The influence of impact angle on the dynamic response of a Hybrid III headform and brain tissue deformation

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ABSTRACT

The objective of this study was to investigate the influence of impact angle on the dynamic response of a Hybrid III headform and brain tissue deformation by impacting the front and side of the headform using four angle conditions (0°, at the impact site and 5, 10 and 15° rightward rotations of the headform from 0°) as well as three additional angle conditions of -5, -10 and -15° (leftward rotations from 0°) at the side location to examine the effects of the neckform. The acceleration-time curves were used as input into a finite element model of the brain where maximum principal strain was calculated. The study found that an impact angle of 15° significantly influences the results when measured using linear and rotational acceleration and maximum principal strain. When developing sophisticated impact protocols and undertaking head injury reconstruction research, it is important to be aware of impact angle.

KEYWORDS: Head impact angle, Hybrid III headform, Finite element analysis,
1 INTRODUCTION

Concussion affects a large portion of the population with over two million people affected annually [1]. The long-term consequences of concussion have been linked to cognitive dysfunction that can affect a person’s daily life at work or school [2, 3]. Concussion can be caused by an impact causing motion of the head and brain, where energy from the impact is transferred to the brain causing injury [4, 5]. Past studies have established a link between head motion and the resulting brain injury [6, 7] and researchers have used engineering metrics such as head acceleration and brain tissue deformation metrics to quantify the severity of head impacts [8].

A head impact event can be described using parameters such as impact mass, velocity, compliance, location and angle. Past research has found that each of these variables affect the results of a head impact. Head impact mass and velocity have been shown to vary the magnitudes of acceleration and impact force experienced by the headform with higher masses and velocities increasing the risk of head injury [9, 10]. Factors affecting the compliance of the impact condition have also been shown to be important when studying a head injury event. In past research, some components of compliance in a head injury event could be manipulated by changing the stiffness of the materials involved in the impact, such as the energy absorbing liners in helmets. These liners contribute to influencing the amount of impact energy transferred to the head, which can reduce the risk for injury [10, 11].

The results of a head impact have been shown to be sensitive to impact location. This has been found from experimental research using animals as well as from computational research using finite element analysis, where the impact location changes the severity and duration of
concussive symptoms in animals[6] and the magnitudes of brain tissue deformation in finite element models of the human brain[12].

One study combined both impact location and angle in producing centric (through the center of gravity) and non-centric impacts to the headform and found that linear dominant head accelerations were mainly from centric impacts and rotationally dominant accelerations were created by non-centric (not through the center of gravity) impacts [13]. These types of impact conditions have been found to influence the magnitudes of headform acceleration and brain tissue deformation contributing to the risk of brain injury [13, 14]. Since impact location has been shown previously to be an important variable when examining the outcome of a head impact, research investigating the influence of impact angle is less clear. Studies in the past have described impact angle as a variable in a combination with location, while other studies have altered impact angle as a variable in a series of parametric studies as part of the head injury reconstruction process [15, 16].

When examining these studies in light of impact angle, the variation of the angle of the incoming objects or projectiles often created conditions where a different site on the head was impacted. For example, when orienting a hemispherical impactor (Figure 3) to create non-centric conditions, the center of the impactor was directed at one location even though the side of the impactor came into contact with a different location on the headform [13]. Therefore, isolating the effect of impact angle on the headform response was difficult. Consequently, the objective of this present study was to isolate impact angle as a variable by impacting the same location on a headform at different incoming angles. Peak resultant linear and rotational accelerations were measured from the headform dynamic response and were used as input into a finite element model of the brain to estimate magnitudes of maximum principal strain.
2 METHODOLOGY

2.1 Equipment

A pneumatic linear impactor system was used to deliver impacts to a headform. The linear impactor system was composed of a steel frame, piston, air tank and impactor arm (mass: 13.1 kg). The impactor arm was propelled forward using compressed air and was covered with a pointed striker that had a layer of modular elastomer programmed (MEP). The MEP layer was 25.0 ± 1 mm thick and was placed between a 25.0 ± 1 mm thick vinyl layer base and a vinyl rounded tip (Figure 2). This striker was different from the hemispherical striker used in past research [ANNA, I THINK YOU NEED A REFERENCE OR TWO HERE TO THE PARTICULAR PAST RESEARCH YOU MEAN] because it had a much narrower cap that allowed the same location on the headform to be impacted at different angles [13]. A sliding table located at the end of the impactor arm attached the headform to an adjustable and lockable base. This allowed the headform to be oriented and remain fixed in space during testing. This feature allowed the headform to be oriented in five degrees of freedom: anterior-posterior (x), lateral (y) and superior-inferior (z) translation as well as anterior-posterior (y) and longitudinal rotation (z) with positive x, y and z axes moving forwards, leftwards and upwards.

A 50th percentile adult male Hybrid III head and neckform were used to collect the dynamic response of the head. The surrogates were composed of a combination of steel, aluminum, butyl rubber and vinyl where the headform weighed 4.54 ± 0.01 kg and the neckform weighed 1.54 ± 0.05 kg. The headform was equipped with 9 single-axis Endevco 7264C-2KTZ-2-300 accelerometers that were arranged in a 3-2-2-2 array to collect headform acceleration data [17].
2.2 Data Collection

Linear and rotational acceleration data was collected at a rate of 20 kHz when the headform acceleration exceeded a threshold level of 3 g. This data was recorded on a personal computer using TDAS Pro Lab computer software (Diversified Technical Systems, Seal Beach, California). An electronic time gate was used to calculate impact velocity by measuring the time it took for a 0.02525 m flag to pass through a laser beam. This data was captured and analyzed using National Instruments VI-Logger (Austin Massachusetts, Texas) and Bioproc 2 (developed by D.G.E. Robertson, University of Ottawa).

2.3 Procedure

The headform was impacted at the front and side location using 4 angle conditions: a 0° condition (perpendicular to the impact location) and a 5, 10 and 15° condition (rightward rotation of the headform in the transverse plane from the 0° condition) (Figure 1). Three additional impact angle conditions were used at the side location, -5, -10 and -15° leftward rotation of the headform to evaluate the influence of the neckform. Impact site accuracy was ensured prior to each impact condition through the use of a laser that was mounted to the center of the impactor arm and that matched marked locations on the headform (front and side 0° conditions). Care was taken to ensure that the first point of contact between the apex of the striker and the headform was at the same location for all angles. The headform was impacted three times per condition at a velocity of 5.5 m/s for a total of 33 impacts.

2.4 Finite element model

The finite element model used in this study was the University College Dublin Brain Trauma Model (UCDBTM). Discretization of the model was done using computed tomography (CT) and magnetic resonance imaging (MRI) scans of a male human cadaver [18]. The model is
composed of approximately 26 000 elements constituting the scalp, skull, pia, falx, tentorium, white and grey matter as well as cerebrospinal fluid (CSF) [18]. To approximate the slip layer of the CSF, elements representing this component were solid elements with a lower shear moduli to allow shear motion between the skull and brain [18]. The characteristics of the different tissues of the brain were taken from the literature and are listed in Table 1 and Table 2 [18]. The linear viscoelastic behavior of the brain tissue is represented with the following shear modulus equation:

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

where $G_\infty$ and $G_0$ represent the long- and short-term moduli and $\beta$ represents the decay factor of the material [18]. A unique $G_\infty$ and $G_0$ constant for each tissue is representative of the time dependent response of that particular tissue to shear loading. The brain model was validated against the intracranial response data of selective cadaver head impacts and relative brain and skull surface motion results [19, 20].

The three-dimensional linear and rotational acceleration-time curves from the bare headform impacts were used as input into the finite element model to estimate brain tissue deformation. The brain tissue deformation metric used in this study was maximum principal strain (MPS), which is a measure that has been used in the past [21, 22] to quantify brain deformation associated with head impacts.

### 2.5 Statistical Analyses

A total of nine one-way ANOVAs were used to compare the effect of impact angle on each of the dependent variables (peak linear and rotational acceleration and maximum principal strain) at the front and side locations. Six one-way ANOVAs were used to analyze the dependent variables for the positive impact angles at the front and side, and three additional one-way
ANOVAs were used to analyze the negative angles at the side location. The alpha level was set to $p<0.05$ in all tests.

3 RESULTS

Table 3 lists the results for peak resultant linear and rotational accelerations as well as maximum principal strain for all impact angle conditions. For peak resultant linear acceleration, significant differences were found at the side location for the negative impact angles where the $-15^\circ$ condition (174.5 g) produced significantly higher linear accelerations than the $-5^\circ$ condition (164.5 g). A different relationship was found for rotational acceleration. At the front location, the 10 and 15° conditions (8778 rad/s$^2$ and 9939 rad/s$^2$, respectively) produced significantly higher rotational accelerations than the 0 and 5° conditions (8083 rad/s$^2$ and 8282 rad/s$^2$, respectively) where the 15° condition was also significantly higher than the 10° condition. For the side location, no differences were found amongst the positive impact angles; however for the negative angles the $-15^\circ$ condition (11 964 rad/s$^2$) had higher rotational accelerations than the 0 and $-5^\circ$ conditions (11 067 rad/s$^2$ and 11 0642 rad/s$^2$, respectively). For maximum principal strain, the 15° condition (0.610) had higher values than the 0 and 5° condition (0.515 and 0.496 respectively) at the front location. At the side location, the positive 15° condition (0.568) had higher MPS values than the 10° condition (0.527) and for the negative impact angles both the -10 and -15° conditions (0.602 and 0.634) produced significantly higher MPS values than the 0 and -5° conditions (0.557 and 0.561 respectively).
4 DISCUSSION

The objective of this study was to examine the effect of impact angle on headform dynamic response and brain tissue deformation. The results of this study found that impact angle had an effect on linear and rotational accelerations as well as maximum principal strain. When measuring head impact angle using peak resultant head acceleration, the largest impact angle condition (15°) produced significantly higher linear and rotational accelerations for the side location amongst the negative impact angles. A similar trend was found for the front location where the larger impact angles (10 and 15°) increased the rotational response of the headform as compared to the smaller angles (0 and 5°). This trend was not found for the positive angles at the side location for either linear or rotational acceleration. Characteristics of the head and neckform and how they interact to an impact could influence the dynamic response results found in this study.

The Hybrid III headform does not have a uniform mass distribution, which would affect the moment of the inertia of the headform. The moments of inertia of a Hybrid III headform are $1.590 \cdot 10^{-2}$ kg·m$^2$ in the x-axis, $2.400 \cdot 10^{-2}$ kg·m$^2$ in the y-axis and $2.200 \cdot 10^{-2}$ kg·m$^2$ in the z-axes, with positive axes oriented frontwards, leftwards and upwards respectively [23]. A lower moment of inertia in the x-axis would contribute to the larger magnitudes of rotational acceleration seen for side impacts relative to frontal impacts.

Characteristics of the neckform could have contributed to affecting the dynamic response as well. The Hybrid III neckform has slits on the anterior portion of the neck which allow for different neck responses in flexion than extension [24]. Under quasi-static bending tests, frontal loading (extension) was found to be more compliant than lateral loading (side bending, and rear loading (neck flexion) [25]. These anisotropic characteristics of the neckform have been found to
contribute to influencing the magnitudes of headform acceleration [25]. It is also possible that the slits on the anterior portion of the neck cause the headform to move in a similar motion across all the impact angle conditions when impacted at the front location which would explain why no significant differences were found for linear acceleration. However, when impacting the headform at the side location using the negative impact angles, the effect of these slits on the motion of the head in a combination with the different headform moments of inertia in the x, y and z-axes could contribute to the differences in acceleration values found in this study.

When examining the effect of impact angle on maximum principal strain, a similar trend exists as for the headform acceleration, where a larger impact angle increased the values of maximum principal strain. As impact angle increases, the impact vector becomes oriented outside of the center of gravity of the headform, most likely producing a rotationally dominant response. Past research [14] evaluating the relationship between headform acceleration and brain tissue deformation variables has reported that rotational acceleration is moderately correlated with maximum principal strain (r = 0.64) as compared to linear acceleration, which had a low negative correlation (r = -0.24).

The magnitudes of linear and rotational acceleration and maximum principal strain found in this study are well above proposed thresholds for concussive injury. An 80% probability of concussion has been proposed to occur at linear and rotational accelerations of 106 g and 7900 rad/s² respectively and a strain level of 0.24 [26]. This would make sense as this study delivered impacts directed at an un-helmeted headform, causing high levels of headform dynamic response and thus high levels of brain tissue deformation.

This study found that impact angle influences the results of a head impact; however these relationships were different when measured using linear and rotational accelerations and
maximum principal strain. When developing sophisticated impact protocols and undertaking head injury reconstruction research, it is important to be aware of impact angle as a variable influencing the impact condition. ANNA; I THINK THIS SENTENCE IS A LITTLE TOO GENERAL. THE WAY IT READS AT PRESENT, IT SEEMS LIKE YOU ARE SIMPLY STATING SOMETHING OBVIOUS. PERHAPS YOU COULD QUALIFY THIS WITH A BIT MORE SPECIFIC DETAIL?

5 CONCLUSION

The objective of this study was to investigate the influence of head impact angle on the response of a Hybrid III headform and brain tissue deformation. This study found that a 15° impact angle produced significantly higher results [ANNA – I SUGGEST YOU EXPLICITLY MENTION WHAT WAS “HIGHER”, AND PERHAPS ALSO MENTION BY HOW MUCH HIGHER WAS IT] when compared to lower angle variations but was influenced by the dependent variable used for analysis. It is important to be aware of impact angle [SEE COMMENT IN PREVIOUS PARAGRAPH] when undertaking head injury reconstructions or when developing sophisticated centric and non-centric impact protocols.

6 LIMITATIONS

The Hybrid III head and neckform are limitations of this study because they are used as human surrogates to head impact. These surrogates are composed of a combination of metal and rubber and may not be representative of a living human’s response to impact. Similarly, the finite element model used in this study is also a limitation because it estimates human brain tissue deformations based on a number of assumptions regarding the behavior of brain tissue properties and its validation. The finite element model was validated against cadaver head impact research
and may not provide an accurate estimation of living human brain tissue deformation to impact, which would make it difficult for use in predicting brain injury. The pointed impactor striker was also a limitation of this study because the geometry of the striker limited the maximum variation of impact angle to 15°. However, impacts directed beyond 15° using this striker created conditions where the side of the striker impacted the headform which did not allow for impact angle as a variable to be isolated and studied. To obtain more impact angle variations, a more acutely pointed striker would be needed; however, the more narrow the tip of the impactor cap becomes, the mechanism of injury becomes more of a penetration type of injury.

7 DELIMITATIONS

This study was delimited to the 50th percentile adult male Hybrid III headform. Although female versions of the headform have been developed, the male headform was readily available in laboratory for use in this study. While different headforms would produce characteristic responses to impact, the objective of this study was to investigate impact angle on the response of a headform, therefore it is assumed that these relationships would be similar, despite the specific headform used. This study was also delimited to the front and side impact location as well as the 0, 5, 10, and 15° impact angles. The locations were chosen because they represent different geometries associated with the head and the angles were chosen because these angles permitted the same location on the headform to be impacted at different angles given the geometry of the impactor striker.

8 REFERENCES

ANNA – SHOULD YOU MENTION ALL AUTHORS BY NAME IN REFERENCES 6, 14 & 20? I DON’T KNOW IF THE ASTM STYLE IS EXACTLY AS YOU HAVE GIVEN BELOW SO I MAY WELL BE WRONG ON THIS!


Figure 1.: Front and side impact conditions. Red represents the impact angle oriented at 0°; blue represents 5°; green represents 10° and orange represents 15°. The solid arrows represent positive impact angles and the double lined arrows represent the negative impact angles (only at side location).

Table 1. UCDBTM Properties

<table>
<thead>
<tr>
<th>Material</th>
<th>Young's Modulus (MPa)</th>
<th>Poisson's Ratio</th>
<th>Density (kg/m³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Scalp</td>
<td>16.7</td>
<td>0.42</td>
<td>1000</td>
</tr>
<tr>
<td>Cortical Bone</td>
<td>15000</td>
<td>0.22</td>
<td>2000</td>
</tr>
<tr>
<td>Trabecular Bone</td>
<td>1000</td>
<td>0.24</td>
<td>1300</td>
</tr>
<tr>
<td>Dura</td>
<td>31.5</td>
<td>0.45</td>
<td>1130</td>
</tr>
<tr>
<td>Pia</td>
<td>11.5</td>
<td>0.45</td>
<td>1130</td>
</tr>
<tr>
<td>Falx</td>
<td>31.5</td>
<td>0.45</td>
<td>1140</td>
</tr>
<tr>
<td>Tentorium</td>
<td>31.5</td>
<td>0.45</td>
<td>1140</td>
</tr>
<tr>
<td>CSF</td>
<td>-</td>
<td>0.5</td>
<td>1000</td>
</tr>
<tr>
<td>White Matter</td>
<td>Hyperelastic</td>
<td>0.499997</td>
<td>1060</td>
</tr>
<tr>
<td>Grey Matter</td>
<td>Hyperelastic</td>
<td>0.499998</td>
<td>1060</td>
</tr>
</tbody>
</table>
Table 2. Tissue characteristics for UCDBTM

<table>
<thead>
<tr>
<th>Location</th>
<th>Shear Modulus (kPa)</th>
<th>Decay Constant (s⁻¹)</th>
<th>Bulk Modulus (GPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Grey Matter</td>
<td>10</td>
<td>2.0</td>
<td>80</td>
</tr>
<tr>
<td>White Matter</td>
<td>12.5</td>
<td>2.5</td>
<td>80</td>
</tr>
<tr>
<td>Brain Stem</td>
<td>22.5</td>
<td>4.5</td>
<td>80</td>
</tr>
<tr>
<td>Cerebellum</td>
<td>10</td>
<td>2.0</td>
<td>80</td>
</tr>
</tbody>
</table>

Table 3. Peak resultant accelerations and maximum principal strain for each impact condition

<table>
<thead>
<tr>
<th>Location</th>
<th>Angle (°)</th>
<th>Peak Linear Acceleration (g)</th>
<th>Peak Angular Acceleration (rad/s²)</th>
<th>Maximum Principal Strain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Front</td>
<td>0</td>
<td>165.6 (0.9)</td>
<td>803 (187)</td>
<td>0.515 (0.03)</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>162.2 (3.3)</td>
<td>8282 (102)</td>
<td>0.496 (0.04)</td>
</tr>
<tr>
<td></td>
<td>10</td>
<td>165.6 (5.6)</td>
<td>8778 (128)</td>
<td>0.558 (0.01)</td>
</tr>
<tr>
<td></td>
<td>15</td>
<td>169.8 (1.6)</td>
<td>9939 (70)</td>
<td>0.610 (0.01)</td>
</tr>
<tr>
<td>Side</td>
<td>-15</td>
<td>174.5 (4.0)</td>
<td>11964 (346)</td>
<td>0.634 (0.02)</td>
</tr>
<tr>
<td></td>
<td>-10</td>
<td>171.7 (2.1)</td>
<td>11145 (106)</td>
<td>0.602 (0.01)</td>
</tr>
<tr>
<td></td>
<td>-5</td>
<td>164.5 (3.0)</td>
<td>10642 (603)</td>
<td>0.561 (0.02)</td>
</tr>
<tr>
<td></td>
<td>0</td>
<td>164.7 (7.1)</td>
<td>11067 (244)</td>
<td>0.557 (0.02)</td>
</tr>
<tr>
<td></td>
<td>5</td>
<td>162.3 (2.6)</td>
<td>10934 (153)</td>
<td>0.548 (0.01)</td>
</tr>
<tr>
<td></td>
<td>10</td>
<td>161.4 (0.7)</td>
<td>11034 (136)</td>
<td>0.527 (0.01)</td>
</tr>
<tr>
<td></td>
<td>15</td>
<td>159.8 (1.5)</td>
<td>11121 (275)</td>
<td>0.568 (0.01)</td>
</tr>
</tbody>
</table>
Figure 2. Hybrid III head and neckform with pointed striker impactor (front condition)

Figure 3. Hemispherical impactor striker