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The characteristics of traumatic brain injury

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Abstract

Traumatic brain injury (TBI) is a common injury and is a leading cause of morbidity and mortality throughout the world. Research has been undertaken in order to better understand the characteristics of the injury event and measure the risk of injury to develop more effective environmental, technological, and clinical management strategies. This research used methods that have limited applications to predicting human responses. This limits the current understanding of the mechanisms of TBI in humans. As a result, the purpose of this research was to examine the characteristics of impact and dynamic response that leads to a high risk of incurring a TBI in a human population. Twenty TBI events collected from hospital reports and eyewitness accounts were reconstructed in the laboratory using a combination of computational mechanics models and Hybrid III anthropometric dummy systems. All cases were falls, with an average impact velocity of approximately 4.0 m/s onto hard impact surfaces. The results of the methodology were consistent with current TBI research, describing TBI to occur in the range of 335 to 445 g linear accelerations and 23.7 to 51.2 krad/s² angular accelerations. More significantly, this research demonstrated that lower responses in the antero-posterior direction can cause TBI, with lateral impact responses requiring larger magnitudes for the same types of brain lesions. This suggests an increased likelihood of incurring TBI for impacts to the front or back of the head, a result that has implications affecting current understanding of the mechanisms of TBI and associated threshold parameters.

Keywords: Traumatic brain injury, fall, brain injury, impact biomechanics

1.0 Introduction

Traumatic brain injury (TBI) is a significant cause of morbidity and mortality around the world. In the United States alone, 1.7 million Americans sustain some type of TBI annually. Of this 1.7 million TBIs, 275,000 hospitalizations, 1.3 million Emergency room visits, 52,000 deaths, and 124,000 disabilities were the result (Rutland-Brown et al., 2003; Faul et al., 2010). These head injuries represent one of the leading causes of mortality and morbidity in adults up to the age of 45 in the United States (Centres for Disease Control and Prevention, 2011). Past research has been undertaken in order to better understand the mechanism of injury for TBI to develop more effective environmental and clinical prevention and treatment strategies (Kleiven, 2007; Deck and Willinger, 2008).

The current understanding of the mechanisms of TBI is based upon research that was conducted in the 1960s through to the 1980s using simple physical models, cadaveric impacts, and primates (Gennarelli et al., 1971, 1972, 1979, 1981; Gennarelli, 1983; Gennarelli and Thibault, 1982; Gurdjian and Gurdjian., 1975; Gurdjian and Gurdjian., 1980). Much of this extensive work focused on the magnitude and duration of linear and rotational acceleration from non-impact and impact events and how that related to the mechanisms of injury for TBI. While this research created considerable insight into TBI, it was conducted using non-human models and none that involved living human subjects. These non-human methods provided information but severely limited the transferability of the results to humans (Ommaya, 1985; Doorly and Gilchrist, 2006; Yoganandan et al., 2011). In particular, any threshold parameter to predict TBI was scaled from brains much smaller than the human brain, leading to difficulties in applying the risk metrics to human populations. Also, while some of the brain tissue characteristics of the animal models were similar to those of human, they were anatomically and geometrically

different, creating uncertainties surrounding the mechanisms of injury linked to TBI. Finally, while cadaveric research was a good representation of the human head and brain geometry, it suffered from limitations surrounding the methods used to characterize the brain tissue and changes in human brain tissue and vasculature after death (Prange and Margulies, 2002; MacManus et al., 2014). Some researchers attempted to account for some of these limitations by using finite element models of the human brain to further the understanding of TBI (Huang et al., 1999; Kleiven, 2003), but these models were simulations and not reconstructions of events that occurred in reality and thus had limited application to describing TBI risk in the living human being.

While this large body of TBI research had its limitations, some conclusions were derived. Primarily, a linear acceleration based threshold around 300g would lead to skull fracture or other form of TBI (Hoshizaki and Brien, 2004; Yoganandan and Pintar, 2004). This use of a blanket threshold for all brain injuries within the category of TBI based on linear acceleration (subdural hematoma, epidural hematoma, contusion, and others) is probably simplistic as these injuries all involve different anatomical parts of the head/brain and likely have differing mechanisms of injury (Post et al., 2014a). Secondly, several researchers suggested that TBI injuries such as subdural hematoma demonstrated a directional sensitivity (Gennarelli et al., 1972; Zhou et al., 1995; Huang et al., 1999; Kleiven., 2003). Since that original research into TBI describing these injury thresholds and directional sensitivity, there has been no research confirming these effects on human subjects, probably reflecting the difficulties in obtaining real world accident data. Characterizing these phenomena would allow for a better understanding of how the impact characteristics influence the occurrence of TBI. Research investigating TBI using human subjects also allow for an understanding of how transferrable previous research using a variety of

models is to the live human. As a result, the purpose of this research is to investigate the influence of event characteristics such as impact direction on the risk of TBI in real-life human subjects.

2.0 Material and Methods

Medical doctors at participating hospitals (Hull Hospital and the Ottawa General Hospital in Canada, and the National Department of Neurosurgery at Beaumont Hospital, in Ireland) identified patients who met the subject inclusion criteria for this research. Each subject was interviewed, reconstructive questionnaire completed (Post, 2013), and informed consent signed. All ethical practices for human research were followed for this study. The questionnaire recorded the characteristics of the impact event to allow for the laboratory reconstructions. As reconstruction information from eyewitness and patient recollections can be fraught with error, stringent selection criteria were applied for subject identification. The information that was required to meet the selection criteria included: age (must be over 18 years of age), sex, mass and height of subject, description of event, impact location, surface, and presence of medical imaging. There were no exclusion criteria based on sex or mass of subject and as a result male of females of any mass were included in the subject pool. If there were no details describing the impact location or surface, that subject was excluded, as this information is critical for the reconstruction. There were no exclusion criteria for the impact surface as all impacts were to common objects and easily transferrable to the laboratory via a suitable impact anvil. In addition, subjects must have incurred a TBI injury that was identified and confirmed by radiologist and/or neurosurgeon by computed tomography (CT) or magnetic resonance imaging (MRI) scan within 24 hours of incurring the injury. The incident in which the subject incurred the TBI must have been a fall, without complications such as being pushed or having impacted other surfaces before

the head made contact with the ground. The impact location was confirmed by both contusions evident on the scalp by CT scan (figure 1) in addition to identification of the site on their head and recorded by physician. In total, over 700 subjects were reviewed from these hospitals and of those only 20 were found to match the selection criteria. This method of collecting reconstruction information from hospital populations has been conducted in the past and has provided results that were consistent with previous human anatomical tolerance levels (Post, 2013; Post et al., 2014a). The subject data collection yielded a variety of TBI (contusions, epidural/subdural hematoma, and subarachnoid hemorrhage), brief descriptions of the falling event and resulting TBI lesions incurred are presented in Table 1.

Table 1. Subject (case), event, and resulting TBI lesion description of each case.

Case	Brief description of event	TBI
1	Fell face-first onto a concrete floor from a standing height.	Subdural Hematoma
2	Slipped, fell backwards and hit head against ground.	Subdural hematoma
3	Fell sideways from a small ladder (height above ground = XX) and hit side of head on concrete floor.	Subdural hematoma; Epidural hematoma; Contusion
4	Tripped, fell forwards, and hit against concrete floor.	Subarachnoid hemorrhage
5	Lost footing and hit right occiput on a wooden deck.	Subdural hematoma; Subarachnoid hemorrhage
6	Fell out of bed and hitting head.	Subdural hematoma
7	Passed out hitting head on concrete floor.	Subdural hematoma; Subarachnoid hemorrhage; Contusion

8	Lost her balance and fell, hitting head on pavement.	Subdural hematoma
9	Fell to the road, hitting head.	Subdural hematoma, Subarachnoid hemorrhage
10	Slipped and hit head.	Subarachnoid hemorrhage; Contusion
11	Slipped and fell hitting head on concrete.	Subdural hematoma; Contusion
12	Slipped, hitting head on pavement.	Epidural hematoma
13	Slipped and hit the posterior aspect of head on wooden deck.	Subdural hematoma
14	Slipped and fell hitting head on the curb.	Subdural hematoma; Contusion
15	Fell backwards off of a step and hit head on wall.	Parenchymal hemorrhage; Contusion
16	Lost balance and fell forward, hitting head on concrete.	Subdural hematoma
17	Lost balance hitting head on concrete	Subdural hematoma
18	Tripped fell forward hitting head.	Subdural hematoma
19	Tripped fell forward and hit head.	Subdural hematoma
20	Slipped and hit head on steel handrail.	Subdural hematoma

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147 The reconstructive questionnaire established the parameters used to conduct the Hybrid III
148 laboratory reconstructions of the accidents resulting in TBI. While specifics such as impact
149 location and vector for the impact reconstruction was determined directly from the questionnaire,
150 the head impact velocity was determined through Mathematical Dynamic Model (MADYMO,
151 TASS International, Livingston USA) simulations. Once the parameters including impact
152 location, surface, surface geometry, and head velocity were determined, an instrumented 50th

Hybrid III anthropometric dummy headform and neck form was attached to a monorail and used to re-create the impact (Figure 2). The anvil at the base of the monorail was equipped with the same impact surface as described in the report forms. The Hybrid III headform was then dropped onto the described surface three times per injury condition.

Figure 1. Computed tomography scans indicating: (red arrows) TBI; and (green arrow) impact location on scalp.

Figure 2. Monorail and Hybrid III head and neckform impact with concrete anvil.

2.1 MADYMO reconstructions

To determine the approximate head impact velocities, MADYMO simulations were conducted (Figure 3). MADYMO is a rigid multibody dynamics software package that allows for an approximation of the kinematics of the body to be ascertained for a falling event (Adamec et al., 2010). The software accomplishes this by allowing the user to place an appropriately sized model of a human being into a virtual space that closely reflects the reality of the environment in which the event occurred. In this case, the events were always falls, so the simulations allowed

171 for a representation of how the individual fell and in this way determine the head contact
172 velocities. This software has a series of ellipsoid pedestrian models that were validated for use in
173 car crash impacts and has been frequently used for forensic reconstruction of falls and other
174 brain and body injury events (Adamec et al., 2010; O’Riordain et al., 2003). While this software
175 has been used to conduct brain injury research to analyze head impacts (Doorly and Gilchrist,
176 2006; Doorly, 2007; Fréchède and McIntosh, 2009), in the present research it was simply used to
177 ascertain the head contact velocities. These types of MADYMO simulations are dependent upon
178 the accuracy of eyewitness reports, which can be unreliable. To account for this variation,
179 researchers commonly approach the simulation with a ‘most likely’ MADYMO reconstruction
180 that is closest to the description of the event. From this ‘most likely’ event, a series of +/-
181 percentages of joint angle and position simulations are conducted to determine the tolerance of
182 error for any particular simulation (Doorly, 2007; Forero Rueda and Gilchrist, 2009; Adamec et
183 al., 2010). Another method of establishing this corridor of variance is to conduct the ‘most
184 likely’ MADYMO reconstruction and then create a corridor of variance based on the possible
185 joint positions for the head to be impacted in the location identified by CT scan, and then
186 determine the upper and lower boundaries for the velocity of the event (Post, 2013; Post et al.,
187 2014a; Post et al., 2014b). This variation would reflect falling mechanics such as having an arm
188 extended to reduce the falling velocity, or having the shoulder contact the ground before the head
189 impact occurs (Usman et al., 2014). These methods have both been used for brain injury
190 reconstructions and produced results consistent with brain injury literature (O’Riordain et al.,
191 2003; Doorly, 2007; Post et al., 2014a). For the present research, the initial body positions were
192 determined from injury reports and used to determine the falling kinematics and final head
193 velocity upon contact with the ground. For each case a series of MADYMO simulations was

conducted to determine the possible values for head impact velocity. These values were affected by initial body position of the ellipsoid model, as well as joint angles and positions. This sensitivity analysis allowed for the determination of the lowest and highest possible head contact velocities for each event. These velocities were used as the input velocity parameters for the Hybrid III monorail reconstructions, creating a corridor of responses. This method allows for measurement of the possible responses of the headform (in this case) that encapsulates part of the error inherent to using eyewitness reports for the description of the initial event, providing conservative through to liberal response values for headform motion.

Figure 3. Sample frames of one MADYMO fall simulation.

2.2 Equipment

To simulate the TBI falling events a monorail drop rig and an instrumented Hybrid III 50th headform and neckform was used. The monorail had a 4.7 m rail to which the headform and neckform was attached using a drop carriage with ball bushings. The carriage runs along the rail and the ball bushings reduced the friction of the drop to achieve a low friction condition. The carriage was released by pneumatic piston, and impact velocity was determined by photoelectric time gate set up within 0.02 m of the impact. The anvil was changed depending on the surface identified in the report forms. The most common impact anvil was concrete, with some wood anvils, one steel, and one carpet with underlay condition.

The headform used for this research was a 50th Hybrid III headform (mass 4.54 ± 0.01 kg) and neckform. The neckform was attached to the drop carriage to allow for translational and rotational movement of the head upon contact with the impact surface in 6 degrees of freedom (Figure 2). The headform was instrumented with a 3-2-2-2 accelerometer array for the measurement of the three dimensional kinematics upon impact (Padgaonkar et al., 1975). The accelerometers used were Endevco 7264C-2KTZ-2-300, and were sampled at 20 kHz. The signals were captured using a DTS TDAS system and filtered using a CFC class 1000 filter. The frame of reference for the headform was x-axis forward, y-axis to the left of the head, and z-axis upwards.

Peak resultant linear and rotational acceleration, as well as the x, y, and z components for both linear and rotational accelerations were examined to determine the effect of magnitude and direction on the resulting TBI cases. Statistical comparisons were made using an ANOVA, with a Tukey post-hoc test when significant main effects were present. The alpha was set to a $p=0.05$ level, thereby establishing significance at a 95% confidence interval. Of the velocities determined by the MADYMO sensitivity analysis, only the lower boundary thresholds were reported and analyzed statistically as these would represent the lower thresholds for the traumatic brain injuries (Post et al., 2014a).

3.0 Results

In total 20 cases of traumatic brain injury from falls were reconstructed. The subject population was an average age of 68 years, ± 14.5 years. Of these 20 cases, 12 were male and 8 female. Of these, there were 19 subdural hematomas, 7 contusions, 6 subarachnoid hemorrhages, 2 epidural hematomas, and 1 parenchymal hemorrhage present. Impact locations, surfaces, and MADYMO determined impact velocities (table 2) for the reconstructions.

236 Table 2. Impact location, velocities, and impact surfaces for each TBI case.

Case #	Impact location	Velocity/Surface	Case #	Impact location	Velocity/Surface
1		3.0 - 4.0 m/s Concrete	11		3.7 - 5.8 m/s Concrete
2		4.7 - 5.2 m/s Concrete	12		3.6 - 4.9 m/s Concrete
3		5.1 - 6.1 m/s Concrete	13		4.8 - 6.2 m/s Wood
4		4.1 - 5.4 m/s Concrete	14		3.8 - 6.0 m/s Concrete
5		3.9 - 6.2 m/s Wood	15		4.8 - 5.8 m/s Concrete
6		2.2 - 3.3 m/s Carpet	16		5.1 - 5.6 m/s Concrete
7		3.9 - 6.2 m/s Concrete	17		4.5 - 4.8 m/s Concrete
8		3.3 - 4.8 m/s Concrete	18		3.5 - 4.2 m/s Concrete
9		4.8 - 6.2 m/s Concrete	19		4.7 - 5.1 m/s Concrete
10		3.6 - 4.9 m/s	20		5.4 - 6.1 m/s

The results were separated into direction of impact and included in tables 3 and 4. An example of the typical loading curve shape and duration is represented in figure 4. There was no significant main effects for velocity of impact or HIC_{15} for the different impact directions ($p>0.05$). There were significant main effects between all measures of resultant and component linear and rotational accelerations ($p<0.05$) except for resultant peak linear acceleration and z-axis linear acceleration ($p>0.05$). The peak resultant rotational acceleration responses demonstrated magnitudes that were smaller for the front (33.7 krad/s^2) ($p=0.007$) and rear (23.7 krad/s^2) ($p=0.001$) impacts when compared to the side impacts (46.7 and 51.2 krad/s^2). The front impacts were of larger magnitude than the rear ($p=0.012$) and there was no significant difference between left and right side impacts for peak resultant rotational acceleration ($p=0.483$).

Table 3. Linear acceleration results for the impacts to the Hybrid III headform. Standard deviations are in brackets.

Impact location	Velocity (m/s)	Linear (g)				
		HIC_{15}	Resultant	X	Y	Z
Front	4.0 (1.1)	181 (113)	335.5 (165.9)	265.2 (172.2)	160.8 (91.3)	91.3 (57.9)
Rear	4.2 (0.5)	219 (71)	334.9 (80.7)	304.6 (98.3)	72.4 (62.6)	90.1 (27.7)
Right side	4.2 (0.8)	196 (82)	442.1 (130.9)	93.9 (43.4)	407.1 (130.7)	115.5 (85.3)

Left side	4.2 (0.8)	220 (80)	445.2 (96.9)	129.6 (44.0)	417.7 (103.4)	76.8 (39.1)
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Table 4. Rotational acceleration results for the impacts to the Hybrid III headform. Standard deviations are in brackets.

Impact location	Velocity (m/s)	Rotational (krad/s ²)				
		HIC ₁₅	Resultant	X	Y	Z
Front	4.0 (1.1)	181 (113)	33.7 (15.5)	18.8 (10.6)	23.2 (18.4)	15.3 (12.4)
Rear	4.2 (0.5)	219 (71)	23.7 (6.8)	7.1 (5.6)	21.1 (8.4)	4.4 (3.5)
Right side	4.2 (0.8)	196 (82)	46.7 (8.7)	41.4 (10.5)	8.8 (2.6)	22.7 (5.4)
Left side	4.2 (0.8)	220 (80)	51.2 (13.8)	43.4 (7.6)	8.2 (1.9)	24.4 (16.3)

Figure 4. Sample linear (left) and rotational (right) acceleration curves for a TBI reconstruction.

When comparing the linear acceleration components, the x-axis component had the largest magnitude response for the front and rear impacts, and the y-axis for the impacts to the left and right side ($p=0.023$). When comparing these dominant axes (x-axis vs y-axis), the front (265.2 g) and rear (304.6 g) impacts incurred lower magnitudes than the side impacts (407.1 g and 417.7 g) for peak magnitude of component ($p=0.023$). When comparing the rotational acceleration components, a similar relationship occurs to that of the linear acceleration components. The front and rear impacts produced the largest rotational accelerations in the y-axis (23.2 and 21.1 krad/s²) ($p=0.001$). The left and right side impacts had the largest magnitude responses in the x-axis with 41.4 and 43.4 krad/s² respectively ($p=0.001$). When comparing the dominant (largest magnitude response by component) component axes, y-axis front/rear to x-axis left/right side impacts, the antero-posterior impact direction has significantly lower magnitudes of response ($p=0.001$).

4.0 Discussion

The peak resultant linear (335 – 445 g) and rotational accelerations (23.7 – 51.2 krad/s²) in this study are consistent with those reported in the literature using human datasets (Tables 5 and 6). The rotational accelerations are significantly larger than those reported by Ommaya et al (1967) and others, but that may be a result of differences in populations and methodologies as earlier comparative datasets were from impacts to animals and not reconstructions of real life injury events. While there were no skull fractures in this population, the peak linear acceleration magnitudes are within the ranges found for cadaveric impacts at similar velocities to those in this research (Hodgson and Thomas, 1971; Mertz, 1985; Prasad and Mertz, 1985; Mertz et al., 1996; Yoganandan and Pintar, 2004). The skull fracture research presented in the literature was conducted with varying anvils, and varied pressure distributions, and demonstrated that skull

fracture can occur at a range of 200 to 600 g depending on methodology (Hodgson, 1967; Schneider et al., 1972; Nahum et al., 1980; Got et al., 1983). It is possible that skull fractures did not occur in this research as a result of the nature of the loading to the skull; where in these cases the forces were not focused enough to create a fracture, but rather an internal brain injury. The impact conditions for TBI in this hospital population were the result of falls occurring at approximately 4.0 m/s, on an unyielding surface such as concrete. These parameters resulted in high energy impact conditions where head contact velocity and low impacting surface compliance (stiff surfaces) combine to create a high risk for TBI (Willinger et al., 1992; Gilchrist and O'Donoghue, 2000). These conditions created acceleration loading curves with high magnitudes and short durations (less than 5ms; Figure4) that would be expected to result in TBI. Gennarelli et al (1979) also reported high acceleration loading curves of short duration as a high risk for subdural hematoma, an injury that was very common in this hospital dataset (Figure 4). Interestingly, the HIC₁₅ magnitudes for all impacts would have predicted that there was a low risk of injury for all the cases that were reconstructed. This could be an indication of the limitations of the HIC₁₅, where it may not be suitable for all impact mechanisms or injury types.

Table 5. Peak resultant linear acceleration responses for unhelmeted head impacts producing a TBI.

Linear acceleration response range (g)	Method	Reference
200	Animal models	Gennarelli and Thibault (1982)
311 - 638	MADYMO	O'Riordain et al (2003)
236.5 - 366.5	MADYMO	Doorly and Gilchrist (2006)
236.5 - 366.5	MADYMO	Doorly (2007)

Table 6. Peak resultant rotational acceleration responses for unhelmeted head impacts producing a TBI.

Rotational acceleration response range (krad/s ²)	Method	Reference
4.5	Cadaver	Lowenhielm (1974)
34.0	Finite element	Kleiven (2003)
26.2 - 44.8	MADYMO	O'Riordain et al (2003)
33.9 - 36.9	MADYMO	Doorly and Gilchrist (2006)
11.4 - 45.2	MADYMO	Doorly (2007)

The components of the resultant accelerations were examined as they are indicative of the direction of the magnitudes of acceleration, which has been reported in the literature to have a significant effect on the injury (Gennarelli et al., 1979, 1987; Hodgson et al., 1983; Zhou et al., 1995; Kleiven, 2003). When examining the results of this research it was discovered that for the present fall reconstructions (TBI), the front and rear head impact had x-axis linear, and y-axis rotational accelerations that were significantly higher than the other components. In addition, for the impacts to the side of the head, the y-axis linear and x-axis rotational accelerations were significantly larger than the other components, which is indicative of a typical lateral impact. Interestingly, the impacts in the antero-posterior direction all had significantly lower linear and rotational responses in comparison to the lateral impacts for the same types of TBI injuries, suggesting a susceptibility to incurring TBI from these directions. In comparison to the literature examining the effect of direction on TBI injury, Zhou et al (1995) identified that it was more likely to incur an SDH from rear impacts than the front; this result however was not consistent with the present data, which found no difference between front and rear responses for similar

TBI lesions. High strains on brain vasculature have been reported for antero-posterior motion using a finite element model of the human brain (Kleiven, 2003). This motion was identified by Kleiven (2003) as producing large relative brain – skull motions that create large deformations on the vasculature tethering the brain to the skull, increasing the risk for subdural or other hematomas. These brain skull motions are much lower for lateral impacts of similar magnitudes, which is likely a result of the presence of the falx cerebri reducing motion of the brain in that direction. This anatomical consideration would result in lower magnitude head motions causing TBI in the antero-posterior direction, whereas large magnitude motions would be required for similar injury in the lateral direction. Zhou et al (1995) also found that antero-posterior motions caused larger strains in regions of the brain associated with SDH than lateral motions. Zhou et al (1995) and Kleiven (2003) employed finite element models to simulate these motions, and the 20 real life reconstructions of TBIs resulting from falls reported in this research are consistent with their findings. The results in the present research support these previous studies and identify that direction of impact has considerable effect on the magnitudes of acceleration required to cause a TBI. This may be one reason why a concrete threshold for TBI has been difficult to determine, as currently research considers every TBI event to be the same when perhaps they should be separated into categories based on where on the head the individual was impacted. As human brain tissue anatomy and characteristics have been reported to be anisotropic on the macro and micro scale, different directions of loading would be expected to have significant influence on the amount of loading required to cause injury in these structures. This theory is supported by concussion injury research indicating a susceptibility of injury to lateral impacts that was reported by McIntosh et al (2014).

4.1 Limitations

339 Reconstructive methodologies conducted in laboratory have certain limitations, which should be
340 taken into consideration when examining the results. The present MADYMO simulations only
341 approximate the actions and body positions of an individual when falling (in this case). The
342 accuracy of these simulations is dependent on accurate estimates of the incident from eyewitness
343 reports and subject descriptions. Since this type of reporting can be in error, the simulations are
344 approximations of the real event, which is why a series of reconstructions were conducted in an
345 attempt to understand the variance inherent to this limitation. The use of the Hybrid III is
346 common in the car crash industry to establish risk of injury and has seen increased use for injury
347 reconstructions. However, while the headform and neckform do provide responses in the range
348 of the human, they would not be perfectly biofidelic (Kendall et al., 2012). The subject
349 population in this research was all adults, but even within an adult population there are a variety
350 of head sizes. The head sizes of these subjects were slightly different from that of the Hybrid III
351 headform and these differences may influence the resulting head motions. A monorail drop rig
352 was used for these reconstructions, which is appropriate for the nature of the falling mechanisms
353 of injury. However, the monorail cannot replicate every possible falling impact type and as a
354 result the values may have an amount of error. Since this bias would be similar for all impacts
355 that occurred, it may be assumed that the comparisons retain a certain amount of validity for this
356 particular subject population. The tolerances shown for these different forms of TBI are very
357 specific to the magnitude, but also the duration of the event. As such, it should be noted that all
358 of these impacts resulted in very brief ($> 6\text{ms}$) acceleration pulses in addition to large
359 magnitudes of response, and any brain injury comparison can only be made with and between
360 similar mechanisms of injury. Finally, the subject population in this research is small ($n=20$), and

as a result it is inappropriate to definitively suggest that the results would be applicable to the entire population.

5.0 Conclusion

This research examined the biomechanics of TBI for 20 fall reconstructions. The impact conditions for TBI from falls were approximately 4.0 m/s head contact velocity with an impact to a hard unyielding surface. These conditions created high magnitudes of head accelerations reflecting a significant risk of TBI. The magnitudes of response for TBI from falls were between 335 and 445 g linear acceleration and 23.7 to 51.2 krad/s² rotational acceleration. More significantly, this research demonstrated that lower responses in the antero-posterior direction can cause TBI, with lateral impact responses requiring larger magnitudes for the same types of brain lesions. This research suggests an increased likelihood of incurring TBI for impacts to the front or back of the head for this subject population. This result that is supported by simulated brain motions (Zhou et al., 1995; Kleiven, 2003) but has now, for the first time, been replicated using reconstructive data based on real events causing human injuries. This finding has significant importance as impact location may be a key consideration in understanding the clinical mechanisms of TBI and establishing threshold parameters that can be used for prediction. These results could potentially aid clinicians in assessing the likelihood of severe brain injuries from the characteristics of the head impact event. Biomechanically, this research demonstrates that thresholds of injury may be dependent on impact direction, which is a consideration that has not been accounted for in recent threshold of TBI and concussion research. Not accounting for this parameter may be one reason why a definitive threshold for injury has thus far eluded researchers investigating brain injury.

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Conflict of interest statement

There are no conflicts of interest with the submitted research.

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