport, mild traumatic brain injury (mTBI) remains prevalent (Wennberg and Tator, 2003). This suggests that linear acceleration alone is not capturing the spectrum of brain trauma.

Recent research has further demonstrated the limitations of using peak resultant linear acceleration to characterize the protective capabilities of football and hockey helmets (Post et al., 2009; 2010). Using a Hybrid III headform and neck Post et al. (2010) impacted helmets in a series of centre of gravity and off axis impacts. From this research they showed that for certain locations similar values for linear acceleration can be achieved with differing rotational accelerations. When the linear and rotational acceleration outputs were used for a finite element model analysis it was discovered that low linear acceleration values when high rotational acceleration was present can produce potentially injurious levels of maximum principal strain when compared with cadaver tolerance data. This research and others suggests that it may be prudent to consider the characteristics of the linear and rotational acceleration loading curves such as total duration and not just peak values when determining risk of injury from injurious brain deformations (Yoganandan et al., 2008; Post et al., 2010). Further research has been conducted using FEM to identify possible thresholds of injury for brain tissue deformation through reconstructions of real world impacts with promising results (Willinger and Baumgartner, 2003; Zhang et al., 2004; Kleiven, 2007). However, the researchers were focused on identifying peak variables and did not consider how the input loading curve characteristics and loading axis influenced brain deformation.

Improvements in computer modeling have allowed the use of finite element analysis as a tool for predicting head injuries and subsequently improve the design of protective equipment (Zhang et al., 2001; King et al., 2003). Finite element modeling allows for the measurement of brain deformation of an impact through the 3 vectors of linear and 3 vectors of angular acceleration over time for a full representation of the event. This allows for the inclusion of direction and magnitude as well as other loading curve characteristics as influencing factors of brain deformation (Horgan and Gilchrist, 2003).

The purpose of this research was to investigate the influence of linear and angular acceleration loading curve shape on the location and severity of brain tissue deformation. This research has applications to not only protective technology design but also in understanding how the model used in this study responds to controlled inputs. Currently protective technologies and helmets in particular are certified against linear acceleration when brain injury has been shown to be influenced by both linear and angular acceleration. Finite element modelling allows for the interpretation of linear and angular acceleration loading curves and how they influence brain deformation. This unique ability of finite element modelling will likely be used to inform manufacturers how to design helmets to be more protective for a wider range of brain injuries in the future. As a result it is essential that there is a fully understanding of how the finite element model in use is influenced by aspects of the loading curves used as input. The current study uses 3 loading curves which were taken from helmet impacts to examine how the characteristics of those loading curves influenced the resulting brain deformation. The end result could hopefully inform researchers as to how to minimize the production of damaging brain deformation by avoiding certain acceleration loading curve shapes when designing helmets for various sports.

2. Methodology

2.1 Finite element model

The finite element model used in this research was the University College Dublin Brain Trauma Model (UCDBTM) (Horgan and Gilchrist, 2003; Horgan and Gilchrist, 2004). The geometry of the model was from a male cadaver as determined by Computed Tomography (CT), Magnetic Resonance Tomography (MRT) and sliced contour photographs (Horgan and Gilchrist, 2003). The head and brain are comprised of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem (table 1, 2). This version of the UCDBTM does not include elements representing the cerebral blood vessels.

The model was validated against intracranial pressure data from Nahum et al. (1977) cadaver impact tests and brain motion against Hardy et al.'s (2001) research. Further validations accomplished comparing real world brain injury events to the model reconstructions with good agreement (Horgan and Gilchrist, 2003).

A linearly viscoelastic material model combined with a large deformation theory was chosen to model the brain tissue. The behaviour of this tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus (Horgan and Gilchrist, 2003). The compressive behaviour of the brain was considered elastic. The shear characteristics of the viscoelastic behaviour were expressed through the formula:

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

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor (Horgan and Gilchrist, 2003). The hyperelastic material model used for the brain in shear was expressed as:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)}$$

where C_{10} and C_{01} are temperature-dependent material parameters, and t is in seconds (Horgan, 2005).

The CSF layer was modelled using solid elements with a low shear modulus and a sliding boundary condition between the interfaces of the skull, CSF and brain was used. The model is comprised of 7,318 hexahedral elements representing the brain, and 2,874 hexahedral elements representing the CSF layer (Horgan and Gilchrist, 2003).

2.2 Input loading curves

To evaluate the effects of the loading curve shape and applied vector on brain deformation three sample loading curves which represent shapes which are commonly produced in helmet testing were created, all with equivalent areas (figure 1, 2). Loading curve A, which had a late 6015 rad/s^2 / 150 g peak at 4.4ms meant to simulate a soft material response. This loading curve would represent a situation where the material was too soft to manage the energy of the impact and would produce a late spike in the resulting linear and rotational acceleration. Loading curve B, which had an early 6015 rad/s^2 / 150 g peak at 0.9ms to simulate a stiff material response. This loading curve would represent a situation where the material was too dense and did not absorb any of the energy of the impact, producing an early spike in linear and rotational acceleration. Finally, loading curve C which had a lower 2215 rad/s^2 / 55 g peak at 0.5ms, to simulate a plateau curve. This loading curve represents what would be considered an ‘ideal’ response for an energy absorbing material in helmet design, where there is a quick rise to a low plateau where the energy is managed over a longer time without bottoming out.

The same loading curve was modified into m/s^2 for the linear inputs. All three loading curve types were applied to the UCDBTM centre of gravity individually in the X, Y and Z axes. The values were meant to represent the possible peak range to create an mTBI as reported by Schreiber et al. (1997), while the plateau (loading curve C) is below known traumatic brain injury thresholds but consistent in total area with loading curves A and B. Each of the three loading curves was run independently using ABAQUS 6.7 finite element solver through the three axes of linear and angular acceleration. The X-axis is facing forward from the head CG, the Y-axis to the left of the head and the Z-axis vertically upwards.

The resulting brain deformation will be measured by commonly used brain deformation metrics which have been proposed to have a correlation to brain injury (Zhang et al., 2004; Kleiven 2007). In this study the brain deformation metrics chosen were maximum principal strain and Von Mises stress.

3. Results

3.1 Maximum principal strain

Overall, the maximum principal strain results indicated that loading curve A produced the highest values (table 3). For the angular acceleration inputs, loading curve A produced the highest strains in the grey matter for the X axis (0.119), white matter for the Y axis (0.143) and grey matter for the Z axis (0.116). For linear acceleration inputs, loading curve A produced the highest values of strain in the brain stem for the X axis (0.164) and in the white matter for the Z axis (0.280). For the Y axis, loading curve B produced the highest strain in the brain stem (0.124). The lowest values for maximum principal strain were commonly produced by loading curve C and were found in the cerebellum for angular and linear inputs (0.013 – 0.038), except for the linear Z axis input where the lowest value was in the grey matter (0.039).

3.2 Von Mises stress

The Von Mises Stress results indicated that loading curve A produced the highest stress values (table 3). For the angular acceleration inputs, loading curve A produced the highest strains in the grey matter for the X axis (5,796 Pa), white matter for the Y axis (6,646 Pa) and in the grey matter for the Z axis (5,650 Pa). For the linear acceleration inputs, loading curve A also produced the highest Von Mises Stress values in the white matter for the X axis (7,107 Pa) and in the brain stem for the Y axis (6,363 Pa). The grey matter was the location of highest stress for the Z axis linear acceleration (11,730 Pa). The lowest values were produced by loading curve C for both linear and angular acceleration inputs and were commonly found in the cerebellum (578 – 1791 Pa), except for linear input in the Z axis which was found in the grey matter (1,612 Pa).

3.3 Maximum principal strain and von Mises stress curves

Figures 2 and 3 show the maximum values of maximum principal strain and Von Mises stress over time. The elements experiencing the largest deformation changes over the course of the simulation and the values reported here are for the elements experiencing the largest magnitude of deformation. The values are from the single element in the model which incurred the highest magnitude of deformation over the duration of the simulation. The X, Y and Z axis curves of the stress and strain response to the linear and angular acceleration inputs show that the time to peak for loading curve B is shorter than that for the other loading two curve types (figure 2 and 3). Areas of high maximum principal strain were associated with corresponding areas of large Von Mises stress except for linear X and Y loading inputs, where the peak values in white matter and brain stem were close. The peak values for maximum principal strain and Von Mises stress produced by curve B was consistently lower than those for loading curve A. The linear acceleration inputs produced the largest values overall for maximum principal strain and Von Mises stress for the X and Z inputs, with angular acceleration influencing the Y axis responses.

3.4 Response by location

The different loading curve (A, B and C) results were compared across brain region to analyze the influence of the type (linear or angular) and direction of loading curve input (x, y or z) on the magnitude of deformation in different locations of brain model. The loading curve results were ranked per region of the brain to examine which location incurred the largest deformation for each type of loading curve input. The cerebellum consistently showed the lowest maximum principal strain and Von Mises stress response for all inputs. The X axis linear acceleration loading curves produced the highest values in the white matter and brain stem regions. The X axis angular acceleration response produced high stress and strain response exclusively in the grey matter region. The Y axis linear acceleration produced highest values in the brain stem, with the Y axis angular acceleration producing the highest values in the white matter. The Z axis linear acceleration inputs produced the highest values in the brain stem. The Z axis angular acceleration loading curves produced the largest values of stress and strain in the grey matter.

4. Discussion

The purpose of this study was to investigate the effects of loading curve shape and direction of the loading on brain tissue stress and strain using the University College

Dublin Brain Trauma Model (Horgan and Gilchrist, 2004). The research in this area has been limited to animal models that lacked the control that can be achieved using finite element modeling. Recent studies have identified low correlation between linear acceleration and angular acceleration peak values with measured brain injury parameters (Post et al., 2010). This study was undertaken to better understand the relationship between the loading curve shape of linear and angular acceleration and maximum principal strain and Von Mises stress for brain tissue.

4.1 Location and Direction

The simulations in this study isolated each direction and type of acceleration in an attempt to elucidate differences in location of peak stress and strain. Upon analysis of the results the shape of the loading curve influenced the peak value, time to peak and the location of strain and stress in the brain as defined by the UCDBTM. When loading curve direction is analyzed independently, the regions of highest stress and strain also vary, with the angular acceleration producing high values in either the grey matter for the X and Z axis rotations or white matter for the Y axis rotation. The linear acceleration inputs in the X, Y and Z direction consistently produced the highest values in the brain stem region. From a modelling perspective this difference is likely a result of the interaction of the tissue material characteristics with the different types of loading curve inputs. The fact that the material behaves differently under linear and angular loading curve inputs is an example of this. Different regions of brain tissue would respond differently to these inputs due to the characteristics defining those regions such as material characteristics in compression, shear and tension as well as geometry and boundary conditions. An example of these complex interactions is shown by the Y axis linear acceleration producing high values in the brain stem and Y axis angular acceleration in the white matter. These are fundamentally different motions and as a result the model produces different responses in different brain regions. Also the grey matter showed high strains for the X and Z axis angular acceleration and white matter for the Y axis angular acceleration. From a biological perspective the grey matter would be less sensitive to directional inputs as it is thought to be comprised of non-directional tissues. The white matter is comprised of directional axonal tracts and would be more susceptible to shearing accelerations. These biological differences are represented in the UCDBTM by a difference in shear characteristics; however the model does not consider directionality of tissue in its current format. These differences are likely a reflection of the interaction of the loading curve input characteristics and the complex interactions between the tissue material characteristics and loading conditions within the model. Within the literature, this variation in level of response to direction is supported by Kleiven (2003), who demonstrated a directional dependence of subdural hematoma using finite element modeling approaches, as did Bradshaw and Morfey (2001) for diffuse axonal injury. Using animal studies Gennarelli et al. (1979) and Adams et al. (1981) also demonstrated that direction influences the area and type of injury.

4.2 Loading curve shape and peak value

The loading curve A condition consistently produced the highest values of maximum principal strain and Von Mises stress for all conditions. Loading curve C produced the lowest values, which were commonly located in the cerebellum. While loading curve A and B had identical peak values; loading curve C had a lower peak which explains the lower response associated with that condition. Loading curve A had a longer time to peak than loading curve B which may explain the resulting large stress and strain values. The maximum principal strain and Von Mises stress output curve shapes for the elements that produced the peak values in the brain have similar profiles to the linear or angular loading curve shapes. Loading curve C output showed a slower rise to peak time without any noticeable plateau producing the lower stress and strain curves. This would suggest a relationship between the loading curve shape and the output curve shape of maximum principal strain and Von Mises stress. The largest values in maximum principal strain and Von Mises Stress were influenced by the linear accelerations and not the rotational accelerations which are in disagreement with research investigating these influences. It is possible that the relative level of linear acceleration (150 g) was higher than that of the rotational acceleration (6,015 rad/s^2) producing larger response in maximum principal strain and Von Mises stress. It is difficult to use equivalent linear and rotational acceleration inputs as they produce different motions and mechanisms to the same injury. Also human variation makes it difficult to choose equivalent magnitudes of these parameters as subdural hematoma has been produced at 4,500 rad/s^2 (Lovenhielm, 1974) and at 130 g (Willinger and Baumgartner, 2003). In hybrid III impacts a value of 140 g can create a rotational component of 12,000 rad/s^2 (Rousseau and Hoshizaki, 2009; Rousseau et al., 2009), indicating comparing 150 g to 6,015 rad/s^2 may be an unrealistic scenario in terms of brain tissue response. An ideal solution to these difficulties in analyzing linear and acceleration loading curves would be to have acceleration loading curves that produced a head injury.

The high values and location of maximum principal strain in this research were found in the same location as the highest values of Von Mises stress. This would imply that the two are correlated and perhaps it would be redundant to use both in brain tissue deformation. This finding warrants further research. The response shown by this model supports the conclusions of Yoganandan et al.'s (2008) research on biphasic acceleration deceleration pulses using a 2 dimensional finite element model. They reported strains that were produced were region and pulse shape specific.

The purpose of this research was to analyze how different loading curve shapes could influence the production of maximum principal strain and Von Mises stress within the University College Dublin Brain Trauma Model. The peaks of these loading curves are commonly used as predictors of injury for protective technologies such as helmets. However it has been indicated by authors that protective technologies must be developed according to brain deformation and not peak linear or angular acceleration (King et al., 2003). This would allow for the interaction between the acceleration loading curves generated at impact and the brain tissue to be accounted for. If a better understanding of these loading curves could be gained, as shown in this research, it could be used to lower the potentially injurious brain deformations incurred from an impact by designing protective devices to limit those types of loading curves and types of acceleration. Future research in this area should focus on the limitations of the modelling and upon attaining further understanding of how loading curves from an impact influence the brain deformations and brain injury metrics which have been suggested to be associated with injuries such as traumatic brain injury and concussion. While this study investigated only a few loading curve characteristics, further studies are warranted using impact loading curves which resulted in an identifiable brain injury to investigate how these parameters influence maximum principal strain.

Limitations of this study revolve around the parameters used to define the University College Dublin Brain Trauma Model (UCDBTM). The finite element model (UCDBTM) is an approximation intended to simulate human responses to injurious loading. The material characteristics, constitutive properties assigned to the cerebral elements are taken from cadavers and as such may not represent the true nature of central nervous system tissue. Also the inclusion of cerebral blood vessels in the UCDBTM would likely result in different magnitudes of stress and strain from this analysis (Ho and Kleiven, 2007). In addition this model uses a hyperelastic material model for the brain tissue in shear, however the results would likely change depending on the material law used or if linear elasticity was used instead. The influence of these changes for this particular model has yet to be quantified. The geometry of the model is taken from CT and MRI images from a single subject. The current mesh density used in this model was optimized for validity of results compared to computational time. A change in the density of the mesh used would provide different results, although the exact magnitude of those differences has not been quantified. While validations for finite element models of the human brain in general have been conducted, a complete verification of its predictions for human response has not been accomplished for all parts of the brain. As a result the deformation of the brain represented by the simulation can only be taken as an estimate and not an absolute.

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