Investigating the effects of impact directions to improve head injury index

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Abstract

Traumatic brain injury (TBI) is one of the most important causes of death and disability. The objective of this study is to develop new head injury criteria which can predict the greatest principal strain and shear stress in the brain considering the impact directions and magnitudes. So, 150 head impact simulations were performed for 3 magnitudes and 50 directions of impact using head finite element model (FEM). Simulations were performed in order to assess the strain and shear stress created in the brain tissues due to different impact directions and magnitudes. Next, new head injury criteria were developed through performing statistical analysis. The simulation results showed that TBI risks in the sagittal and frontal planes were higher than those with impacts in transverse plane. Furthermore, new brain injury indices were developed to predict maximum principal strain and shear stress in the brain, which had correlation coefficients of 0.85 and 0.89 with head FEM responses, respectively. However, finding of present research showed the effects of impact directions on TBI risks. They also demonstrate that impact magnitude, direction, and duration should be used to develop a brain injury index.

Keywords: Brain injury; Computational modeling; Finite Element Method; Injury Criteria; Safety.

1- Introduction

Traumatic brain injury (TBI) is one of the most important causes of death and disability. TBIs often decreases persons' ability to function and interact socially [1]. Almost 1.7 million cases of TBI per year are reported in the United States[2]. As an example, about 170 000 TBI events occur in Canada every year, which the elderly and children being the frequent victims[3]. These injuries decrease ability and productivity of those who are not able to work. These TBI reports are underestimated because almost 25% of all mild and moderate cases are not included in these medical reports and many concussions do not come under clinical attention. The most important cause of TBI is falls, followed by vehicle events and striking accidents in which a person is hit by an object. TBIs may also occur during military events, vehicle crashes, and sport events [1,4–8]. This study aims to better understand the head injury mechanisms.

Various studies have been performed on brain injury to decrease TBI risks, which is facilitated by understanding those factors playing important roles in causing such injuries, including what characteristics an event should have in order to cause a head injury [9–12]. Head responses during impacts were measured using helmet-mounted accelerometers[13,14]. Moreover, determination of types of head impacts causing such injury can be one of the main areas of research [15–17]. Any types of TBI can be evaluated based on the events by identifying the effects of certain impacts and directions on TBIs. Furthermore, this knowledge will help designers and engineers make new protective devices and safer equipment being used to decrease TBI risks.

Most of injury criteria based on head kinetics are obtained from experimental results; therefore, they have become a basis for many head injury protocols, including vehicle safety standards and helmet design standards[18–22]. Head Injury Criterion (HIC) [18] and Gadd Severity Index (GSI) [19] are applied to predict TBI risks based on weighted integrals of translational acceleration-time during impacts. But, BRain Injury Criterion (BRIC) evaluates head injury using head rotational kinematics [20]. Rowson proposed injury risk curves for TBIs based on translational and rotational impacts based on injury events among collegiate athletes [23]. These head injury criteria are usually calculated to assess TBI risks by using impact time and magnitude without considering the impact orientation. Therefore, these criteria do not show any differences between the input impacts with the different directions

and identical profiles for TBI risks. Some researchers demonstrated that impact directions should be considered as predicting TBI risks because human head is composed of geometries with varying material properties [24–27].

Research done on cadavers and test animals has shown the effects of impact directions on the TBI risks [28]. The use of the animal studies for human subjects has many limitations. Coronal impact effects on the brain reactions were studied by Gennarelli et al [25], who also assessed the effects of sagittal plane impacts on the brain injury. Also, some research has been done by Kleiven [29] and Huang et al [30] to identify the effects of impact directions on TBI risks by using FEM simulations in the sagittal plane. Although some studies have shown impact orientation effects, few studies demonstrated the effects of impact directions on human brain injury considering human anatomical aspects.

Finite element simulations have been used to study the relationships between head kinetics and TBIs. Also, investigation of the brain reactions and prediction of TBI risks have been performed using finite element head models in soccer events, motor vehicle crashes, falls, and other impact scenarios[20,31–33]. Brain stress and strain responses in FEM are very useful tools to study TBI risks [34–39]. Although FEM is a useful tool to study TBI risks by taking impact directions into consideration, it is time-consuming. So, kinematic-based head injury criteria should be proposed to prevent head injury and predict brain stress and stress responses like finite element head models.

In this study, the relationships between brain reactions and head kinematics were evaluated using a 3D FE human head model in different anatomical planes. Therefore, relationships between impact directions and acceleration magnitudes with TBI risks based on maximum principal strain and shear stress were investigated using 150 FE head impact simulations. Finally, statistical analysis was performed to propose maximum principal strain criteria and shear stress criteria for predicting brain reactions.

2- Method

2-1 Head Finite Element Model

Horgan and Gilchrist's head model was applied to create the finite element head model. Their initial head FEM involved brain, specific subarachnoid for cerebro-spinal fluid (CSF), tentorium cerebral, falx cerebri, skull and face. In original Horgan

and Gilchrist's FE head model, skull, facial bone, CSF, dura mater, pia mater, and combined properties of the grey and white matter of the brain were modeled using brick, shell, brick, shell, membrane, and brick elements, respectively. [27,40]. Different anatomic layers were applied to this original model to improve the 3D model in accurate modeling of head layers. The improved model had scalp, three layer skull including cortical and spongy bone, dura mater, trabeculae and CSF under subarachnoid, pia mater, tentorium cerebral, falx cerebri, brain and face[26].

The improved model consisted of 27031 elements including 7107 shell elements and 19924 hexahedral elements. An illustration of improved FEM is given in Fig. 1. Detailed mechanical properties of all components of this head model are illustrated in Table 1.

Cortical and spongy bone, dura mater, arachnoid and pia mater were defined to Abaqus software (Abaqus is a commercial software package for finite element analysis, which developed by Dassault Systemes used in various disciplines of Engineering) by Prony series to control large deformations[27]. Other parts of the model were considered as being linear elastic. For viscoelastic section, the relaxation shear modulus $G_R(t)$ was introduced in ABAQUS software by dimensionless function $g_r(t)$ expressed by Prony series by Equation 1. Zhao work group [41] has developed the following Equation 1:

$$g_r(t) = 1 - \sum_{i=1}^{N} G_i \left[1 - \exp(-t/\tau_i)\right] \quad (1)$$

Where G_0 is the instantaneous shear modulus, $g_r(t) = G_R(t)/G_0$ and τ_i is stress relaxation time. When the time t takes an infinite value, long time shear modulus is calculated as Equation *2*[26].

$$G_\infty = G_R(\infty) = g_R(\infty) \times G_0 \quad (2)$$

The improved model was validated by the results of two research works by Nahum including impact along the mass center of the head and by Trosseille including off center impact in order to validate brain reactions. Six samples of Nahum test (1977) were applied to validate pressure prediction in finite element head model [42]. Moreover, the Trosseille MS 428_2 test (1992) was applied to validate the brain response with the cadaver that was in a seated position and impacted by a 23.4 kg hammer with 7 m/s velocities in the anterior posterior direction[27,43].

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160 **2-2 Injury Criteria for traumatic brain injury (TBI)**

161 The injury criteria for TBI were divided into those that relate to the motion of
162 the whole head (macro scale) and those that relate to the deformation of tissues
163 within the head (tissue level). The macro-scale criteria are measured in experimental
164 studies; therefore, they have become a basis for numerous head injury events.
165 Tissue level thresholds have been measured using finite element models[22].

166 HIC (Head injury criteria) is one of the most important macro-scale injury
167 criteria that was improved and advanced upon SI (Severity Index) by Versace
168 (1971)[18]. It is the current injury index for head injury used in the FMVSS 208
169 standard. The HIC equation is defined as equation 3.

$$HIC = \max\left(\left(t_2 - t_1\right)\left(\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a\left(t\right) dt \right)^{2.5} \right) \tag{3}$$

170 Where *a(t)* is the resultant linear acceleration at the head center of
171 gravity, t_1 and t_2 are the initial and final times, respectively, by which HIC is
172 calculated (t_1 and t_2 are selected to maximize HIC). The HIC value should be limited
173 below 1000 for preventing serious brain injury[18].

174 Several tissue level injury criteria have been proposed to predict injury in brain
175 finite element models. Most of these criteria have been investigated by
176 reconstructing real life accidents using anthropomorphic test devices and finite
177 element head models. These reconstructed brain trauma events include falls, hockey
178 collisions, pedestrian accidents, soccer events, and vehicle accidents, and crashes.
179 Developed injury criteria usually use stress and strain values. Although FE head
180 models have different anatomical structures with different injury criteria, the similar
181 value of brain injury tolerances have been proposed to predict TBI risks[27]. Head
182 tissue injury criteria including maximum principal strain and shear stress were
183 applied to investigate head trauma in this study. These criteria were selected based
184 on previous research, in which they were proposed in order to determine brain
185 injuries [44–47]. The proposed tolerance limit for a risk of head trauma is about
186 7.5kPa (KiloPascal) and 0.25 for shear stress and maximum principal strain,
187 respectively.

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2-3 Computational Simulations

This study was performed using the FE head model having multiple anatomic parts with various material properties and validated by the results of previous experimental data. Simulations were carried out in Abaqus software using the Dynamic Explicit method. They were performed with three different magnitude impact pulses applying about 50 different directions of impact, resulting in 150 simulations (Tables 2, 3). using a cube as an example, translational impacts were applied on each of the following 50 vectors: 8 vectors from the center to the middle of each cube face, 8 vectors from the center to the cube corners, 12 vectors from the center to the midpoint of each cube edge, and 24 vectors from the center to the middle between the middle of each cube face and each cube edge (Figs. 2; Table 3)[27]. The spherical coordinate system (r, θ, φ) was used to describe the impact orientations, as illustrated in Figure 2, where Z and X axes were aligned with superior and anterior directions in the head, respectively. Translational accelerations were applied to the head center of gravity as loading conditions on the skull being assumed rigid. The input load to the FE head model was an acceleration [27,42] shown in Fig. 3, which was applied with different magnitudes and directions. The model was not allowed to have a rotation in order to isolate the effects of translational direction and magnitude on response of the brain model.

2-4 Post-Processing of Simulations

Effects of orientation and peak linear impact on the brain responses (including principal strain and shear stress) were evaluated. Seventeen pairs of impact directions were symmetric relative to the FE head model due to the symmetry across the sagittal plane, resulting in 33 unique impact directions as shown in Table 3 [27]. To simplify the reporting of the results, only 33 unique impact orientations were evaluated while studying the principal strain and shear stress in the brain. Calculations were run in order to indicate the principal strain and shear stress responses in brain tissues due to changing the impact directions. Finally, statistical analysis was performed on the proposed new head injury metrics to predict brain reactions.

3- Results

3-1 Effects of impact parameters on brain greatest principal strain

The brain response diagrams show that head impact orientation has a significant effect on the brain greatest principal strain. The greatest of the maximum principal strain in the sagittal, frontal, and transverse planes occurred in the superior, superior, and posterior directions, respectively, while the minimums occurred in the inferior, inferior, and anterior directions, respectively. This finding shows that changing the impact direction in the sagittal, frontal, and transverse planes leads to changes up to 5.08, 2.89 and 1.70 times in the greatest principal strain (Figs. 4-6). Impact directions change the greatest principal strain since the head has multiple structures with varying material properties and geometries. Also, the most sensitive orientation for changing the greatest principal strain is the inferior direction in the sagittal and frontal planes because of the higher degree of relative brain/skull motions in that diction compared to others. The significantly higher value of the greatest principal strain were observed in sagittal plane simulation compared to others (Fig. 4). Value of the greatest principal strain values were lower in simulations with a pulse in the transverse plane than in others with no pulse in the transverse plane (Fig. 5). Maximum of the greatest principal strain was obtained at 0.20 in the sagittal plane, 0.20 in the frontal plane, and 0.14 in the transverse plane for the highest peak impact (R3) simulations (Figs. 4-6). For the greatest principal strain, some similar trends were calculated in three input peaks by changing impact directions.

3-2 Effects of impact parameters on the brain shear stress

Diagrams of the brain maximum shear stress in different planes and orientations illustrate that impact direction had a considerable effect on the brain shear stress. The gap sizes between three input magnitudes for one direction of impact in Fig. 7 show a non-linear relationship between shear stress and input peaks. Figs. 7-9 demonstrate the maximum brain shear stress in the sagittal, frontal, and transverse planes occurred in the inferior, anterior, and inferior directions while the minimums' occurred in the superior, right-left, and superior directions, respectively. Changing the impact orientations in the sagittal, frontal, and transverse planes led to changes up to 13.44, 9.06 and 10.21 times in the brain shear stress. Furthermore, sagittal plane had the highest sensitivity to changing impact directions, which was observed when the direction was changed from superior to inferior one. However, considerably greater values of maximum brain shear stress were seen in

simulations with the transverse plane impacts compared to those with transverse and frontal plane. These regional differences in shear stress with varied impact directions are likely due to the geometry and specific anatomy in the head model. Considerably higher values of shear stress were observed in simulations with sagittal plane impact compared to those with other planes (Fig. 7). The shear stress values in the transverse plane impact simulations were lower than those in simulations with no pulse in the transverse plane (Fig. 8). Maximum shear stress was obtained at 7.29 k Pa in the sagittal plane, 7.29 k Pa in the frontal plane, and 3.49 k Pa in the transverse plane for the highest peak impact (R3) simulations. (Figs. 7-9). For shear stress, similar trends were seen in three input magnitudes by changing impact orientations.

3-3 Development of new improved head injury criteria

Brain injury risk, as calculated from brain MPS and shear stress, varied with impact directions and increased with an increase in the input peak for MPS and shear stress indices for any given impact directions. However, injury risk for MPS or shear stress varied with translational directions when the pulse magnitude was controlled (Table 4). The average injury risks for the highest magnitude (R1) simulations for the brain injury index were 2.77 kPa for the shear stress and 0.11 for MPS. Maximum injury risk from all simulations at the highest input peak (R1) was 7.29 for brain maximum shear stress and 0.20 for brain MPS injuries. The lowest variation was observed with MPS risk at magnitude R3 with an injury risk differential of 0.10. In contrary, the highest differences were seen in shear stress risk at magnitude R1 with an injury risk differential of 6.95 kPa. Variation in injury risks due to the impact directions increased with an increase in the input magnitude. Also, variations of shear stress due to changing the impact directions were higher than brain MPS variations.

Non-linear relationships were seen between HIC values and input magnitudes for gap sizes of three peak pulses in each impact direction shown in Table 2. The results of the head impact FEM simulations showed that head impact directions could have a significant effect on the brain reactions. Moreover, they revealed that TBI risks can be increased by changing the impact magnitudes and HIC levels. So, effective factors were evaluated based on the findings from brain reactions (maximum brain shear stress and normal strain) by performing statistical analyses

on three different magnitude linear accelerations applying about 50 different orientations, resulting in 150 simulations (Table 3). Then, the simulation results were studied and new head injury indices were proposed in the equations 4 and 5 using the Design Expert software. These indices included Shear Stress Criteria (equation 4) and Maximum Principal Strain Criteria (equation 5), predicting shear stress and maximum principal strain, respectively. In these equations, *A* and *B* are defined as $cos\,\varphi$ and $cos\,\theta$ respectively. Results from regression analysis showed high correlations between FEM simulation results and proposed brain injury indices (Table 5). These equations predicted brain responses for head FEM with material properties indicated in table 5.

$$SSC = ShearStressCriteria = \frac{\left(0.0129HIC^{0.57}\right)\times 100}{7.5}\times \left[1.11-1.64A-1.36B+0.67AB+4.05A^{2}+0.73\times B^{2}\right] \quad (4)$$

$$MPSC = MaximumPrincipalStrainCriteria = \frac{\left(0.076HIC^{0.03}\right)\times 100}{0.25}\times \left[0.11+0.078A-0.029B+0.025A^{2}+0.029A^{2}B-0.016A^{3}+0.011B^{3}\right] \quad (5)$$

4- Discussion

Different layers and material properties composing head geometry may affect the brain response values during impacts with different loading directions. Present research indicates that the highest brain stress and pressure do not occur in specific head impact directions. Most of the injury criteria such as HIC [18] and BRIC [20] predict TBI injury risks only through with peak accelerations, not through the direction applied pulses. So, the risk of injury for inputs with the same acceleration profiles will be the same regardless of impact directions. Thus, this study used FEM simulations to investigate the effects of impact orientations and peaks for TBI injury risks using FEM head simulations. The results clearly showed that impact directions had significant effects on the brain reactions and head injury risks. Therefore, new brain injury indices should take impact directions into consideration. New head injury criteria were developed in present research using HIC to predict intracranial responses and TBI risks considering impact directions, magnitudes and time. Moreover, previous head injury criteria could not determine how inputs affected brain reactions, only being capable of assessing TBI risks. But, newly developed head

injury criteria, including MPSC and SSC, can be applied to determine the maximum principal strain and shear stress in different head impact incidents. Findings of Previous studies are consistent with the results of the present research, showing that motions in the sagittal direction exert strains and stresses on the brain, causing high TBI risks [35]. These stresses and strains are created by a high degree of relative brain/skull motions. This is why impacts on the head sides result in fewer TBI risks and, in general, need higher pulse magnitudes to cause TBIs.

5- Conclusion

Relationships between brain reactions and orientations of head impacts were investigated using a 3D finite element head model with maximum principal strain and shear stress metrics. Results indicated that impact directions should be considered in order to determine head injury risks. Moreover, they showed that stress and strain imposed on the brain in sagittal and frontal planes were higher than those in the transverse planes. Also, analysis of head linear accelerations from different directions showed that Head Injury Criteria (HIC) should be developed for predicting TBI risks. So, new brain injury criteria were proposed using statistical analysis and head FEM to predict the brain reactions. Maximum Principal Strain Criteria (MPSC) and Shear Stress Criteria (SSC) had correlation coefficients of 0.85 and 0.89 with finite element results, respectively. The findings of this research can be used to better understand the relationships between head impact orientations with stress and strain imposed on the brain. Finally, protective equipment and safety systems may be improved to protect humans against head impact events using the results of this research.

Conflicts of interest

The authors declare that they have no conflicts of interest.

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Figure Captions

Figure 1. Developed finite element head model[27].

Figure 2. Directions of the input impacts, showing spherical coordinate system.

Figure 3. Kinematic input shape for translational accelerations.

Figure 4. The effects of impact directions on maximum brain normal strain in the sagittal plane.

Figure 5. The effects of impact directions on maximum brain normal strain in the transverse plane.

Figure 6. The effects of impact directions on maximum brain normal strain in the frontal plane.

Figure 7. The effects of impact directions on maximum brain shear stress in the sagittal plane.

Figure 8. The effects of impact directions on maximum brain shear stress in the transverse plane.

Figure 9. The effects of impact directions on maximum brain shear stress in the frontal plane.

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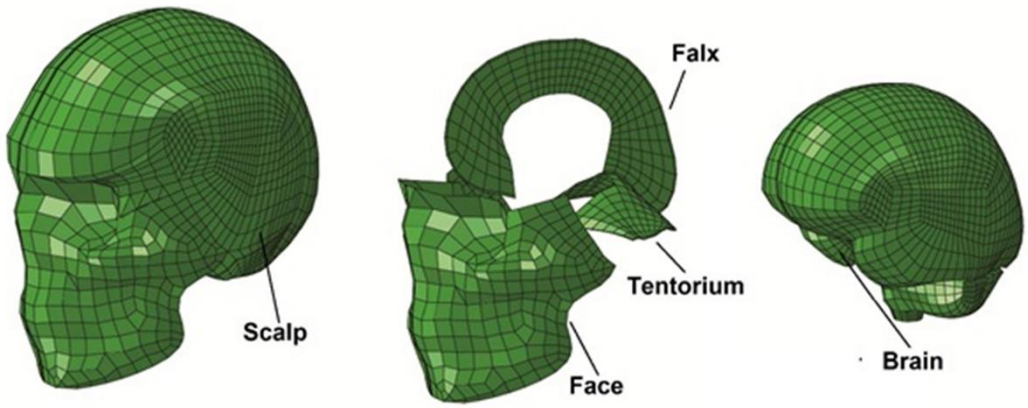

536 **Figure 1.** Developed finite element head model[27].

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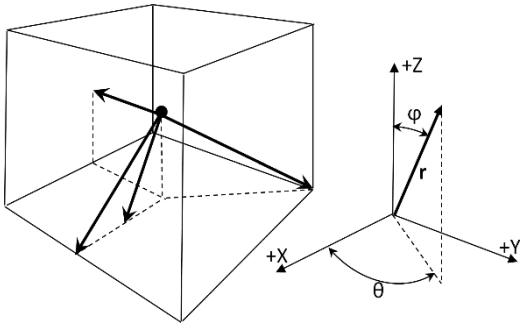

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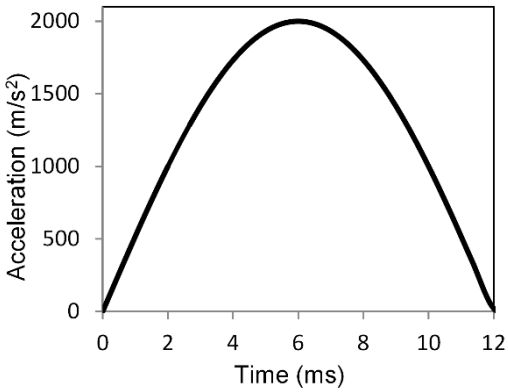

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544 **Figure 3**. Kinematic input shape for translational accelerations.

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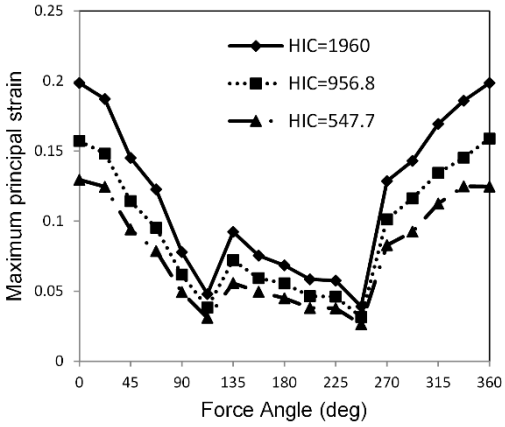

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Figure 4. The effects of impact directions on maximum brain normal strain in the sagittal plane.

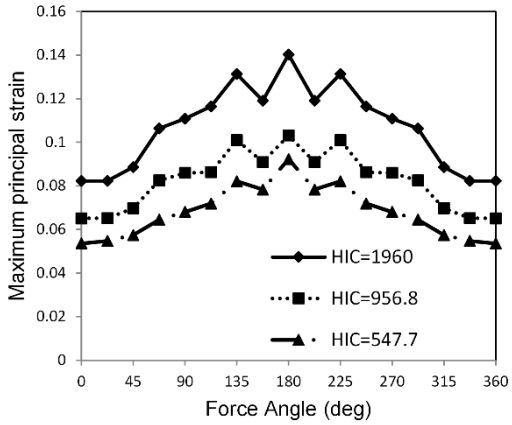


Figure 5. The effects of impact directions on maximum brain normal strain in the transverse plane.

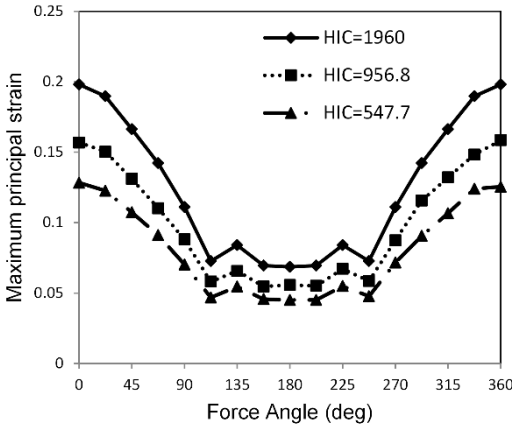


Figure 6. The effects of impact directions on maximum brain normal strain in the frontal plane.

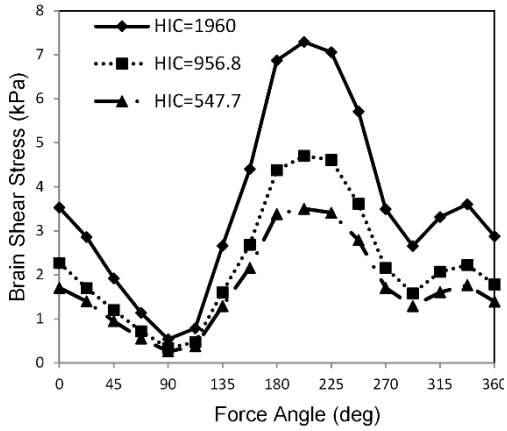
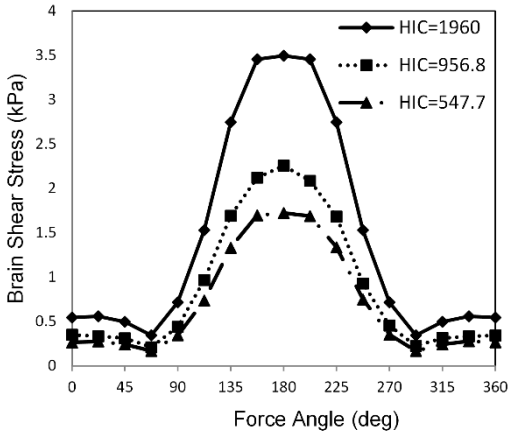


Figure 7. The effects of impact directions on maximum brain shear stress in the sagittal plane.



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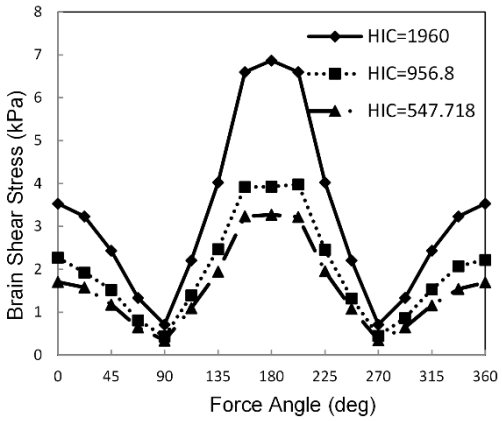

Figure 9. The effects of impact directions on maximum brain shear stress in the frontal plane.

Table 1. Mechanical properties of components of finite element head model[27].

Layer	Behavior	Thickness (mm)	Density (kg/m3)	Long-term elastic modulus (MPa)	Bulk modulus (MPa)	Poisson's ratio	Viscoelastic response ($g_R(t)$)
Skin	Elastic	5	1130	16.7	35	0.42	-
Cortical bone	Viscoelastic	4	2000	13000	8929	0.22	$1-0.0293\left(1-e^{-t/9}\right)-0.0656\left(1-e^{-t/950}\right)-0.0278\left(1-e^{-t/9500}\right)-0.107\left(1-e^{-t/90.000}\right)$
Trabecular bone	Viscoelastic	2	1300	888	740	0.3	$1-0.6224\left(1-e^{-t/711.23}\right)-0.2143\left(1-e^{-t/4267.4}\right)$
Dura mater	Viscoelastic	0.4	1140	11.72	7	0.23	$1-0.1088\left(1-e^{-t/40}\right)-0.0959\left(1-e^{10.00}\right)-0.0922\left(1-e^{-t/1000.00}\right)$
Arachnoid mater	Viscoelastic	0.35	1130	19.32	64	0.45	$1-0.919\left(1-e^{-t/0.002}\right)$
CSF	Elastic	0.15	1000	0.015	2273	0.499989	-
Pia mater	Viscoelastic	0.15	1130	19.32	64	0.45	$1-0.919\left(1-e^{-t/0.002}\right)$
Trabeculae	Elastic	1.5	1130	0.050	22	0.48	-
Brain	Viscoelastic	-	1040	0.0228	2278	0.49998	$1-0.815\left(1-e^{-t/0.0143}\right)$

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Table 2. HIC values and the peaks of linear input impacts.

Input	Peak of linear acceleration (m/s^2)	HIC value
R1	2000	1960
R2	1500	956.7
R3	1200	547.7

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Table 3. Unit vectors of applied impact directions[27].

Run	Anatomical direction	Direction	ϕ	θ	x	y	z
1	Anterior	x	90	0	1	0	0
2	Posterior	x(-)	90	180	-1	0	0
3	Left	y	90	90	0	1	0
3*	Right	y(-)	90	-90	0	-1	0
4	Superior	z	0	0	0	0	1
5	Inferior	z(-)	180	0	0	0	-1
6	Anterior- Left	x, y	90	45	0.707	0.707	0
6*	Anterior- Right	x, y(-)	90	-45	0.707	-0.707	0
7	Posterior- Left	x(-), y	90	135	-0.707	0.707	0
7*	Posterior- Right	x(-), y(-)	90	-135	-0.707	-0.707	0
8	Anterior- Superior	x, z	45	0	0.707	0	0.707
9	Posterior- Superior	x(-), z	45	180	-0.707	0	0.707
10	Anterior- Inferior	x, z(-)	135	0	0.707	0	-0.707
11	Posterior- Inferior	x(-), z(-)	135	180	-0.707	0	-0.707
12	Superior- Left	y, z	45	90	0	0.707	0.707
12*	Superior- Right	y(-), z	45	-90	0	-0.707	0.707
13	Inferior- Left	y, z(-)	135	90	0	0.707	-0.707
13*	Inferior- Right	y(-), z(-)	135	-90	0	-0.707	-0.707
14	Anterior- Left	x, y	90	22.5	0.923	0.382	0
14*	Anterior- Right	x, y(-)	90	-22.5	0.923	-0.382	0
15	Anterior- Left	x, y	90	67.5	0.382	0.923	0
15*	Anterior- Right	x, y(-)	90	-67.5	0.382	-0.923	0
16	Posterior- Left	x(-), y	90	112.5	--0.382	0.923	0
16*	Posterior- Right	x(-), y(-)	90	-112.5	-0.382	-0.923	0
17	Posterior- Left	x(-), y	90	157.5	-0.923	0.382	0
17*	Posterior- Right	x(-), y(-)	90	-157.5	-0.923	-0.382	0
18	Anterior- Superior	x, z	22.5	0	0.382	0	0.923
19	Anterior- Superior	x, z	67.5	0	0.923	0	0.382
20	Posterior- Superior	x(-), z	22.5	180	-0.382	0	0.923
21	Posterior- Superior	x(-), z	67.5	180	-0.923	0	0.382
22	Anterior- Inferior	x, z(-)	112.5	0	0.923	0	-0.382
23	Anterior- Inferior	x, z(-)	157.5	0	0.382	0	-0.923
24	Posterior- Inferior	x(-), z(-)	112.5	180	-0.923	0	-0.382
25	Posterior- Inferior	x(-), z(-)	157.5	180	-0.382	0	-0.923
26	Superior- Left	y, z	22.5	90	0	0.382	0.923
26*	Superior- Right	y(-), z	22.5	-90	0	-0.382	0.923

27	Superior- Left	y, z	67.5	90	0	0.923	0.382
27*	Superior- Right	y(-), z	67.5	-90	0	-0.923	0.382
28	Inferior- Left	y, z(-)	112.5	90	0	0.923	-0.382
28*	Inferior- Right	y(-), z(-)	112.5	-90	0	-0.923	-0.382
29	Inferior- Left	y, z(-)	157.5	90	0	0.382	-0.923
29*	Inferior- Right	y(-), z(-)	157.5	-90	0	-0.382	-0.923
30	Anterior- left- Superior	x, y, z	45	45	0.577	0.577	0.577
30*	Anterior- Right -Superior	x, y(-), z	45	-45	0.577	-0.577	0.577
31	Anterior- left- Inferior	x, y, z(-)	135	45	0.577	0.577	-0.577
31*	Anterior- Right -Inferior	x, y(-), z(-)	135	-45	0.577	-0.577	-0.577
32	Posterior - left- Superior	x(-), y, z	45	135	-0.577	0.577	0.577
32*	Posterior-Right -Superior	x(-), y(-), z	45	-135	-0.577	-0.577	0.577
33	Posterior- left- Inferior	x(-), y, z(-)	135	135	-0.577	0.577	-0.577
33*	Posterior- Right-Inferior	x(-), y(-), z(-)	135	135	-0.577	-0.577	-0.577

* marks inputs that are symmetric relative to sagittal plane.

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Table 4. Summary of brain injury index grouped by input peak.

Injury index	Input magnitude	Ave	SD	Min	Max	Differential
Shear stress (kPa)	R1	2.77	2.02	0.34	7.29	6.95
	R2	1.72	1.25	0.21	4.70	4.49
	R3	1.35	0.98	0.17	3.50	3.34
MPS	R1	0.11	0.04	0.04	0.20	0.16
	R2	0.09	0.03	0.03	0.16	0.13
	R3	0.07	0.03	0.03	0.13	0.10

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Table 5. Coefficient of determinations (R_2) and p-values between the brain reactions and developed brain injury.

Head injury criteria	Correlation coefficient	p value
MPSC	0.85	<0.001
SSC	0.89	<0.001

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