



<b>Title</b>	The Influence of Impactor Mass on the Dynamic Response of the Hybrid III Headform and Brain Tissue Deformation
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**Paper Title:**

The Influence of Impactor Mass on the Dynamic Response of the Hybrid III Headform and Brain Tissue Deformation

**Abstract:** When determining head injury risks through event reconstruction, it is important to understand how individual impact characteristics influence the dynamic response of the head and its internal structures. The effect of impactor mass has not yet been analysed in the literature. The purpose of this study is to determine the effects of inbound mass on the dynamic impact response and brain tissue deformation. A 50<sup>th</sup> percentile Hybrid III adult male head form was impacted using a simple pendulum system. Impacts to a centric and a non-centric impact location were performed with six varied inbound masses at a velocity of 4.0 m/s. The peak linear and peak angular accelerations were measured. A finite element model, (UCDBTM) was used to determine brain deformation, namely peak maximum principal strain and peak von Mises stress. Inbound mass produced significant differences for peak linear acceleration for centric ( $F_{5,24} = 217.55, p=.0005$ ) and non-centric ( $F_{5,24} = 161.98, p=.0005$ ), and for peak angular acceleration for centric ( $F_{5,24} = 52.51, p=.0005$ ) and non-centric ( $F_{5,24} = 4.18, p=.007$ ) impact locations. A change in inbound mass also had a significant effect on peak maximum principal strain for centric ( $F_{5,24} = 11.04, p=.0005$ ) and non-centric ( $F_{5,24} = 5.87, p=.001$ ), and for peak von Mises stress for centric ( $F_{5,24} = 24.01, p=.0005$ ) and non-centric ( $F_{5,24} = 4.62, p=.004$ ) impact locations. These results confirm the inbound mass of an impact should be of consideration when determining risks and prevention to head and brain injury.

**Keywords:**

Mass, acceleration, concussion, head injury, impact biomechanics

## **Introduction**

With the advent of helmet standards in sports, Traumatic Brain Injuries (TBI) have decreased dramatically[1]. However, even with the use of protective headgear the incidence of mild traumatic brain injury (mTBI) known as concussions have not, and continue to be a concern in both recreational and professional sports[2-5]. A recent survey reported 110 in 100,000 Canadians suffer from a concussion, or mild traumatic brain injury each year[6]. Of the reported concussions, over 54% were sports-related incidences. Although the terms concussion and mTBI are synonymous, the severity of concussion injuries is broad, where symptoms and recovery times differ, and may not always be described as a 'mild' injury[7]. Often post-concussion symptoms are used in the diagnosis and prognosis of concussion injuries, even despite the diversity of these symptoms among individuals[8, 9]. In 2011, Meaney and Smith reported that in the United States there are 225,000 new patients each year suffering long-term deficits from mTBI. They put this into perspective, indicating this is approximately equal to the combined number of patients per year diagnosed with breast cancer, multiple sclerosis, and traumatic spinal cord injuries [10]. In order to decrease the incidence and risks of head injuries, it is important to better understand the mechanism of these brain injuries. Impact parameters are described using a number of characteristics including impact site, mass, velocity, angle of impact and compliance of impactor, creating unique dynamic head responses and consequently different head and brain injuries[11-15].

Helmets are designed to attenuate the impact energy or magnitude of the impact force through deformable materials increasing the duration of an impact and reducing the linear acceleration experienced by the head, resulting in a decrease of injury severity[16, 17]. For any given impact energy, the resultant impulse experienced by the head remains the same, however the force-time

curve shape changes depending on the energy-absorbing material[18]. The energy-absorbing material increases the duration of the force, resulting in a decreased magnitude. Impacts in which a high force is experienced in a short time duration can cause more serious injury, or earlier material failure, than those impacts where the force transmission is prolonged and dampened[18]. However this force-time curve shape is not only dependent on the presence of a helmet, but also depends on the interaction of the striking object with the skull, brain, and surrounding soft tissues[18], all of which therefore may influence the peak accelerations experienced and resulting head injury.

Present day helmet certification standards primarily use centric impact testing protocols and resulting linear accelerations as measures of helmet performance[19]. This has led to successes in reducing incidences of traumatic brain injury, however they may not fully encompass and protect against the mechanisms for concussion[1]. Angular accelerations and brain tissue deformation measures have also been shown to correlate with the risk of mTBI, whereas the peak linear acceleration does not always fully describe this injury risk [11, 20-27].

Typically the safety characteristics of helmets are determined using an apparatus that drops the helmet and headform on to a hard surface replicating a fall to the ground [28-30]. However, the majority of concussive injuries in sport are the result of other types of impacts, such as body to body contact [31], where the striking mass of the impactor may vary considerably. Player to player impacts involving helmets, shoulders, elbows and hands are most often non-centric impacts which may lead to higher risks of concussion.

A better understanding of the mechanisms and impact parameters that describe the risk associated with mTBI injuries provides valuable insight for injury prevention and management of the risks associated with concussions. It is important to determine the response of the head and

brain tissue under variable inbound mass conditions to help establish appropriate strategies for decreasing the risk of mTBI in sport. The purpose of this study is to describe the effects of inbound or striking mass on the dynamic impact response of the Hybrid III head form, as measured by peak resultant linear and angular accelerations. The influence of variable inbound mass on brain tissue deformation as measured by peak maximum principal strain and peak von Mises stress has also been examined. Dynamic head response and brain tissue deformations were observed for centric and non-centric impact conditions.

## **Materials and Methods**

### *Test Equipment*

*Pendulum System:* The impacts for this study were performed using a pendulum system. This impacting system involved a hollow metal frame weighing 3.36 kg (Fig. 1) and was suspended using 3/32" aviation cable (length 9.25 ft.). The impactor was free to swing using four cables attached to a rigid ceiling-mounted metal beam directly above the head form. Circular metal plates weighing 1 kg and 2 kg were fastened into the center of the cylinder to achieve the appropriate mass (Fig. 1). A hemispherical nylon striker (diameter 12.6 cm, weight 0.92 kg), containing a compliant 25.4 mm modular elastomer programmer (MEP) 60 Shore Type A middle layer pad, was attached to the impacting end of the pendulum cylinder (Fig 1). MEP is a stiff material that offers very little compliance and therefore may have the least amount of influence on the results in terms of energy absorption. For this reason, it was chosen as the impact surface for this study so that effects of varying inbound mass were measured via the response of the head, rather than the response of the impacting surface. In addition, this is a common material

used in helmet standard testing protocols[28, 30]. The pendulum system was positioned so that the center of the metal steel frame was aligned with the impact site on the head form.

*Hybrid III Head and Neck Form:* For this experiment, an unhelmeted 50<sup>th</sup> percentile adult male Hybrid III head and neck form was impacted. The Hybrid III head form with mass of  $4.54 \pm 0.01$  kg, was attached to a Hybrid III neck form with mass of  $1.54 \pm 0.01$  kg (Fig. 2). This head and neck form is currently the most widely used and advanced human surrogate with biofidelity built into its impact response. The Hybrid III geometry and its characteristics such as head mass and the location of its center of gravity was established based on cadaveric experimentation. This means that it represents human impact characteristics in terms of the essential biomechanical responses[32, 33].

The Hybrid III head and neck form was attached to a low-friction sliding table. The sliding table ( $12.782 \pm 0.001$  kg), allowed the head form to be adjusted within 6 degrees of freedom, and is mounted on rails to provide a low friction surface. The low friction system allows the head form to slide upon impact. The dynamic response of the head form, namely the three-dimensional motion during impact, was measured using nine calibrated single-axis Endevco7264C-2KTZ-2-300 accelerometers (Endevco, San Juan Capistrano, CA), orthogonally positioned in a 3-2-2-2 array and fixed near the center of gravity of the head form[34]. Angular acceleration was calculated using the first principles of rigid body dynamics and linear acceleration from the linear accelerometer orthogonal array with the following equations:

$$\vec{\alpha}_x = \frac{a_{zS}^{\vec{}} - a_{zC}^{\vec{}}}{2S} - \frac{a_{yT}^{\vec{}} - a_{yC}^{\vec{}}}{2T} \quad (1)$$

$$\vec{\alpha}_y = \frac{a_{xT}^{\vec{}} - a_{xC}^{\vec{}}}{2T} - \frac{a_{zF}^{\vec{}} - a_{zC}^{\vec{}}}{2F} \quad (2)$$

$$\vec{\alpha}_z = \frac{a_{yF}^{\vec{}} - a_{yC}^{\vec{}}}{2F} - \frac{a_{xS}^{\vec{}} - a_{xC}^{\vec{}}}{2S} \quad (3)$$

Where  $\alpha_i$  is the angular acceleration for the component  $i$  ( $x, y, z$ ) and  $a_{ij}$  is for the linear acceleration for component  $i$  ( $x, y, z$ ) along the orthogonal arm  $j$  (S, T, F)[34]. The left-hand rule is used to define the coordinate system for the head form. The positive axis is directed anteriorly towards the right ear, and caudally for the  $x, y$  and  $z$ , respectively[35].

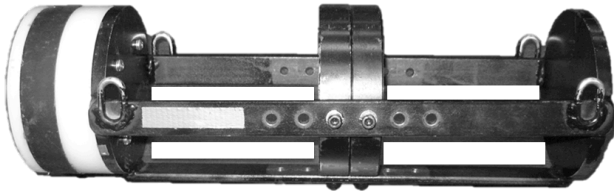


Fig. 1: Pendulum frame, circular metal weights, and MEP impactor cap.

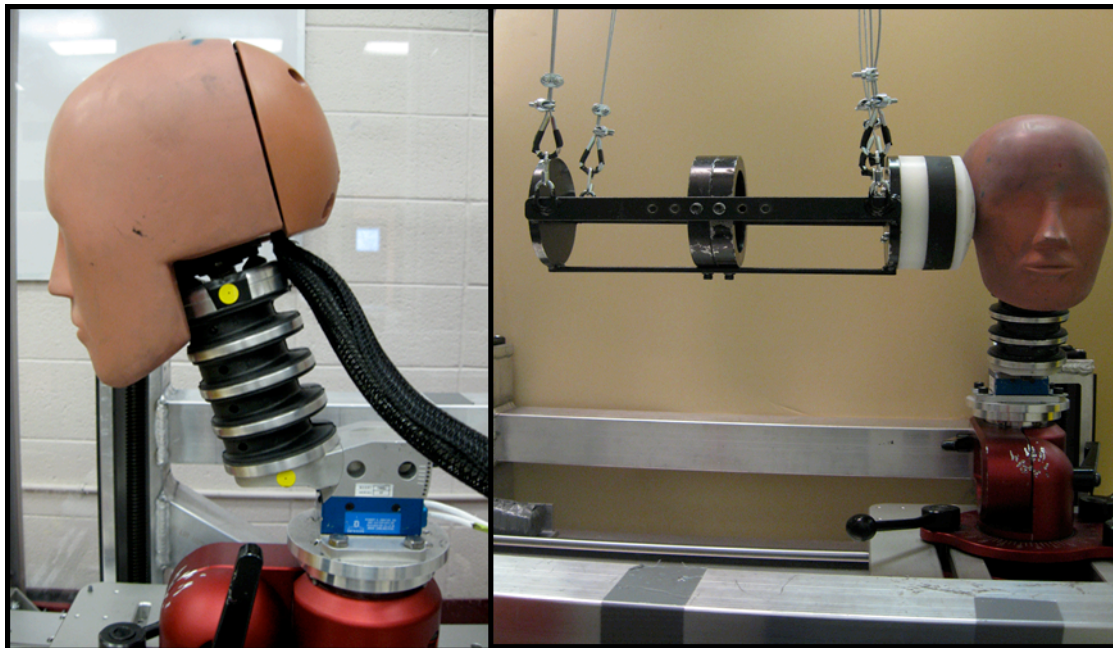


Fig. 2: Hybrid III head and neck form; Pendulum system

*University College Dublin Brain Trauma Model:* In order to predict the brain tissue deformation metrics, a partially validated finite element brain model. The model used was the University College Dublin Brain Trauma Model (UCDBTM) which was developed at the University College Dublin by Horgan & Gilchrist (2003; 2004), and produces brain tissue

deformation corresponding to head accelerations upon impact. This model has been partially validated by comparing model responses against cadaveric experiments measuring intracranial pressure[16] and relative brain motion[36]. Furthermore this model was validated using reconstructions of actual traumatic brain injury incidents[37]. The head/brain geometry of this model was determined through Computed Tomography (CT), Magnetic Resonance Imaging (MRI), and sliced contour photographs of a male cadaver[38]. The components of this model include the scalp, a 3-layered skull, dura, CSF, pia, falx, tentorium, cerebral hemispheres, cerebellum, and brainstem. This model mathematically equates the response characteristics of neural tissue of various anatomical locations within the brain, accounting for their respective material properties. Table 1 & 2 summarizes the mechanical properties of the anatomical model components, which were established from previously conducted research using cadavers[17, 39-41].

Table 1: Material properties for UCDBTM

<b>Material</b>	<b>Young's modulus (MPa)</b>	<b>Poisson's ratio</b>	<b>Density (kg/m<sup>3</sup>)</b>	<b>Reference</b>
Scalp	16.7	0.42	1000	Kl & von 2002
Cortical Bone	15000	0.22	2000	Kl & von 2002
Trabecular Bone	1000	0.24	1300	Kl & von 2002
Dura	31.5	0.45	1130	Kl & von 2002
Pia	11.5	0.45	1130	Zhou 1995
Falx and Tentorium	31.5	0.45	1130	Kl & von 2002
Grey Matter	30	0.49	1060	Zhou 1995
White Matter	37.5	0.49	1060	Zhou 1995
CSF	Water	0.50	1000	Ruan 1994

Table 2: Material characteristics of brain tissue components for UCDBTM.

<b>Material</b>	<b><math>G_0</math></b>	<b><math>G_\infty</math></b>	<b>Decay Constant (GPa)</b>	<b>Bulk Modulus (<math>s^{-1}</math>)</b>
Cerebellum	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19

A hyperelastic material model is used for the brain in shear, in conjunction with a viscoelastic material property (Horgan 2005). The hyperelastic law is given by:

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)} \quad (5)$$

where  $C_{10}$ , and  $C_{01}$  are temperature dependent material parameters, and  $t$  is seconds. The behaviour of the brain tissue was modeled as viscoelastic in shear combined with large deformation theory, and the compressive behaviour of the brain is considered elastic[38]. The shear modulus characteristics of the brain tissue, modeled as being viscoelastic, and are given by:

$$G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t} \quad (4)$$

where  $G_\infty$ , is the long term shear modulus,  $G_0$ , is the short term shear modulus, and  $\beta$  is the decay factor[38]. Modeling of the CSF was done using solid elements with low shear and high bulk moduli values. The skull brain interface is represented by the models interaction between the CSF and the brain with a sliding boundary condition, thus allowing for a sliding like behaviour similar to that of water. Overall the UCDBTM consists of approximately 26, 000 hexahedral elements[38, 39]. Horgan and Gilchrist (2003; 2004) provide additional detailed information regarding specificities of model characteristics, boundary conditions, mesh densities etc.

## *Procedures*

The main objective of this study was to determine the effects of impactor mass on the dynamic response (peak resultant linear and peak resultant angular acceleration) of the unhelmeted Hybrid III head form, and brain tissue response, as represented through finite element modeling (FEM). Altering the Hybrid III head form position about the  $x$ ,  $y$ , and  $z$ -axes made it possible to discriminate between impact conditions. The Hybrid III head form was impacted at one centric (Side Center Gravity, SGA) and one non-centric (Front Boss 45° Positive Azimuth, FBPA) impact site based on the University of Ottawa Testing Protocol (uOTP)[35] (Table 3). A centric and a non-centric impact condition were chosen, as the direction of the applied force was shown to have an effect on the dynamic impact response[43] and risks to injury[35]. When the direction of the impact vector is through the center of gravity of the head form it is defined as a centric impact, whereas an impact vector that is not collinear with the center of gravity is considered to be a non-centric impact. Both impact sites were to the side of the head, which is a common impact site in sport and presents high risk[13, 15, 44]. Impacts at each site were performed with six inbound masses that differed by 2 kg increments: 4.3 kg; 6.3 kg; 8.3 kg; 10.3 kg; 12.3 kg; 14.3 kg. These masses were chosen as they encompass a range of inbound masses that are seen with sport impacts[45-48]. The head form was impacted 5 times under each impact condition for a total of 60 impacts. A low impact velocity of 4 m/s was chosen to avoid material saturation from impacts to the bare headform. Higher velocity impacts seen within sport typically involve head protection. The velocity of the pendulum impactor prior to impacting the head form was measured using a High Speed Imaging PCI-512 Fastcam running at 2 kHz and Photron Motion Tools computer software (Photron, San Diego CA) for consistency. Signals from the nine accelerometers were sampled at 20 kHz and filtered through a 1000 Hz low pass Butterworth

filter using SAE J211 Class 1000 protocol[49]. Data collection was triggered once a 3 g threshold was reached by any one of the accelerometers; data was collected for a period of 15 ms. Signals from the accelerometers were passed through TDAS Pro Lab module (DTS, Calabasas, CA), before being processed with TDAS software. Data analysis and filtering was done using BioProc3, software developed by Robertson, University of Ottawa (2008). A series of one-way ANOVAs were performed for each dependent variable and impact condition to compare the different impact conditions. When significance was found at  $\alpha = 0.05$ , a post hoc Tukey test was performed to establish significant differences between individual inbound masses.

The three-dimensional dynamic response of the Hybrid III head form, the  $x$ ,  $y$ , and  $z$  component loading curves for both linear and angular acceleration, were inputted into the UCDBTM [38] for finite element analysis. These accelerations were applied to the centre of gravity, CG, of the model similarly to Zhang et al. (2004) and Kleiven (2007) in their reconstructions of football concussions. This produced the deformation response characteristics, von Mises stress (VMS) and maximum principal strain (MPS), within the brain tissue. Research employing finite element analysis has measured a variety of brain deformation variables which have highlighted the best predictors of brain injury. The brain deformation metrics used for brain injury prediction and risk include maximum principal strain and von Mises stress (Willinger et al., 2000; Anderson et al., 2003; Willinger & Baumgartner, 2003; Kleiven, 2007).

Table 3: Two impact conditions defined by impact location and angle.

<b>Condition</b>	<b>Impact Location</b>	<b>Impact Angle</b>
Non-centric (FBPA)	Midpoint between the anterior mid-sagittal and right coronal planes in absolute transverse plane	45° rotation in the transverse plane
Centric (SCG)	Right intersection of the coronal and absolute transverse planes	No vertical or horizontal rotation was applied to the vector

## Results

To determine the influence of inbound mass on the dependent variables, eight ANOVAs were completed. These are the results from impacts performed at a velocity of 4.0 m/s to a centric (SCG) and non-centric (FBPA) location using six different inbound masses. The means and standard deviations were determined from five trial impacts performed for each condition.

### *Dynamic Impact Response*

Table 4 displays the mean peak resultant linear and angular accelerations of a Hybrid III head form for each inbound mass tested. A change in inbound mass was found to have a significant effect on peak linear acceleration for centric ( $F_{5,24} = 217.55, p < 0.0005$ ) and non-centric ( $F_{5,24} = 161.98, p < 0.0005$ ) impact locations. Significance was also found for peak angular acceleration for centric ( $F_{5,24} = 52.51, p < 0.0005$ ) and non-centric ( $F_{5,24} = 4.18, p = 0.007$ ) impacts.

Table 4: Mean peak dynamic response ( $\pm 1$  standard deviation) of the hybrid III head form resulting from impacts with six inbound masses performed at 4.0 m/s to a centric and non-centric impact location.

<b>Inbound Mass (kg)</b>	<b>Centric (Side CG)</b>		<b>Non-Centric (Front Boss PA)</b>	
	<b>Linear Acc. (g)</b>	<b>Angular Acc. (krad/s<sup>2</sup>)</b>	<b>Linear Acc. (g)</b>	<b>Angular Acc. (krad/s<sup>2</sup>)</b>
<b>4.3</b>	177.8 $\pm$ 1.9	15.2 $\pm$ 0.2	158.7 $\pm$ 6.8	18.4 $\pm$ 0.8
<b>6.3</b>	199.4 $\pm$ 2.3	16.1 $\pm$ 0.2	180.7 $\pm$ 4.1	18.8 $\pm$ 0.7
<b>8.3</b>	210.5 $\pm$ 0.8	17.0 $\pm$ 0.1	191.6 $\pm$ 1.8	17.8 $\pm$ 0.5
<b>10.3</b>	228.4 $\pm$ 5.9	18.0 $\pm$ 0.3	197.6 $\pm$ 3.4	18.2 $\pm$ 0.7
<b>12.3</b>	228.4 $\pm$ 3.1	17.9 $\pm$ 0.2	200.4 $\pm$ 3.9	18.8 $\pm$ 1.4
<b>14.3</b>	229.6 $\pm$ 2.6	17.5 $\pm$ 0.6	226.5 $\pm$ 0.5	19.9 $\pm$ 0.2

Mean peak resultant linear and angular accelerations for the centric condition ranged from 177 - 230 g and 15,300 - 17,500 rad/s<sup>2</sup>, respectively. The resulting centric dynamic response increased as the inbound mass of the impact increased, however they reached their maximum values at the 10.3 kg inbound mass condition. Impacts using an inbound mass of 10.3 kg, 12.3 kg and 14.3 kg were not significantly different from one another. There was no observed significant difference in mean peak resultant angular acceleration between an inbound mass of 8.3 kg and 14.3 kg.

Mean peak resultant linear and angular acceleration for the non-centric impact condition ranged from 158 - 226 g, 18,450 - 19,900 rad/s<sup>2</sup>, respectively. An increase in inbound mass resulted in an increase in the linear acceleration response for the non-centric impact condition. Significant differences in the mean peak resultant linear accelerations resulting from a non-centric impact were found between all inbound masses tested excluding those between an 8.3 kg and 10.3 kg mass, and between a 10.3 kg and 12.3 kg inbound mass. Mean peak resultant angular

accelerations resulted in significant differences between an inbound mass of 8.3 kg and 14.3 kg, and between 10.3 kg and 14.3 kg.

### *Brain Tissue Deformation*

The resulting mean MPS and VMS values are reported in table 5 for each inbound mass tested. A change in inbound mass had a significant effect on peak MPS for both centric ( $F_{5, 24} = 11.04$ ,  $p < 0.0005$ ) and non-centric ( $F_{5, 24} = 5.87$ ,  $p = 0.001$ ) impact locations. Significance was also found for peak VMS for centric ( $F_{5, 24} = 24.01$ ,  $p < 0.0005$ ) and non-centric ( $F_{5, 24} = 4.62$ ,  $p = 0.004$ ) impacts.

Table 5: Mean peak brain tissue deformation metrics ( $\pm 1$  standard deviation) as determined from finite element analysis resulting from impacts with six inbound masses performed at 4.0 m/s to a centric and non-centric impact location.

<b>Inbound Mass (kg)</b>	<b>Centric (Side CG)</b>		<b>Non-Centric (Front Boss PA)</b>	
	<b>maximum principal strain</b>	<b>von Mises stress (kPa)</b>	<b>maximum principal strain</b>	<b>von Mises stress (kPa)</b>
<b>4.3</b>	0.285 $\pm$ 0.009	10.1 $\pm$ 0.4	0.499 $\pm$ 0.024	19.9 $\pm$ 1.0
<b>6.3</b>	0.304 $\pm$ 0.004	11.1 $\pm$ 0.2	0.536 $\pm$ 0.017	21.3 $\pm$ 0.8
<b>8.3</b>	0.315 $\pm$ 0.009	11.3 $\pm$ 0.2	0.493 $\pm$ 0.017	19.3 $\pm$ 0.8
<b>10.3</b>	0.330 $\pm$ 0.016	12.2 $\pm$ 0.4	0.527 $\pm$ 0.012	20.6 $\pm$ 0.6
<b>12.3</b>	0.326 $\pm$ 0.016	12.1 $\pm$ 0.3	0.562 $\pm$ 0.042	22.1 $\pm$ 2.0
<b>14.3</b>	0.325 $\pm$ 0.009	11.7 $\pm$ 0.5	0.516 $\pm$ 0.016	19.9 $\pm$ 0.7

Centric impacts resulted in mean peak MPS and VMS values ranging from 0.29 – 0.33 and 10.1 – 11.7 kPa, respectively. The finite element modeling contours for MPS resulting from centric impacts of 4.3-14.3 kg masses are shown in figure 3. The centric impact condition experienced a plateau in the brain deformation response at an inbound mass of 10.3 kg. This condition showed

that mean peak MPS resulting from an inbound mass of 4.3 kg was significantly lower than MPS from all other inbound masses excluding 6.3 kg. In addition, MPS resulting from 6.3 kg inbound mass and 10.3 kg inbound mass were significantly different. Mean peak VMS resulting from an inbound mass of 10.3 kg, 12.3 kg and 14.3 kg were not significantly different from each another. The 6.3 kg, 8.3 kg and 14.3 kg inbound mass impacts produced mean peak VMS that were not statistically different. The brain tissue experienced much larger deformations as a result of the non-centric impacts with mean peak MPS and VMS values ranging from 0.50 – 0.56 and 19.9 – 22.1 kPa, respectively. The 12.3 kg inbound mass impacts created significantly higher brain deformations than the 4.3 kg, 8.3 kg and 14.3 kg mass impacts for both MPS and VMS.

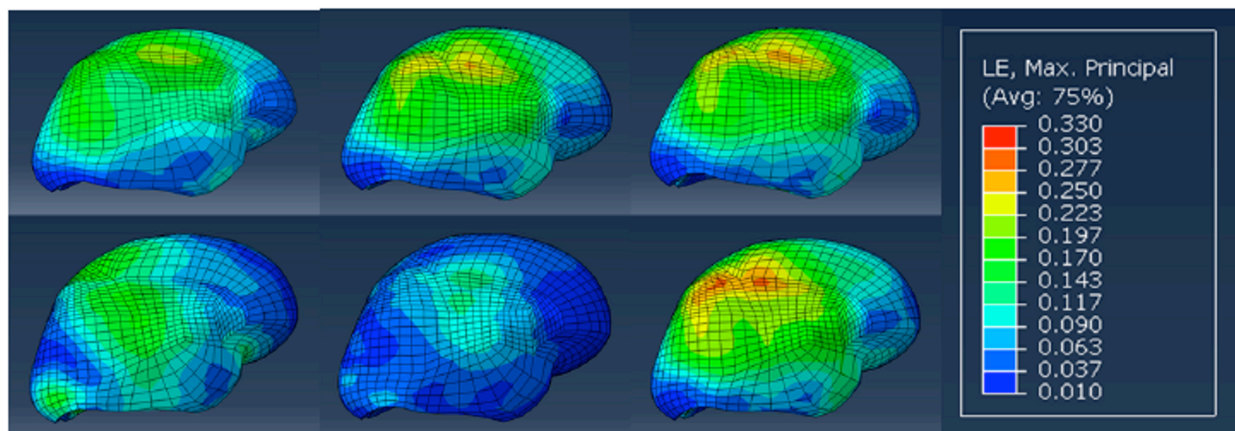


Fig. 3: FEM brain deformation contours of peak maximum principal strain resulting from a centric impact with inbound mass of 4.3kg (top left), 6.3kg (top middle), 8.3kg (top right), 10.3kg (bottom left), 12.3kg (bottom middle), and 14.3kg (bottom right).

## **Discussion**

The findings from the present study showed that differences in inbound mass had a significant effect on the dynamic response of the head, and subsequent brain deformation values. Interestingly, the effect of increasing the striking mass resulted in differences in both the linear and angular acceleration responses. However, the non-centric impact results did not follow the same trend as for the centric impacts. The mean peak resultant linear and angular accelerations did not follow a linear relationship with increases in inbound striking mass. Peak linear accelerations for both centric and non-centric impact conditions did increase as the inbound mass increased, however peak angular acceleration responses were more variable. Peak angular accelerations also demonstrated greater standard deviations between impact trials. The increased variation in the peak angular acceleration values may be explained by more complex impact characteristics specific to using a pendulum system. This variability was more pronounced for the non-centric impact condition. The pendulum impact system allows greater degrees of freedom of the impact event, resulting in an increase in measurement variability. Depending on the impact parameters, the resulting dynamic response of the head is affected by the velocity, compliance, direction, mass and location of the impact. Linear accelerations for the centric impact location plateaued as the inbound mass increased from 10.3 to 14.3 kg. This may be due to saturation in the compliance or energy absorption of the materials involved in the impact (bottoming-out of material), namely the Hybrid III head and neck form and MEP impactor cap. This plateau was not observed for the non-centric impacts; however these impacts did not reach the same level of acceleration, approximately 228 g.

Peak angular accelerations and the peak deformation variables for both the centric and non-centric impacts revealed a similar trend with the increase of inbound mass. This is in accordance

with other studies that have reported angular response of the headform is more closely associated with brain deformation values. High positive correlations between these two response variables were reported in the literature for helmeted impacts[26, 27, 51], as well as research that associates both angular acceleration and brain deformation with concussion type injuries[11, 23]. The results from this experiment suggest that the inbound mass of an impact influences these dependent variables differently depending on the location of the impact. This finding was consistent with previous research that shows variation occurs in the head kinematics and brain tissue response by changing the parameters of the impact[11-15, 21].

To compare centric and non-centric impacts in terms of the proportionate increase within the response variables due to increasing inbound mass, percentage difference calculations were completed. The percentage differences between centric and non-centric impacts for peak linear and angular accelerations ranged from 1.3 to 14.5 % and 1.1 to 18.7 %, respectively.

Interestingly though, the percentage differences in peak tissue deformation variables were considerably higher, ranging from 44.1 to 55.3 % and 51.6 to 65.4 % for peak maximum principal strain and peak von Mises stress, respectively. This greater change may be explained by considering the shape of the individual curves created by the impact, within each of the three axes[52]. The brain model uses an integration of these acceleration time curves, rather than peak resultant values, when determining brain tissue responses. Examining the impact from this standpoint, results showed similar single axis responses for linear acceleration between centric and non-centric conditions (Fig. 4, 5), however much higher angular accelerations about the z-axis were observed under the non-centric impact condition (Fig. 6), when compared to centric (Fig. 7). Deformation of brain tissue may be more sensitive to rotations about the z-axis and

could explain why the non-centric location produced higher stress and strain (Table 5). This occurrence was not observed in view of the peak resultant dynamic response alone (Table 4).

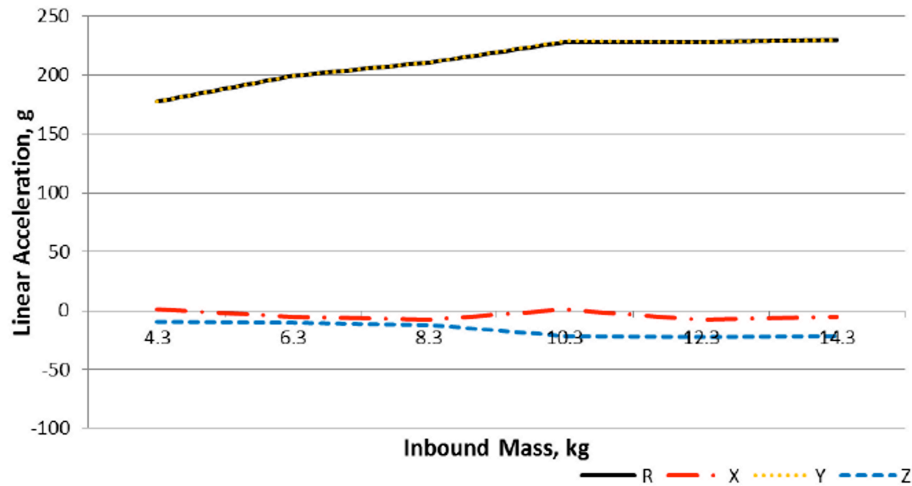


Fig. 4: Peak linear accelerations within the x, y, and z axes across six inbound masses from the centric (SCG) impact location. Peak resultant follows the y axis.

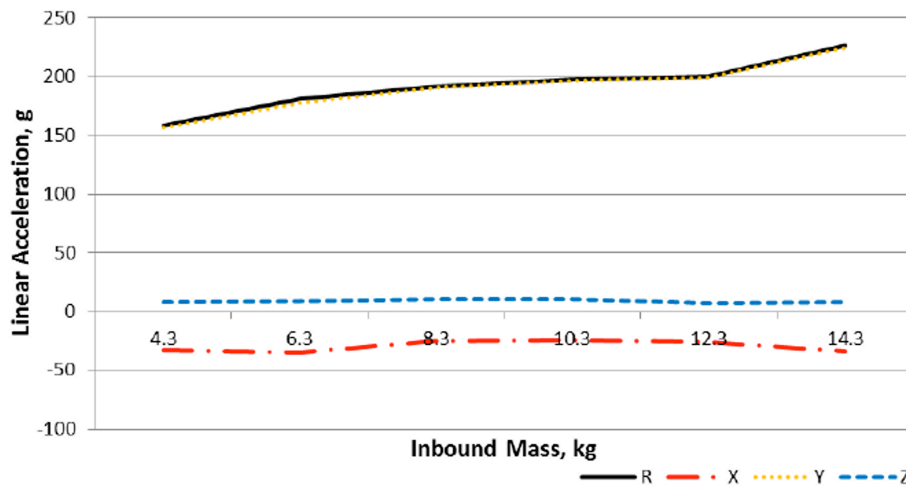


Fig. 5: Peak linear accelerations within the x, y, and z axes across six inbound masses from the non-centric (FBPA) impact location. Peak resultant follows the y axis.

Non-centric impacts have been shown to create a more rotation dominant response, or angular acceleration of the head[35]. This method was demonstrated by Viano and associates (2005), where boxing punches created proportionately higher head rotational acceleration when compared to NFL impacts, even when a similar inertial force of the head was created from the

impact[46]. This was attributed to the effective radius from the head cg. The results from this study support this finding where the peak maximum principal strain and von Mises stress values were much higher for the non-centric impact location. These results describe the effect of inbound mass on brain deformation depending on the characteristics of the impact, for example head impact location and angle. Moreover, peak accelerations alone may not adequately represent the severity of brain deformation when impacts to the head are non-centric, individual axes loading-curve analysis may enhance this understanding.

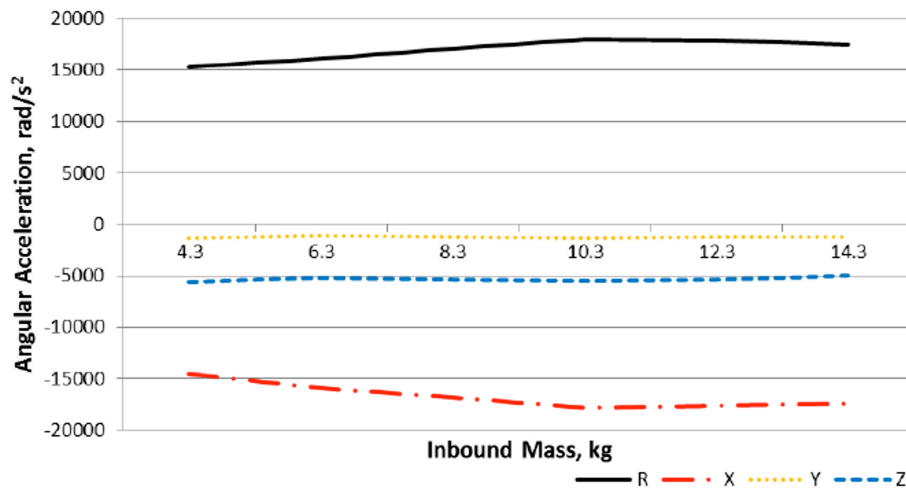


Fig. 6: Peak angular accelerations within the x, y, and z axes across six inbound masses from the centric (SCG) impact location.

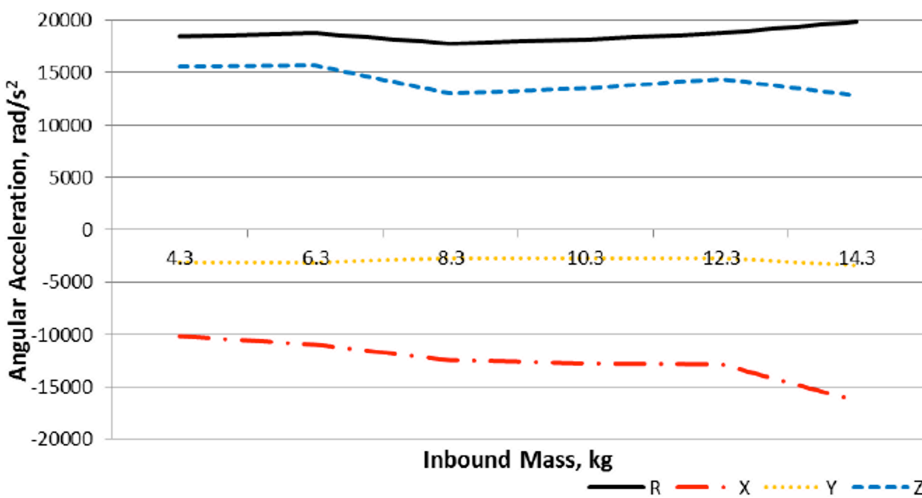


Fig. 7: Peak angular accelerations within the x, y and z axes across six inbound masses from the non-centric (FBPA) impact location.

The results from this research show that even with low inbound impact mass, high head accelerations can be achieved. As observed in the dynamic response data, the unpredictability of angular accelerations, particularly resulting from non-centric impacts results in high risk for concussion. This mechanism of injury can be created even with low mass impacts, and may be important in understanding the high number of concussions occurring in sports where non-centric impacts are common.

## **Conclusion**

This research describes the relationship between inbound mass and the dynamic impact response of the head and brain tissue deformation characteristics for centric and non-centric impact conditions. An increase in inbound mass resulted in a significant effect on both peak linear and angular acceleration for centric and non-centric impact locations. As well, increases in inbound mass had a significant effect on both peak MPS and peak VMS for both centric and non-centric impact locations. This study has described the effect of inbound mass on the dynamic response of the headform and brain tissue using physical reconstructions representing real life impact parameters. Inbound mass is an important parameter to consider when undertaking injury reconstructions, establishing head protection strategies and measuring risk of injury.

## **Limitations**

There are inherent limitations when using physical and finite element models to evaluate head dynamic response and neural tissue damage during impact testing. Although widely accepted, the

Hybrid III head- and neckform is an approximation of the head and neck geometries of an adult male. These physical models are composed of steel and rubber and are therefore only used to estimate the dynamic properties and impact response of an actual human head, with the understanding of the compliant nature of live tissues. In addition, these dummies were designed and validated primarily for antero-posterior directional inertial loading, and as such would may produce a stiffer response within other planes. It must be noted that this study was not intended to reproduce actual human head injury, but rather create a controlled environment in which response characteristics could be compared. The finite element model used in this research to evaluate the response of brain tissue under impact also has limitations to the methodology. Tissue responses are determined from the material characteristics and parameters defined to this specific model. It is recognized that the peak brain deformation results using the UCDBTM in this present study would likely differ if compared to another model. However, these peak values were only used for impact condition comparisons specific to this study, and therefore under the same model parameters.

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