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Analysis of the influence of independent variables used for reconstruction of a traumatic brain injury incident

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Abstract

Traumatic brain injuries contribute to a high degree of morbidity and mortality in society. To study traumatic brain injuries researchers reconstruct the event using both physical and finite element models. The purpose of these reconstructions is to correlate the brain deformation metric to the type of injury as a measure for prediction. These reconstructions are guided by a series of independent variables which all have influence upon the outcome variables. This research uses a combination of physical and finite element modelling to quantify how independent variables such as velocity and impact vector (angle) contributes to the resulting variance in brain deformation metrics. The results indicate that increased velocity increases the magnitude of the response. Also, the impact angle influences the response depending on brain deformation metric chosen as predictor variable. This research demonstrates the need to account for a wide variety of independent variables when engaging in reconstructive research,

Keywords

Finite element modelling, Traumatic brain injury, Reconstruction, Injury biomechanics, Subdural hematoma

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Introduction

Traumatic brain injuries such as subdural hematoma and traumatic axonal injury contribute to a high degree of morbidity and mortality in society[1,2,3]. As a direct result of the high social and personal costs of these types of injury a great deal of research has gone into their prediction in an effort to prevent their occurrence. Finite element modelling in conjunction with brain injury accident reconstructions has been employed by researchers to better understand traumatic brain injury[4,5,6]. Often, the purpose of these reconstructions is to correlate brain deformation based dependent variables to the type of traumatic brain injury.

Much of the reconstructive research has been conducted in sporting situations because it provides a semi controlled environment for analysis where head injuries are relatively common. Zhang et al[5] used data from American National Football League impacts producing concussion to reconstruct concussive injuries using their brain model. Using a combination of a Hybrid III head form, impact system and finite element modelling they proposed brain injury thresholds using maximum principal strain, intracranial pressure and strain rate amongst others. Other researchers have used a similar data set and have suggested the same variables although with varying magnitudes[4,6]. While this research provides an excellent avenue for future research involving thresholds and dependent variables associated with brain injury they have their limitations. Newman et al[7] described the data from the American National Football League reconstructions as having error up to 25% based upon the kinematic video data alone. This error was suggested to reside in incorrect impact vector identification and inbound velocity approximations. The error would be compounded further by the finite element modelling process used to calculate brain deformation metrics. These studies used similar pools of data, and many of the differences and similarities reflect the head models used.

Advancements in methodology using medical imaging to define brain lesion site for FE analysis was developed by Kleiven[6] and Doorly and Gilchrist[8]. This type of methodology allowed for specific regions of damaged brain tissue to be examined for the local magnitudes of brain deformation metric. While considered a refinement of the finite element analysis of the injured region, limitations still existed in accurately defining the vector of the impact, impact mass and low sample sizes from which to develop a statistically significant result.

Investigations into impact vector using Hybrid III dummy systems have been conducted by several researchers[9]. This research has shown that for a variety of impact vector conditions centred around the same impact site can produce different linear and rotational acceleration loading curves. These variations would produce different brain deformation locations and magnitudes depending on whether the impact produces linearly or rotationally dominant loading curves[10,11,12,13]. This research has also been conducted on helmeted Hybrid III head forms with similar variation due to impact vector[14,15,16,17].

It is apparent from this research that there are many difficulties to correctly reconstructing a brain injury scenario. Any impact to the head can be a product of the inbound velocity, mass and location. All of these things when varied could produce completely different loading curves and therefore create a different mechanism of injury and location of damage to the brain. While these studies do acknowledge these limitations, the quantification of how these parameters affect the brain deformation has yet to be undertaken.

The purpose of this research is to reconstruct a traumatic brain injury event using a combination of hybrid III head form, mechanical impact system, FE modelling and medical imaging to examine brain deformation based metrics which may be linked to the injury. A secondary purpose is to examine how the various impact conditions produced in this reconstruction, specifically velocity and impact angle, influence the brain deformation dependent variables.

Methodology

The participant in this case was 85 years old and had received a subdural hematoma from a fall without the presence of skull fracture. The subdural hematoma presented itself on the left frontal area of the brain. Clinical assessment of the brain injury was conducted by a neurosurgeon on site and corresponding CT scan along with injury reports were used to define the initial conditions for the reconstructive protocols.

According to the subject and confirmed by a witnesses, the 85 year old man fell head first into concrete with the contact site being to the right side of the frontal bone. The subject did not experience a loss of consciousness (assessed at a GCS of 15) and filled out a reconstructive report form describing the incident. The information on the form was further validated by a description of the incident by the subject's wife. Initial CT scans showed a large hematoma over the impact site (Figure 1). A CT scan three weeks later showed a subdural hematoma in his left frontal lobe (Figure 2).

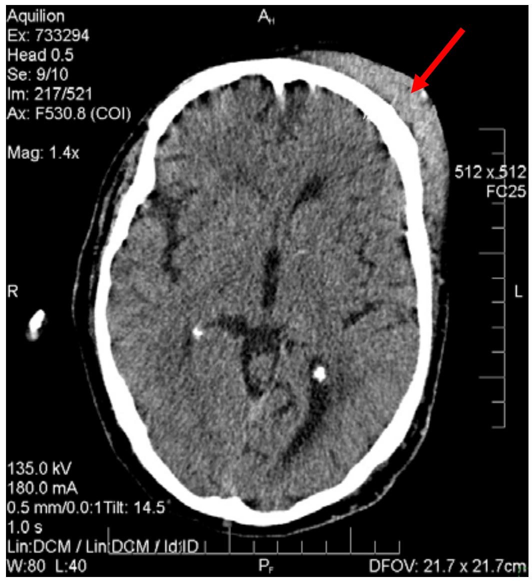

Figure 1. Surface hematoma indicating impact site (indicated by red arrow)

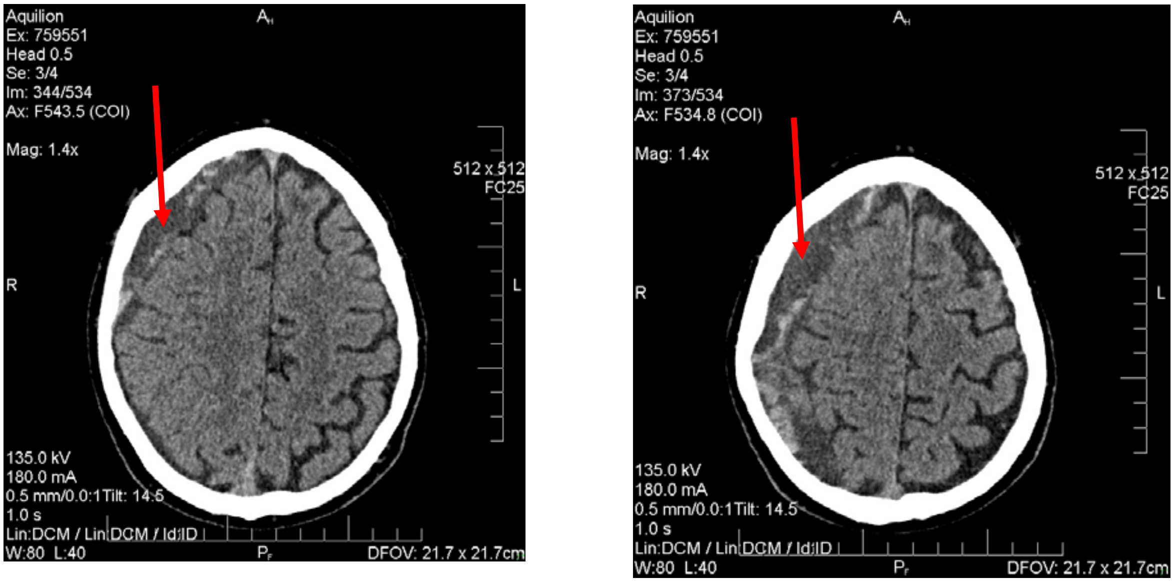

Figure 2. Pre-operative CT scans of the 85 year old subject at two different levels with subdural hematoma evident in the upper left of the image (indicated by red arrow)

As this incident was a falling injury, the reconstruction was conducted using a monorail device with a 50th percentile Hybrid III instrumented with a 3-2-2-2 accelerometer (Endevco 7264C-2KTZ-2-300) array[18]. The accelerometers were sampled at 20 kHz and the resulting loading curves were filtered with a 1000 Hz Butterworth filter. The impact velocity was calculated based upon the subject's falling height. The impact conditions were a velocity of 4.5 m/s +/- 0.5 m/s and angle of impact variations of 0, 12 and 24 degrees as depicted in Figure 3.

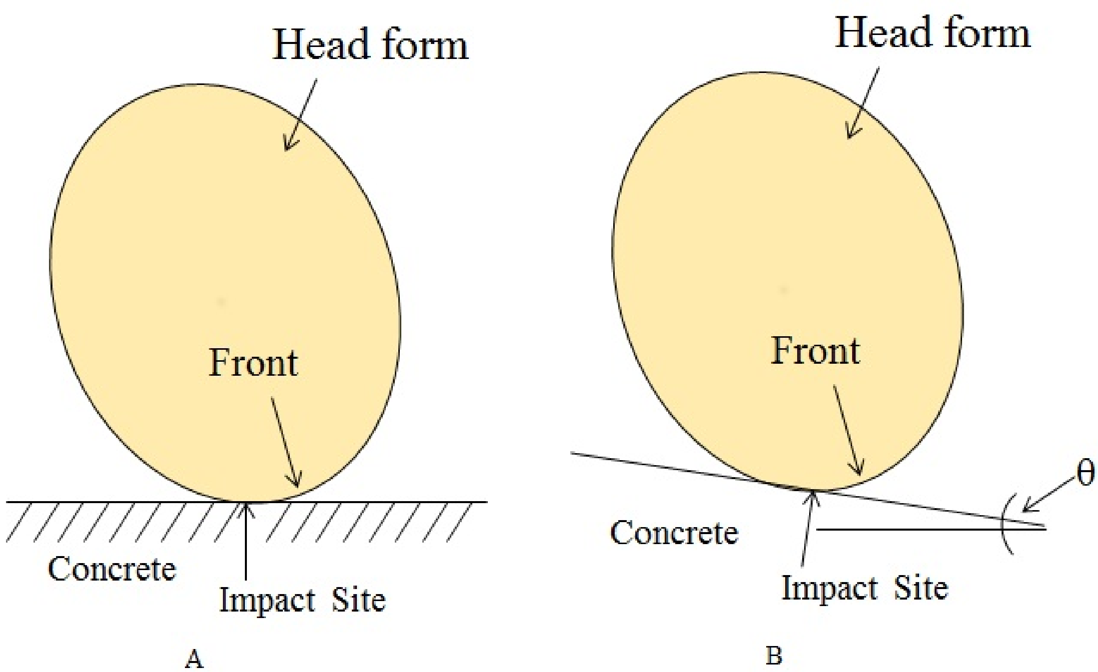

Figure 3. Diagram showing the angle (θ) as calculated for the reconstruction

The angle variations were accomplished by tilting the concrete block to achieve the appropriate variations (Figure 4). These conditions were chosen to produce a corridor of response through which the influence of the independent variables on the brain deformation metrics could be examined. The material was composed of concrete similar to the original impact surface. The impact location on the Hybrid III was chosen to mimic the impact site as shown on the initial CT scan.

Figure 4. Simulated injury area impact using a Hybrd III head form and neck form

The finite element model used for the reconstruction was the University College Dublin Brain Trauma Model (UCDBTM)[19,20]. The model geometries were determined through medical imaging of a male cadaver. The head was comprised of the scalp, skull, pia, falx, tentorium, cerebrospinal fluid, grey and white matter, cerebellum and brain stem. In total, the model had approximately 26,000 elements[19]. The validation of the model was found to be in good agreement with intracranial pressure and brain motion data taken from cadaver experiments[21,22]. In this reconstruction the UCDBTM was scaled to the head size of the subject, and extra element sets were created to represent the area affected by the subdural hematoma and the bridging veins local to the injury site. The subdural hematoma area was defined by matching the CT scan images overtop of the FE model images and selecting the appropriate elements to represent the volume.

The material properties of the model were derived from anatomical testing (Table 1). A linearly viscoelastic material model was used to define the brain tissue. The brain tissue behaviour was characterized as viscoelastic in shear and the compressive behaviour as elastic. A sliding boundary between the CSF and brain was used to mimic the brain skull interface.

Table 1. 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
Bridging veins	9.5	-	-

The loading curves from the physical reconstructions were input at the centre of gravity of the model and brain deformations calculated. Results in von Mises stress, maximum principal strain, strain rate, product of strain and strain rate were taken from the subdural hematoma region identified by CT scan as well as global peaks from the cerebellum element set of the model.

Results

The results for the reconstruction are shown in tables 2, 3 and 4. Table 2 shows the values for the entire cerebrum as a result of the reconstructive impacts. Table 3 shows the kinematic input and brain deformation response for the subdural region and table 4shows the values for the bridging veins closest to the injury site.

Table 2. The brain deformation metric response for the cerebrum

	4.0 m/s			4.5 m/s			5.0 m/s		
	0°	12°	24°	0°	12°	24°	0°	12°	24°
Linear acceleration (g)	408 (12.5)	346 (16.9)	285 (14.7)	506 (9.62)	425 (10.1)	362 (4.47)	601 (12.1)	490 (17.7)	433 (13.4)
Rotational acceleration (krad/s^2)	29.2 (1.54)	29.6 (1.72)	28.2 (2.04)	37.0 (0.52)	36.4 (0.79)	36.0 (0.44)	40.8 (0.20)	42.1 (1.78)	43.2 (0.90)
Maximum principal Strain	0.75 (0.04)	0.64 (0.06)	0.73 (0.04)	0.98 (0.09)	0.81 (0.09)	0.79 (0.05)	1.30 (0.02)	0.85 (0.02)	0.92 (0.02)
von Mises Stress (kPa)	25.2 (1.97)	19.9 (3.15)	22.0 (1.87)	34.2 (2.37)	26.3 (2.36)	27.2 (0.68)	44.5 (0.38)	28.2 (0.85)	29.8 (0.37)
Strain rate (s^{-1})	123 (15.2)	79.6 (33.5)	72.8 (3.79)	224 (112)	125 (17.3)	118 (32.7)	373 (5.99)	171 (58.7)	112 (33.1)
Strain X Strain rate	92.0 (7.90)	52.2 (27.4)	53.1 (5.52)	223 (124)	101 (24.3)	92.8 (20.4)	486 (15.6)	144 (47.3)	103 (28.3)

Table 3. The brain deformation metric responses for the subdural hematoma region

	4.0 m/s			4.5 m/s			5.0 m/s		
	0°	12°	24°	0°	12°	24°	0°	12°	24°
Linear acceleration (g)	408 (12.5)	346 (16.9)	285 (14.7)	506 (9.62)	425 (10.1)	362 (4.47)	601 (12.1)	490 (17.7)	433 (13.4)
Rotational acceleration (krad/s^2)	29.2 (1.54)	29.6 (1.72)	28.2 (2.04)	37.0 (0.52)	36.4 (0.79)	36.0 (0.44)	40.8 (0.20)	42.1 (1.78)	43.2 (0.90)
Maximum principal Strain	0.29 (0.06)	0.26 (0.002)	0.27 (0.01)	0.36 (0.09)	0.34 (0.03)	0.32 (0.001)	0.33 (0.03)	0.36 (0.02)	0.38 (0.004)
von Mises Stress (kPa)	9.73 (0.84)	9.3 (0.31)	9.1 (0.52)	11.6 (2.36)	10.9 (0.68)	10.9 (0.16)	11.5 (1.68)	11.7 (0.45)	12.5 (0.10)
Strain rate (s^{-1})	51.7 (9.02)	73.5 (0.74)	77.7 (2.74)	63.4 (8.71)	95.8 (7.29)	91.9 (0.26)	52.9 (18.2)	107 (4.53)	110 (1.38)
Strain X Strain rate	14.8 (3.70)	18.9 (0.38)	21.1 (1.48)	22.8 (7.80)	32.2 (4.99)	29.5 (0.17)	17.2 (4.20)	38.1 (2.65)	42.0 (1.06)

Table 4. The brain deformation metric responses for the bridging veins closest to the subdural hematoma region

	4.0 m/s			4.5 m/s			5.0 m/s		
	0°	12°	24°	0°	12°	24°	0°	12°	24°
Linear acceleration (g)	408 (12.5)	346 (16.9)	285 (14.7)	506 (9.62)	425 (10.1)	362 (4.47)	601 (12.1)	490 (17.7)	433 (13.4)
Rotational acceleration (krad/s^2)	29.2 (1.54)	29.6 (1.72)	28.2 (2.04)	37.0 (0.52)	36.4 (0.79)	36.0 (0.44)	40.8 (0.20)	42.1 (1.78)	43.2 (0.90)
Maximum principal Strain	0.16 (0.01)	0.11 (0.004)	0.10 (0.002)	0.16 (0.02)	0.12 (0.02)	0.10 (0.001)	0.23 (0.02)	0.14 (0.004)	0.12 (0.01)
Strain rate (s^{-1})	25.0 (1.43)	17.5 (1.12)	17.5 (0.31)	25.1 (2.82)	19.1 (2.55)	18.6 (0.77)	20.5 (2.82)	22.4 (1.19)	18.8 (1.12)
Strain X Strain rate	4.09 (0.46)	1.87 (0.08)	1.68 (0.06)	4.12 (0.89)	2.39 (0.64)	1.86 (0.25)	2.63 (0.80)	3.08 (0.23)	2.17 (0.18)

The results produced a range of dynamic responses across the three velocities of 285 to 601 g and 28.2 to 43.2 krad/s^2. The peak resultant linear acceleration decreases when the impact angle is increased, however the peak resultant rotational acceleration shows no significant change across angle ($\alpha = 0.05$). When examining the results across velocity, the cerebrum showed an increase in brain deformation metrics as velocity increased for 0˚ ($\alpha = 0.05$). For the 12˚ condition, the MPS and VMS increased from 4.0 m/s to 5.0 m/s, however the 4.5 m/s velocity results were not different form the 5.0 m/s results. At 24˚ the cerebellar values increased across velocity for MPS. Von Mises stress values were smallest at 4.0 m/s and largest for 5.0 m/s, but 4.5 m/s values were not different from 5.0 m/s. For the subdural site there is no significant increase in brain deformation metric across the 0˚ condition for velocity. At the 12˚ condition, the magnitudes at 4.0 m/s were smaller than 5.0 m/s for all metrics. However, the 4.5 m/s condition was not different in magnitude for the 5.0 m/s condition. For the 24˚ condition there was an increase in all metrics across velocity. For the bridging veins there were no significant differences in the results at 0˚. For the 12˚ condition the values for the brain deformation metrics were smaller at 4.0 m/s than for 5.0

m/s. When comparing 4.5 m/s to 5.0 m/s there is no difference in the values ($\alpha = 0.05$). This pattern is followed in the 24˚ condition for the bridging vein location.

When the results are examined by angle of impact, there is also variance showing an interaction between impact angle and resulting dynamic response and brain deformation metric. At 4.0 m/s the linear acceleration magnitudes decreased significantly across angle, but not for rotational acceleration ($\alpha = 0.05$). The cerebrum site showed no difference in any brain deformation metric across angle. The subdural site showed no difference in maximum principal strain and von Mises stress. However there were differences in strain rate with the 0° condition values being lower than the 12° and 24° conditions ($\alpha = 0.05$), but 12° and 24° were not significantly different. Product of strain and strain rate at this site had lower values at 0° and 24° with values of 14.8 and 21.1 respectively ($\alpha = 0.05$). The bridging veins showed differences in all brain deformation metrics with 0° producing higher magnitudes than 12° and 24° for all the variables.

The results for the 4.5 m/s condition produced results similar to the 4.0 m/s condition with some variations. The values all increased as the velocity increased. For the cerebrum, the MPS, strain rate and product of strain and strain rate were not different across angle. The VMS values at 0° were larger than the values at 12° and 24° with 34.2 kPa and 26.3, 27.2 kPa respectively. At the subdural site the values for MPS, VMS, and product of strain and strain rate were not different across angle. The values for strain rate were larger for strain rate at the 0° condition (224) than for the 12° (125) or 24° (118) impact conditions. When examining the bridging vein values the MPS, VMS strain rate and product of strain rate all showed similar results. The 0° condition was consistently larger than the 12° and 24° conditions ($\alpha = 0.05$).

The results for the 5.0 m/s condition were all larger than the previous two velocities. At the cerebrum location, the MPS and VMS values were largest for the 0° condition, followed by the 24° condition, and finally the 12° condition ($\alpha = 0.05$). For strain rate and product of strain and strain rate, the 0° condition produced larger magnitudes than the 12° or 24° conditions ($\alpha = 0.05$). The subdural site showed no difference between angle conditions for MPS and VMS variables. Strain rate had very low values (52.9 s^{-1}) at the 0° condition, followed by 107 s^{-1} and 110 s^{-1} for the 12° and 24° conditions respectively. A similar trend was evidenced by the product of strain and strain rate, with similarly low values for the 0° condition and larger values for the 12° and 24° condition. The bridging veins site at 5.0 m/s did not show any significant differences between the brain deformation variables for any impact angle condition.

Discussion

The results of this reconstruction suggest that for this case the subdural hematoma was produced at a range between 26 to 38% maximum principal strain, which is somewhat higher than those values found in the literature[8]. The values found for this reconstruction for the overall brain response (cerebrum) was also found to be largely in excess of the current values for human reconstruction that exists in literature (130%).

The dynamic response and resulting brain deformation as a result of the different impact conditions were found to vary. The change in velocity as expected tended to produce a more severe impact. At the cerebellar and subdural hematoma regions the 4.0 m/s impact velocity tended to produce the lowest magnitudes in brain deformation response. However the 4.5 m/s and 5.0 m/s brain deformation metrics were commonly not significantly different. This might indicate that changing the velocity by 0.5 m/s is not enough to induce a change in the response which would be reflected in the brain tissue deformations. It is also possible that this form of physical modelling leads to a saturation point were the values do not increase appreciably past a certain impact energy. The bridging vein location did not show any significant increase in brain deformation metric across velocity at the 0˚ impact condition, but did show changes at the 12˚ and 24˚. This region was modelled differently, with beam elements instead of hexahedral brick elements which may contribute to this phenomenon.

Changes in impact angle produced corresponding decreases in linear acceleration. Notably, the rotational acceleration component did not change. It is possible that this is due to a coupling contact phenomenon where the vector is far enough outside the centre of gravity that would result in low linear accelerations, and due to the neck condition of the Hybrid III head form also reduced rotation upon impact. Previous work has shown that relatively minor changes in impact angle can change the kinematic responses of the head form in linear and rotational acceleration[9,23]. These studies were conducted using a system which allows more free movement of the head form post-impact. These results indicating no change in rotational acceleration across impact angle using a monorail methodology might contribute to the lack of difference amongst the rotational accelerations. This is likely due to the fixed neck condition and no movement of the head form post-impact and a complete transfer of energy. Ordinarily in a more free system when an impact vector is outside the centre of gravity the rotations increase until there is not enough contact to cause the head form to rotate[23]. These types of changes produced unique loading conditions which influences the location, type and severity of the resulting response, be it measured in kinematic or brain deformation variables.

The brain metric deformations were also affected by changes in impact angle condition. Each site analysed showed different magnitudes of response. Also, as can be seen in the cerebrum response at 4.0 m/s, there is no difference in any brain deformation variable, whereas the 0° condition produced larger magnitudes in the bridging veins and lower values in the subdural site. The influence of angle on the resulting response shows that the 0° condition typically produces a lower magnitude strain rate and product of strain and strain rate response in the subdural site and the bridging vein site than the 12° and 24° condition; however this reverses for the cerebrum at 5.0 m/s. These results would infer that there is a velocity/angle interaction for the cerebrum at the 5.0 m/s condition, where previously non-significant variables become different from each other.

When the brain deformation independent variables are examined for the subdural and bridging vein sites certain relationships appear. The maximum principal strain and von Mises stress parameters for the subdural site show no difference across angle however the strain rate shows differences between 0° and the other angle condition. A similar relationship exists for the bridging veins at 4.0 and 4.5 m/s, where strain rate and product of strain and strain rate find differences between the variables. This relationship could suggest that these deformation metrics are more sensitive to changes in angle than using maximum principal strain and von Mises stress alone. The results also indicated that the linear acceleration was influenced by the change in angle however the brain deformation metrics produced from the linear and rotational acceleration loading curves were not necessarily significantly different. These results would suggest that the peak magnitudes of these variables may not be the only factor in the production of these brain deformation metrics. Also, when comparing the subdural site at 4.5 m/s the linear acceleration magnitudes are different however the maximum principal strain and von Mises stress magnitudes are the same, 32 – 34% and 10.0 – 11.6 kPa across angle. This supports research showing peak kinematic variable may not predict the magnitude of brain deformation response[15,17].

This reconstruction differs from some of the previous TBI reconstruction research as it employs the use of physical models as well as finite element. In the case of physical models, the head is more rigid than most computational models under impact, which may influence the results. The impacting mass can also be difficult to ascertain from injury reports, though in this case the weight of the head itself was likely the impacting mass. Other factors that would influence the dynamic response data would also be the type of impacting surface used (compliance of the system) and the mechanism of injury which is being reconstructed (head impacting object vs object impacting the head). This particular reconstruction shows a corridor of response which may reasonably represent the conditions that caused the injury. However, the loading curves created with the Hybrid III head form are not the same as those created when hitting a living head and thus the brain deformation metrics when calculated by applying the loading curves from this source would reflect error. This is one of the primary reasons to use a corridor of response type accident reconstruction, where the impact characteristics used should account for a wide variety of possibilities and perhaps an average with upper and lower boundaries to represent the injury would be more appropriate. This type of corridor would require variations in impact vector, velocity and mass to fully cover all possibilities. In this study only impact vector and velocity were used as corridor variables, however in further studies the influence of mass as well may have be considered.

While this study focuses primarily on the difficulties of physical model reconstructions, there are further limitations as is evidenced through the use of finite element models. Finite element models are best approximations and not perfect representations of human physiology. In the case of this research, the results are specific to the UCDBTM's material characteristics and unique boundary conditions. Also, in this case the unique physiology of the elderly brain was not reflected by the UCDBTM, although the model was adjusted to size.

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