Traumatic Brain Injuries: The influence of the direction of impact

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30 *Abstract*

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32 *Background:* Head impact direction has been identified as an influential risk factor in the risk of
33 TBI from animal and anatomical research, however to date, there has been little investigations into
34 this relationship in human subjects. If a susceptibility to certain types of TBI based on impact
35 direction was found to exist in humans it would aid in clinical diagnoses as well as prevention
36 methods for these types of injuries.

37 *Objective:* The purpose of this research was to examine the influence of impact direction on the
38 presence of TBI lesions, specifically, subdural hematomas, subarachnoid hemorrhage, and
39 parenchymal contusions.

40 *Methods:* Twenty reconstructions of falls that resulted in a TBI were conducted in a laboratory
41 based on eyewitness, interview, and medical reports. The reconstructions involved impacts to a
42 Hybrid III anthropometric dummy, and finite element modelling of the human head to evaluate the
43 brain stresses and strains for each TBI event.

44 *Results:* The results showed that it is likely that increased risk of incurring a subdural hematoma
45 exists from impacts to the frontal or occipital regions, and parenchymal contusions from impacts to
46 the side of the head. There was no definitive link between impact direction and subarachnoid
47 hemorrhage. In addition, the results indicate that there is a continuum of stresses and strain magnitudes
48 between lesion types when impact location is isolated, with subdural hematoma occurring at lower
49 magnitudes for frontal and occipital region impacts, and contusions lower for impacts to the side.

50 *Conclusion:* This hospital dataset suggests that there is an effect that impact direction has on TBI
51 depending on the anatomy involved for each particular lesion.

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53 Key Words: Traumatic brain injury; biomechanics; subdural hematoma, contusion, subarachnoid
54 hemorrhage
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57 Running title: Effect of direction on TBI
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Introduction

Traumatic brain injuries (TBI) contribute to disability and affect the quality of life for many in society, often compromising the persons' ability to work and interact socially. The costs of these injuries are including lost productivity and care facilities for those no longer able to care for themselves. In the United States, there are an estimated 1.7 million cases of traumatic brain injuries every year¹. While Canadian statistics in this area are not complete, it can be estimated that 170,000 traumatic brain injuries per year occur in Canada. In addition, 150,000 Canadians are likely to suffer nonfatal traumatic brain injuries that will not require hospital treatment. The elderly and children are the most likely to experience a traumatic brain injury, with falls being the most common cause². Of all injuries, brain trauma has the singular distinction of having the highest morbidity and mortality³. As a result, a great deal of research has been undertaken to help prevent head injuries. Recently much attention has been paid to brain injuries due to their serious effects on the nervous system and the repercussions to the quality of life of the injured person.

The severe nature and serious repercussions of TBI has resulted in research designed to better understand the factors that contribute to these injuries; such as what are the characteristics of the event that are required to cause a TBI⁴⁻⁷. In particular, identifying the types of head motions that can cause a TBI has been an important area of research⁸⁻⁹. Identifying susceptibility to incur TBI from certain impact locations would be knowledge that could allow for clinicians to better assess likelihood of type of brain injury based on the event. In addition, it would allow for biomechanists and engineers to innovate technologies and create safer environments that would prevent and protect the brain from TBI through understanding of how the brain responds to impacts from certain directions. There has been some research involving monkeys and cadavers examining the influence of impact location, and thus, the direction of motion on pathological responses, which identified a link between severity of brain injury and direction^{8,10-11}. They reported axonal injury in the brain was proportional to the degree of coronal motions¹⁰. The same group identified an association of sagittal plane motions with severe brain injury in primates that were likely caused by the lack of anatomical structures such as the falx¹¹. Since the falx acts in the Sagittal plane, it was proposed that the falx played a role in reducing the motion of the brain for lateral impacts thus reducing high strains¹¹. The influence of impact direction on subdural hematoma (SDH) was studied by Kleiven¹² and Huang et al¹³ using finite element modelling simulations with motions in the sagittal plane causing an increased likelihood of injury.

While this research demonstrated a mechanism of injury involving the influence of impact location on the resulting brain injury, it did not reflect an event involving human subjects and human anatomical considerations. This limitation of transferring animal and anatomical data to human responses is commonly mentioned in these and other publications¹⁴⁻¹⁶. The mechanisms of injury for animals have long been considered as having severe limitations when applied to our understanding of injury mechanisms for humans¹⁴. Thresholds associated with brain injury obtained from animal research are speculative when applied to human injuries, which may be one reason why there is currently no definitive threshold or measurement variable for TBI. In addition, traumatic brain injury defines a broad spectrum of lesions and it is likely that each lesion type is influenced by the nature of the motion from unique impact directions causing stresses and strains to specific anatomical regions^{9,17}. As a result, the purpose of this research is to examine the relationships between impact direction and the presence of different traumatic brain injury lesions in a human population.

Methods

The cases were selected by physicians from the Hull Hospital in Hull, Canada, the Ottawa General Hospital in Canada and the National Department of Neurosurgery at Beaumont Hospital, Dublin, Ireland. For this research reconstructions were conducted based on eyewitness and patient reports. As a result, falls were the only mechanism of injury represented in this dataset as they were more feasible to reconstruct in a laboratory and represented the most common injury method for TBI^{15,18}. In total, over 700 cases of head injuries were presented at the participating hospitals during the data collection period. These cases were then reviewed based on the quality of the description of the traumatic event, the cranial imaging, and the outcome. As eyewitness and patient recollections of the event can be inaccurate, rigorous inclusion criteria was applied to the collection of reconstructable cases to ensure the minimum possible error for the simulations. As previously stated, the mechanism of injury was limited to falls, with a clearly defined TBI present on computed tomography (CT) or magnetic resonance imaging (MRI) scan which was confirmed by neurosurgeon or radiologist. The imaging must have taken place within 24 hours of the original event, and the fall must have been a slip or trip without any external push, or contact with any object before contact with the impact surface. Impact location must have been identified by both bruising on the skin in addition to identification by the patient of the site on

their head and recorded by physician. The subjects' height, weight, gender was also have been recorded to ensure accuracy of the falling simulations. In addition, to be included the subjects must not have been taking antiplatelet or anticoagulation medications. From that analysis, the following injuries were included: parenchymal contusions, subdural hematomas (SDH), and subarachnoid hemorrhage (SAH). From these cases, twenty (20) were identified as having met the strict criteria for accurate laboratory reconstructions and each subject signed an informed consent form. This group was represented by 13 males and 7 females, with an average age of 68 years (15 years standard deviation). This group of subjects were evaluated to have a Glasgow Coma Scale (GCS) of 14/15, with no post-traumatic or retrograde amnesia. This information from the eyewitness and subject reports established the initial parameters for the computer and physical model simulations and contained information such as: impact vector, location of the impact on the head, velocity of impact, and impact surface. To accomplish the reconstruction of the TBI event both MATHematical DYnamic MOdels (MADYMO, TASS International, Livingston USA) and Hybrid III anthropometric dummies were used. This method for researching brain injury for falls has been conducted and published in the past and has produced informative results that are consistent with previous data based on human anatomical research^{9,17,19-23}.

MADYMO reconstructions

Mathematic dynamic models are tools commonly used to conduct reconstructions of falling accidents in a human population from eyewitness and subject reports^{18,20,24-25}. This software allows for the estimation of human kinematics for a fall from a starting position to point of impact with a surface. While having some limitations this tool is the best available for this type of research and allows for a better estimation of impact parameters than using simple mathematical equations to determine head impact based on gravity or pendulum motion^{18,24}. In the case of this research, this software is used to estimate head impact velocity as that cannot be determined from the eyewitness and subject reports directly. This particular tool is quite effective because it has a large database of human body models, which includes a series of ellipsoid pedestrian models^{19,24}. The pedestrian models were validated using a variety of impactors to determine the risk of injury to pedestrians from vehicle impacts²⁴. The MADYMO simulations were used to reconstruct the falling motions based on the anthropometrics of the subject, initial

and final body positions of each subject as described by eyewitnesses and subjects^{18,21,25-26}. The MADYMO simulations were used in this research to approximation of the final kinematics of the head as it impacted the ground.

For each reconstruction, an appropriately sized ellipsoid pedestrian model closely matching the anthropometrics of the subject was chosen. The ellipsoid pedestrian model was then placed in a simulated environment constructed to match the environment in which the subject fell (Figure 1). As there can be some variation from injury reports, a sensitivity analysis was conducted on each fall^{9,17,20,25}. Several simulations were conducted representing a variety of reasonable joint angles and body positions^{18,20-21,25}. From this sensitivity analysis, the fastest and slowest head impact velocities were calculated, which then defined the most likely corridor of impact head velocities. This analysis completed the necessary information required for conducting a physical reconstruction of the fall event.

Figure 1. Example MADYMO simulation for a trip up a step and a forehead impact; (A) initial walking position; (B) position prior to impact.

Laboratory reconstruction

The falling impact reconstructions of each incident were carried out by a guided monorail device equipped with a hybrid III head and neck form.

Monorail

To simulate falling impacts, a monorail was used (Figure 2). The monorail consisted of a 4.7 m long rail to which the drop carriage was attached. The drop carriage runs along the rails on ball bearings to reduce the effects of friction on inbound velocity and was released by pneumatic

piston. A hybrid III 50% headform and neck was attached to the drop carriage to obtain three dimensional impact characteristics. The impact velocity was measured using a photoelectric time gate placed within 0.02 m of the impact.

Figure 2. Monorail reconstruction of a fall using the Hybrid III headform; (A) fall to concrete; (B) fall onto carpet with underlay on a concrete anvil.

The anvil at the base of the monorail was composed of steel with the surface material representative of the material contacted by the head as described in the report. The base of the monorail was 0.67 m high, 0.30 m wide, and 0.38 m deep and fixed to the floor using 6 concrete bolts. The 6 meter vertical track was bolted to the wall and the ceiling to minimize movement of the system and background noise or vibration.

Hybrid III

A 50th percentile adult Hybrid III headform (mass $4.54 \pm 0.01\text{kg}$) and neck was instrumented for measurement of three dimensional kinematics according to Padgaonkar's²⁷ 3-2-2-2 accelerometer array (Figure 2). The accelerometers used were Endevco 7264C-2KTZ-2-300.

Laboratory reconstruction procedure

Once an event was identified for reconstruction the injury reports were used to identify the impact parameters. As previously described, these parameters defined the characteristics of the event to allow for the closest possible representation of the event. The parameters of impact

surface and location were easily determined from the report and CT/MRI scans (in the case of impact location). The impact velocity is the parameter that is most sensitive to error as the MADYMO simulations depend to a large part on the accuracy of the patient and eyewitness recollection. The experimental procedure involved a monorail to which the headform and neckform was attached and allowed to drop and impact the same type of surface in the same location on the head as the subject. In the case of these reconstructions, the Hybrid III 50% headform was affixed to the monorail and dropped from the described height to obtain the falling impact velocity as determined by the MADYMO reconstructions. For each reconstruction, three trials were conducted for each velocity. This resulted in between six and nine trials per injury reconstruction as there were consistently at least two or three reasonable impact velocities. The impact surface was matched to the accident impact surface and thus its characteristics. The impact site was identified from images of the impact location on the subject and the location of scalp swelling on the CT/MRI scans (Figure 3).

Figure 3. Medical images showing: (A) subdural hematoma (red arrow), and (B) impact site indicated by green arrow.

The impacts were sampled at 20 kHz and recorded using DTS TDAS PRO software. All data was filtered with a 1650 Hz lowpass Butterworth filter according to the SAE J211 head impact conventions. This reconstruction provided a three dimensional description of the kinematics of the impact event. The resulting x, y and z linear and rotational acceleration loading curves that are representative of the impact to the Hybrid III head and neckform complex were captured and used for the finite element model simulations. The loading curves were applied to the UCDBTM at the centre of gravity of the model.

Finite element model (UCDBTM)

The model used in this research was developed in Dublin and is known as the University College Dublin Brain Trauma Model (UCDBTM)²⁸⁻²⁹. The finite element solver used was ABAQUS (Simulia, Providence, RI, USA). The geometric parameters of the model were from a male cadaver as determined by medical imaging techniques²⁸⁻²⁹. The head and brain finite element model was comprised of ten parts: the scalp, skull (cortical and trabecular bone), cerebellum, brain stem, tentorium, falx, grey and white matter and cerebrospinal fluid (CSF). The CSF layer was modeled using solid elements with a high bulk modulus and a low shear modulus to create a sliding boundary condition between the interfaces of the CSF and brain. The algorithm allowed no spaces between the pia and the CSF layer. For the sliding surfaces a friction coefficient of 0.2 was used³⁰. The fluid present between the falx and the brain and between the tentorium, cerebrum and cerebellum also had the same fluid algorithm. In total, the brain model was comprised of approximately 26,000 hexahedral elements.

The current model uses the parameters by Mendis et al³¹ and Kleiven and von Holst³² which are nonlinear viscoelastic material law under large deformation. A hyperelastic material model was used for the brain in shear in conjunction with the viscoelastic material property. The hyperelastic law was given by:

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

where C_{10} and C_{01} are the temperature-dependent material parameters, and t is time in seconds. The compressive behaviour of the brain was considered elastic. The shear characteristics of the viscoelastic behaviour of the brain were expressed by:

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

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). As fluid has a high bulk modulus and zero resistance to shear, the CSF layer was modeled using solid elements with a low shear modulus as was used in other research to simulate the sliding interaction between the brain and skull³³⁻³⁷.

The characteristics of brain tissue are those approximated from cadaveric and scaled animal anatomical testing. The material properties for the brain and skull were taken from the literature and the values are shown in tables 1 and 2^{32-34,38}. The model was validated against Nahum et al's³⁹ cadaver impacts measuring cranial pressures and Hardy et al's⁴⁰ research of

brain motion. Further validations of the model for brain injury research was conducted using real life incidents by Doorly and Gilchrist¹⁵ and Post et al⁷ and were found to be in good agreement with the values in the literature for TBI lesions from falls.

Table 1. Finite element model material properties (Mpa: megapascals)

Material	Young's modulus (Mpa)	Poisson's ratio	Density (kg/m ³)
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	Hyperelastic	0.49	1060
White Matter	Hyperelastic	0.49	1060

Table 2. Material characteristics of the brain tissue used for the UCDBTM (kPa: kilopascals; GPa: gigapascals)

Material	Shear Modulus (kPa)			Bulk Modulus (GPa)
	G ₀	G _∞	Decay Constant (s ⁻¹)	
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

Region of interest identification

The CT and MRI scans of each subject were assessed by radiologists and neurosurgeons at the hospital to identify the type of lesion and the location of the damage^{7,15,17,20,41}. The CT/MRI images were then compared to the UCDBTM and a suitable region of the model was selected to represent the volume of the region of interest (ROI) of the injured area (Figure 4). This allowed for an examination of the stresses and strains in particular regions of the brain associated with SDH, SAH, and contusion separately. This type of analysis allowed for the examination of the

influence of impact direction for particular TBI types. The FE model was also scaled to the closest dimensions of the brain of the subject to account for differences in brain found in the population such as age and gender^{17-18,32}.

Figure 4. Image showing the UCDBTM region of interest representing the region of a subdural hematoma (red).

The brain tissue deformation parameters

The brain tissue deformation parameters chosen for this study were pressure, maximum principal strain (MPS), von Mises stress (VMS), shear stress, and shear strain. These variables were chosen based upon previous research which has used these parameters to describe brain injuries^{20,41-43}. The peak magnitude in the ROI was represented by the element experiencing the largest deformation using the same method as previous research^{17,20,41-43}. For each simulation an examination of the aspect ratios of each element in the model was conducted to catch any errors present in the finite element mesh. Statistical comparisons between lesion types (contusion, subdural hematoma, and subarachnoid hemorrhage) were conducted for each direction of impact (front, rear, left and right side) by ANOVA. Further analysis was conducted by ANOVA, with a Tukey post-hoc test, comparing the magnitude of dependent variable for each lesion type within each impact location. Confidence interval was set to 95%^{7,17-18}. The software used for statistical analyses was SPSS version 17.0 (IBM, NY, USA).

Results

Twenty cases of falls were reconstructed that had a total of 20 subdural hematomas, 7 subarachnoid hemorrhages, and 7 contusions. Descriptions of each case and impact locations can be found in figure 5. In total, there were 15 impacts that occurred in the antero-posterior direction and 5 that occurred in the lateral direction. The magnitudes of response presented in tables 3 through 5 are the results of the lowest head impact velocities used from the MADYMO simulations and thus represent the lowest possible threshold for injury for the SDH, SAH, and parenchymal contusions for this methodology. The relationships between lesion and impact direction were consistent (linear) throughout the other head impact velocities used from the MADYMO simulations.

Case #	Impact location	Velocity/Surface	TBI	Case #	Impact location	Velocity/Surface	TBI
1		3.0 m/s Concrete	SDH	11		3.7 m/s Concrete	Contusion; SDH
2		4.7 m/s Concrete	SDH	12		3.6 m/s Concrete	SDH; SAH
3		5.1m/s Concrete	Contusion; SDH	13		4.8 m/s Wood	SDH
4		4.1 m/s Concrete	Contusion; SAH	14		3.8 m/s Concrete	Contusion; SDH
5		3.9 m/s Wood	SDH; SAH (2)	15		4.8 m/s Concrete	Contusion
6		3.3 m/s Carpet	SDH (2)	16		5.1 m/s Concrete	SDH (2)
7		3.9 m/s Concrete	Contusion; SDH; SAH	17		4.5 m/s Concrete	SDH
8		3.3 m/s Concrete	SDH	18		3.5 m/s Concrete	SDH
9		4.8 m/s Concrete	SDH; SAH	19		4.7 m/s Concrete	SDH (2)
10		3.6 m/s Concrete	Contusion; SAH	20		5.4 m/s Steel	SDH

Figure 5. Impact location, velocity, and impact surface, and TBI types for each fall case. Bracket for TBI type denotes number of lesions found and analyzed.

Results by lesion type

Subdural hematoma

From the TBI events that were reconstructed (table 3) there were a total of 20 subdural hematomas. The majority of the lesions incurred from impacts to the frontal (11) and occipital (6) regions of the head with the rest to the sides (3). Upon statistical analyses of the UCDBTM stresses and strains for the regions of interest representing the lesions, main effects were found for all dependent variables ($p=0.01$). The maximum principal strain magnitudes for the frontal (0.334; right side $p=0.01$; left side $p=0.04$) and occipital (0.272; right side $p=0.01$; left side $p=0.01$) impacts were significantly lower than those impacts from the sides (0.456; 0.444). A similar result was found for magnitudes of VMS and shear stress, with the front (10.8; $p=0.01$ and 4.6 kPa; $p=0.019$) and rear (9.2; $p=0.01$ and 4.0 kPa; $p=0.001$) being significantly lower in response in comparison to the right side (15.8 and 6.2 kPa) impacts. The shear strain responses were of lower magnitude for just the rear location (0.393; front $p=0.009$; right side $p=0.001$; left side $p=0.006$) in comparison to impacts to the other regions. The pressure responses were significantly greater for the occipital impacts (995.7 kPa) in comparison to the frontal and right side impacts (724.5 kPa, $p=0.026$ and 591.5 kPa, $p=0.043$). The majority of comparisons to the left side showed no significance (ns), which may be reflective of the low sample number.

Table 3. Brain stress and strain responses from the UCDBTM for subdural hematoma. (Standard deviations in brackets; kPa: kilopascals)

Impact location	# of lesions	Pressure (kPa)	MPS	VMS (kPa)	Shear stress (kPa)	Shear strain
Front	11	724.5 (333.8)	0.334 (0.070)	10.8 (2.2)	4.6 (0.9)	0.496 (0.092)
Right side	2	591.5 (66.8)	0.456 (0.030)	15.8 (0.5)	6.2 (1.7)	0.620 (0.151)
Rear	6	995.7 (346.1)	0.272 (0.074)	9.2 (3.2)	4.0 (1.4)	0.393 (0.119)
Left side	1	794.0 (52.2)	0.444 (0.011)	14.4 (0.3)	5.9 (0.1)	0.621 (0.007)

Subarachnoid hemorrhage

From the twenty TBI reconstructions, there were a total of 7 subarachnoid hemorrhages. The impacts from: one frontal, one right side, three occipital, and two left side. When comparing the results of the UCDBTM for impact direction (table 4), significant main effects were found for VMS ($p=0.002$), shear stress ($p=0.011$), and pressure ($p=0.022$). When comparing impact direction, the VMS response was larger for impacts to frontal regions (18.9 kPa) in comparison to the occipital (11.2 kPa) ($p=0.002$). The occipital impacts produced magnitudes of response that were lower than the left side (15.2 kPa, $p=0.041$), but not significantly different from the right ($p=0.379$). The shear stress magnitudes were significantly larger for impacts to the frontal region (10.6 kPa) than the right side and rear impact locations respectively ($p=0.037$; $p=0.009$). The pressure responses for impacts to the frontal region were significantly smaller (1194.8 kPa) than for the right side (2117.5 kPa, $p=0.017$) but not for the other impact directions (ns, occipital: 763.6 kPa; left side: 838.3 kPa).

Table 4. Brain stress and strain responses from the UCDBTM for subarachnoid hemorrhage. (Standard deviation in brackets; kPa: kilopascals)

Impact location	# of lesions	Pressure (kPa)	MPS	VMS (kPa)	Shear stress (kPa)	Shear strain
Front	1	1194.8 (709.2)	0.461 (0.008)	18.9 (0.3)	10.6 (0.1)	0.860 (0.012)
Right side	1	2117.5 (474.4)	0.442 (0.041)	14.0 (0.9)	5.8 (0.1)	0.594 (0.007)
Rear	3	763.6 (220.0)	0.341 (0.062)	11.2 (1.9)	5.8 (0.7)	0.567 (0.052)
Left side	2	838.3 (594.7)	0.363 (0.108)	15.2 (4.1)	7.4 (0.1)	0.608 (0.276)

Parenchymal Contusion

There were seven (7) parenchymal contusions in the 20 reconstructed cases, with two from impacts to the right side, 4 from impacts to the occipital region and one to the left side. No contusions were present from impacts to the front of the head. When comparing the dependent variables for influence of impact direction (table 5), significant main effects were found for MPS

(p=0.002), VMS (p=0.003) and shear strain (p=0.014). The impacts to the right (0.244; 8.3 kPa) and left side (0.242; 7.1 kPa) were significantly lower (right side: p=0.004, p=0.007; left side: p=0.025, p=0.033) in magnitude than impacts to the occipital region (0.406; 13.6 kPa) of the head for MPS and VMS. When examining the influence of impact direction using shear strain, the occipital impacts (0.688) were of higher magnitude than the right side (0.333) impacts (p=0.013), but not the left (0.465) (p=0.286).

Table 5. Brain stress and strain responses from the UCDBTM for contusions. (standard deviations in brackets; kPa: kilopascals)

Impact location	# of lesions	Pressure (kPa)	MPS	VMS (kPa)	Shear stress (kPa)	Shear strain
Front	0	-	-	-	-	-
Right side	2	595.1 (313.8)	0.244 (0.049)	8.3 (1.8)	3.8 (1.1)	0.333 (0.120)
Rear	4	966.4 (220.3)	0.406 (0.106)	13.6 (3.8)	6.6 (2.8)	0.688 (0.271)
Left side	1	617.6 (14.4)	0.242 (0.004)	8.1 (0.2)	4.4 (0.1)	0.465 (0.011)

Overall lesion comparisons by direction

The frontal impacts produced just SDH and SAH, and for all the dependent variables calculated in the UCDBTM, the SDH occurred at lower magnitudes (pressure, p=0.042; MPS, p=0.004; VMS, shear stress and strain, p=0.001). For impacts to the right side of the head, the contusions occurred at lower magnitudes of response than the SDH and SAH for MPS (p=0.001 respectively), VMS (p=0.001 respectively), and shear strain (SDH, p=0.005; SAH, p=0.030). For shear stress, the contusions resulted from significantly lower magnitudes for impacts from the right side in comparison to SDH (p=0.018), but not SAH lesions (p=0.104). Pressure responses were significantly higher for SAH lesions than for SDH (p=0.004) and parenchymal contusions (p=0.04). Impacts to the left side of the head showed significant main effects when the lesions were compared using MPS and VMS (p=0.039 and p=0.025). For impacts to the left side of the head the contusions resulted at lower magnitudes than the SDH lesions (p=0.033) but not the

SAH lesions ($p=0.141$). For VMS, the contusions occurred at significantly lower magnitudes than the SDH ($p=0.023$) but not the SAH ($p=0.075$) for impacts to this location on the head. Finally, for occipital impacts, significant main effects were found for MPS ($p=0.001$), VMS ($p=0.003$), shear stress ($p=0.002$) and shear strain ($p=0.001$) for all. The SDH's occurred at lower magnitudes for occipital impacts than contusions for MPS ($p=0.001$), VMS ($p=0.002$), and shear stress ($p=0.002$) and shear strain ($p=0.001$), but not in comparison to SAH for MPS ($p=0.116$), VMS ($p=0.284$) and shear stress ($p=0.595$). The SDH's were incurred at lower magnitudes for occipital impacts for shear strain than the lesion types (SAH, $p=0.046$; contusion, $p=0.001$).

Discussion

This research investigated the influence of direction on the presence and magnitude of brain stresses and strains for real world TBI lesions arising from falls. The falls were all characterized by slips or trips which led to impacts to the head on hard, non-compliant surfaces. Overall, for the 20 fall reconstructions conducted, there were a total of 20 SDH lesions present, 7 SAH lesions, and 7 contusion lesions present. The distribution in the numbers of lesions suggests that the SDH may be a more common injury for falls to concrete, ice, or other non-compliant surfaces than the other TBI's, particularly for impact velocities that are present for falls from standing height (4 - 6 m/s). In comparison to the existing finite element modelling literature the current research is consistent with the range of magnitudes of brain stresses and strains for TBI in general and different lesion types (table 6)^{15,20}. The current research also produced magnitudes of stress and strain that are consistent with anatomical testing, indicating TBI vascular damage in the range of 0.30 strain or more.

Table 6. Ranges of brain stress and strain found for different TBI lesions from previous finite element modelling research^{15,20}. (kPa: kilopascals)

Lesion type	Pressure (kPa)	MPS	VMS (kPa)	Shear stress (kPa)	Shear strain
Subdural hematoma	55 – 308	0.14 - 0.53	5 - 17	1.8 - 4.75	0.17 - 0.50
Subarachnoid hemorrhage	270	0.34	12.2	4.3	0.40
Contusion	93 – 123	0.16 - 0.23	5 - 8	2.2 - 3.8	0.25 - 0.40

The purpose of this research was to investigate the influence that impact direction has on the presence of particular TBI and the magnitudes of stresses and strains associated with those damaged regions of the brain. This knowledge is important for both the biomedical engineering community as well as clinician. Biomechanically, it is important to know the impact parameters, in this case direction, that are likely to cause a high risk of incurring damage to particular anatomical structures resulting in SDH, SAH, and contusion. For the clinician, information may aid in the assessment of likely outcome injuries from patient descriptions of the head impact event. The results of this research showed that subdural hematomas primarily occurred from impacts to the frontal and occipital regions of the head, which was indicated by the large number of lesions from these impacts (18) in comparison to the frequency from the side impacts (3). The SDH occurred at lower magnitude stresses and strains for many of the dependent variables used to describe the injury, indicating a higher risk of this type of injury for frontal and occipital impacts. These results are consistent with previous research indicating that motions in the sagittal direction (frontal or occipital impacts) lead to strains on the bridging veins that cause a high risk of vascular injury¹²⁻¹³. These strains are caused by the high degree of relative brain/skull motion, motions that would be mitigated by the falx for impacts to the sides of the head¹¹. This may account for why impacts to the sides of the head result in fewer SDHs, and generally required a larger magnitude of brain stress and strain to cause a SDH.

When examining the influence of impact direction on the presence of SAHs there was little agreement between the dependent variables, with only three producing a significant result. Lower magnitude responses for SAH were present for the left side and occipital impacts in comparison to the frontal and right side, but little difference between these impact locations (left side vs occipital; right side vs frontal). The results indicate that unlike SDH, the likelihood of SAH does not seem to be affected by the location of impact on the head for this falling population, which may be indicative of a different mechanism of vascular damage. Like the SAH, there were fewer contusions (7) than SDH, and none of these contusions present from frontal impacts. Three of the dependent variables (MPS, VMS, and shear strain) produced significant results when examining the effects of direction on the presence of parenchymal contusions. Impacts to the sides of the head produced contusions at lower magnitudes in these MPS, VMS and shear strain, than the impacts to the occipital region of the head. This suggests

that movements of the head caused by lateral impacts (side) are more likely to cause contusions for lower magnitude events. It has been proposed that impacts from these directions are more likely to cause movements that result in the brain bumping against the irregular surfaces on the inside of the cranium and causing contusions as previously described in the literature⁵⁰⁻⁵¹. Overall, when examining the sensitivity of brain stress and strain dependent variables used to characterize the effect of impact direction, there was little agreement between different lesion types (SDH, SAH, contusion). Pressure responses in particular produced results that were often opposite to those of the other metrics. However, von Mises stress did produce significant differences for each direction of head motion for SDH, SAH, and contusion injuries suggesting it may be the most sensitive metric for directional analyses of brain injury.

In addition to examining the effect of direction on the presence of TBI types, this research also provided an opportunity to examine if a hierarchy exists among lesions caused by impacts as suggested by Hoshizaki et al⁵². Upon comparison of the different lesions (SDH and SAH) for frontal impacts, the SDH were incurred at lower magnitudes. This result indicates that SDH occur at lower severity impacts, with SAH present when a more severe event has occurred. A similar relationship was present for occipital impacts with SDH and SAH being produced at the lowest magnitudes of response, with parenchymal contusions the product of higher magnitude events. When comparing the different lesions for impacts to the side of the head, the contusions were generally produced at lower brain stress and strain magnitudes than for SDH lesions but not the SAH. This suggests that contusions are more likely to be present for lower severity impacts to the sides of the head than SDH. This analysis supports the theory of a hierarchy between the lesions based on magnitude of brain stress and strain for this dataset, but much more cases need to be analyzed.

Limitations

This research used a variety of biomechanical and computational models as tools for brain injury investigation and is subject to some limitations that must be considered when evaluating the results. These results are biased in that they use a small group of subjects that were presented to hospital from falling events. The small population used is indicative of the difficulty in receiving quality reconstructable cases from hospital datasets. As a result, while these results support many of the previous literature on the influence of impact direction on head injury, the small number of

subjects may limit its application to a broader population. Also it is likely that since the age of the subjects in this research is older adult to elderly, the magnitudes of stress and strain in which these injuries were incurred would be perhaps lower than those for a younger more able bodied group. In addition, it is likely that other mechanisms of injury may produce different results from these analyses. The limitations of using patient reports and questionnaires were discussed in the methodology, and this source of error was controlled as best as possible by applying extremely rigorous inclusion criteria to the dataset. The Hybrid III headform is comprised of a steel head covered with a vinyl skin to simulate human head response, and while it does produce results in the range of cadaveric data⁵³, is not identical to that of a human. In addition, the neckform used is also a simulation of real neck responses. The UCDBTM finite element model represents the current state of the art for mechanical brain injury research and is comprised of material characteristics and geometries from human cadavers, and while it has been subject to, and passed, validation procedures²⁸⁻²⁹, its results may differ from those of live human subjects. The results of this research should be considered in light of these limitations.

Conclusion

The purpose of this research is to examine the relationships between impact direction and the presence of different traumatic brain injury lesions in a human population. This research demonstrated that for this sample of falls causing different TBI lesions that SDH is more likely to occur for impacts to the frontal and occipital regions of the head, and parenchymal contusions for impacts to the side of the head. The SAH that were investigated as part of this study were not characterized by a clear effect of direction, which may be reflective of a unique injury mechanism. These injuries were also shown to occur for events that caused a head impact to hard non-compliant surfaces from standing height (4 – 6 m/s head impact velocity). When examining the brain stress and strain metrics, only von Mises stress was sensitive enough to these impact conditions to produce significant results for each comparison that was made, suggesting this variable may be suitable for this type of brain injury analysis. The data obtained in this research supports the notion that the location of an impact influences the thresholds and types of traumatic brain injury incurred from falls.

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