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Authors(s)	Post, Andrew; Oeur, Anna; Hoshizaki, Thomas Blaine; et al.
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The influence of centric and non-centric impacts to American football helmets on the correlation between commonly used metrics in brain injury research

Andrew Post[1], Anna Oeur[1], Blaine Hoshizaki[1] and Michael D. Gilchrist[2,1]

Abstract Concussion has become recognized as an injury which can be a source of long term neurological damage. This has led to research into which metrics may be more appropriate to define risk of injury. Some researchers support the use of linear acceleration as a metric for concussion, while others suggest the use of linear and rotational acceleration as well as brain deformation metrics. The purpose of this study was to examine the relationships between these metrics using a centric and non-centric impact protocol. A linear impactor was used to impact a Hybrid III headform fitted with different models of American football helmet using a centric and non-centric protocol. The dynamic response was then used as input to the FE model for analysis of brain deformations. The results showed that linear acceleration was correlated to rotational acceleration and brain deformation for centric conditions, but under non-centric conditions it was not. These results indicate that the type of methodology used will influence the relationship between the variables used to assign risk of concussion. These results also support the use of a centric/non-centric protocol and measurement of rotational acceleration and brain deformation when it comes to the development of helmet technologies.

Keywords American football, Brain injury, Concussion, Impact biomechanics

I. INTRODUCTION

Brain injury and concussion are common in contact sports such as American football and ice hockey [1]. Concussion in particular has become a cause of concern among sporting institutions since research has been published indicating that multiple concussions over the period of a player's career could lead to permanent degenerative brain damage [2]. In an effort to reduce the incidence of concussion in sport, researchers have been examining the relationship between the linear and rotational acceleration and brain deformation metrics to assess what variables might be most useful in the design of better sporting protective equiment [3]-[4]-[5].

Head impacts are currently characterised using linear and rotational acceleration [6]-[7]. Linear acceleration has been shown through research to be associated with traumatic brain injuries such as skull fracture and subdural hematoma [8]. Rotational acceleration can lead to the type of diffuse shearing of brain tissue which is thought to contribute to the incidence of a concussion [9]-[10]. These two metrics, however, have had a low correlation to injury, which has led to the use of brain deformation metrics (e.g. strain) to bridge the gap between response and injury [6]-[7]. How the linear and rotational accelerations influence the resulting brain deformation metrics is an ongoing source of debate. Some researchers maintain that linear and angular acceleration are correlated [11], and as a result when evaluating the protective capacity of helmets measurements of linear acceleration are sufficient to capture the risk of injury. Other researchers, however, have demonstrated that under many types of sporting impact conditions linear and rotational acceleration are not correlated [12]. It has been suggested that when the line of impact passes throught the centre of gravity (centric), there will be high correlations between linear and rotational acceleration and the resulting brain deformation metrics. Impact conditions that do not pass through the centre of gravity (non-centric) would create lower correlations [5]. Also, the influence of these accelerations on brain deformation needs to be established to aid in the development of technologies which can reduce stress and strain within the brain.

[1]Andrew Post is a PhD student in Biomechanics at the University of Ottawa, Canada (613-562-5800 ext 7210, apost@uottawa.ca). [1]Anna Oeur is a MSc student in Biomechanics at the University of Ottawa. [1]T. Blaine Hoshizaki is Prof. of Biomechanics in the Human Kinetics Department at the University of Ottawa. [2]Michael D. Gilchrist is Prof. of Engineering at the School of Mechanical & Materials Engineering at the University College Dublin, Ireland.

The purpose of this research is to examine the correlation of linear and rotational acceleration to maximum principal strain (MPS) under centric and non-centric impact conditions using football helmets. The results will be analyzed for peak brain deformation metrics as well as an average brain deformation metric representing nine separate functional regions of the cerebrum.

II. **METHODS**

Experimental Testing

Three different models (vinyl nitrile liner, Skydex liner, and three dimensional structure liner) of American football helmet were impacted at 7.5 m/s using a linear impactor system (Fig. 1.) [11]. The linear impactor consists of an impacting frame and a sliding table frame. The impacting frame consists of the impacting arm and compressed air canister. The sliding table frame consists of a sliding table which has a 50th percentile Hybrid III headform attached to it in such a way that allows selection of impact sites in five degrees of freedom. The impactor arm had a mass of 13.1 ± 0.1kg with a hemispherical nylon tip with a VN 602 layer underneath. The impactor striker was designed to simulate the compliance of a helmet to helmet impact in American football. The 50th percentile Hybrid III headform was fitted with Endevco 7264C-2KTZ-2-300 uniaxial accelerometers in a 3-2-2-2 array for the measurement of three dimensional kinematics [13]. The signals were collected at 20 kHz and filtered with a 1000 Hz 2nd order lowpass butterworth filter using Diversified Technical Systems TDAS Pro lab module software.

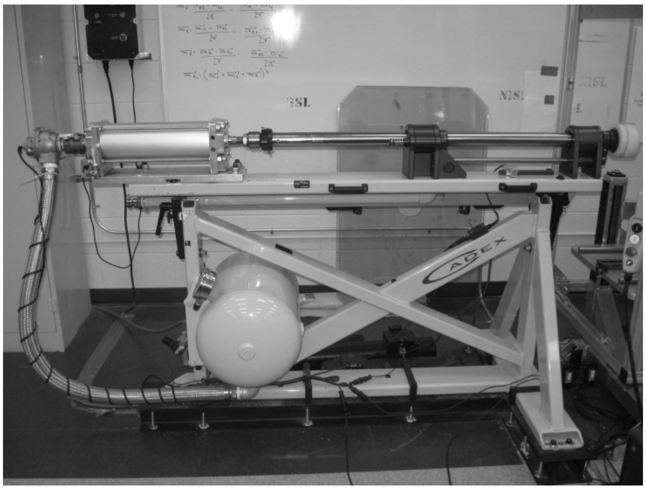

Fig. 1. Linear impactor frame with impacting arm

The helmets were fitted on the 50th percentile Hybrid III headform and impacted in nine sites (Fig. 2 and 3), 5 of which were centric (through the centre of gravity of the headform) in nature, and the remaining 4 non-centric (not through the centre of gravity of the headform) [12]. Each helmet was impacted three times per site, with a new helmet used for each impact condition. Overall, 81 impacts were conducted, 71 of which were useable for the statistical analyses. The resulting dynamic responses were used as input for the University College Dublin Brain Trauma Model for analysis of the resulting brain deformations. Analysis of the relationship between the dynamic response and brain deformation for all helmets as a group, measured in maximum principal strain (MPS) at the peak point in the cerebrum and an average MPS from nine separate regions of the cerebrum, were conducted using a Pearson correlation.

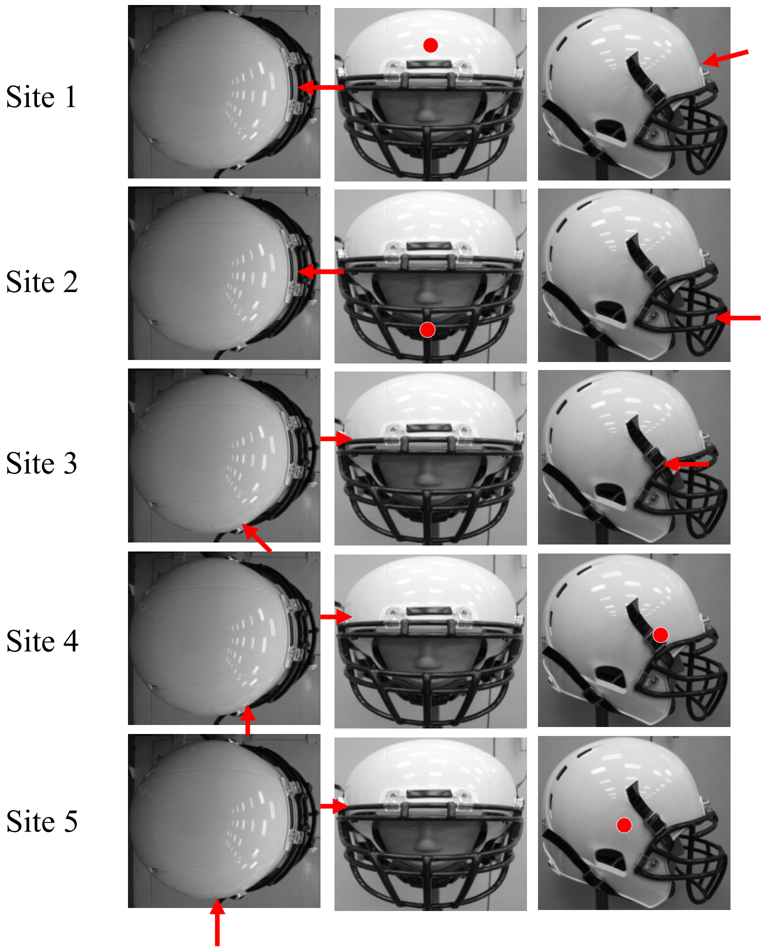

Fig. 2. Impact sites 1-5 on the American football helmets, impact vector indicated by arrow

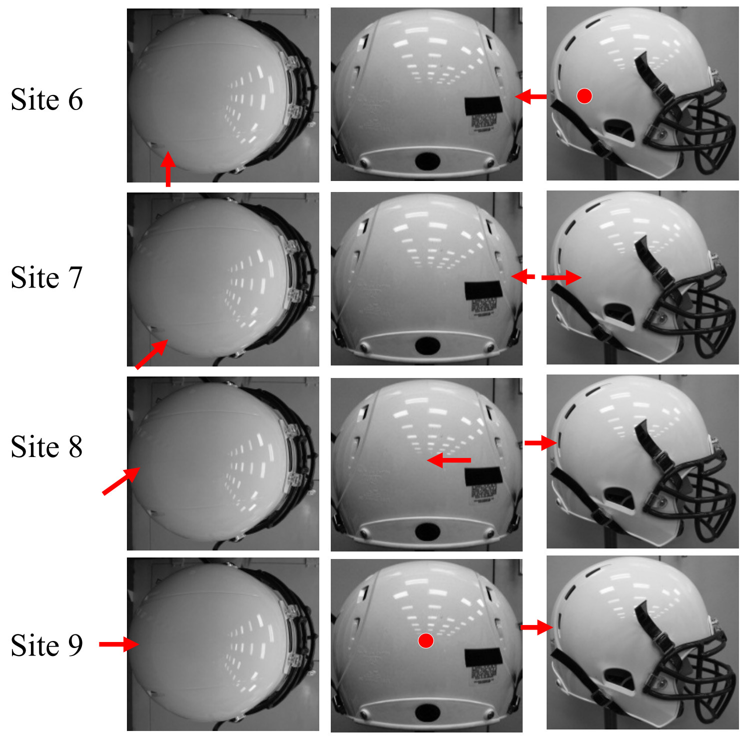

Fig. 3. Impact sites 6-9 on the American football helmets, impact vector indicated by arrow

Computational Modeling

The computational model used in this research is the University College Dublin Brain Trauma Model (UCDBTM) developed by Horgan and Gilchrist [14]-[15]. The geometry of this FE model was developed through CT and MRI scans of a male cadaver. The model is comprised of a scalp, three-layered skull (cortical and trabecular bone), dura, cerebrospinal fluid (CSF), pia, falx, tentorium, cerebral hemispheres, cerebellum and brain stem. In total, the UCDBTM has approximately 26 000 elements. The model was validated against Nahum et al's [16] experiments and Hardy et al's [17] neutral density tracking data from cadaver impacts. Further validations were carried out by Doorly and Gilchrist [18] simulating real world TBI incidents with good results.

Material properties (Tables 1 and 2) of the cortical and trabecular bone, scalp and intracranial membranes for the model were derived from work done by Ruan [19], Willinger et al [20], Zhou et al [21] and Kleiven and von Holst [22]. The brain material properties were taken from Zhang et al [23]. A linearly viscoelastic material model combined with large deformation theory was used to model the brain tissue [14-[15]-[21]-[24]. The compressive behaviour of the brain was defined as elastic. The shear characteristics of the brain were defined as:

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

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor [14]. The skull brain interface was defined by modelling the CSF as solid elements with a high bulk modulus and low shear modulus to allow it to behave as a fluid. The contact definitions at this interface were assigned as no separation and used a friction coefficient of 0.2 [25].

Table 1. Material properties of the entire finite element model

Material	Poisson's Ratio	Density (kg/m^3)	Young's Modulus (Mpa)
Scalp	0.42	1000	16.7
Cortical Bone	0.22	2000	15000
Trabecular Bone	0.24	1300	1000
Dura	0.45	1130	31.5
Pia	0.45	1130	11.5
Falx	0.45	1140	31.5
Tentorium	0.45	1140	31.5
CSF	0.5	1000	-
Grey Matter	0.49	1060	30
White Matter	0.49	1060	37.5

Table 2. Material characteristics of the brain tissue

	Shear Modulus (kPa)			
Material	G_0	G_∞	Decay Constant (Gpa)	Bulk Modulus (s^{-1})
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19
Cerebellum	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19

III. RESULTS

The centric conditions (Table 3) showed significant correlations between linear and rotational acceleration

(0.42), average MPS (0.66) and peak MPS (0.68). Rotational acceleration was significantly correlated to linear acceleration (0.42) and average MPS (0.42). For the non-centric conditions (Table 4) linear acceleration was not significantly correlated to either rotational acceleration or average/peak MPS. Rotational acceleration was significantly correlated to average MPS (0.47) and peak MPS (0.43).

Table 3. Pearson correlations between dynamic response and maximum principal strain for the centric impact conditions. (* 95% confidence interval).

		Linear acceleration	Rotational acceleration	average MPS	Peak MPS
Linear acceleration	Pearson Correlation	1	.422(*)	.662(*)	.675(*)
	Sig. (2-tailed)	.	0.012	0	0
	N	35	35	35	35
Rotational acceleration	Pearson Correlation	.422(*)	1	.422(*)	0.245
	Sig. (2-tailed)	0.012	.	0.012	0.156
	N	35	35	35	35
average MPS	Pearson Correlation	.662(*)	.422(*)	1	.822(*)
	Sig. (2-tailed)	0	0.012	.	0
	N	35	35	35	35
Peak MPS	Pearson Correlation	.675(*)	0.245	.822(*)	1
	Sig. (2-tailed)	0	0.156	0	.
	N	35	35	35	35

Table 4. Pearson correlations between dynamic response and maximum principal strain for the non-centric impact conditions. (* 95% confidence interval).

		Linear acceleration	Rotational acceleration	average MPS	Peak MPS
Linear acceleration	Pearson Correlation	1	-0.27	-0.163	-0.053
	Sig. (2-tailed)	.	0.112	0.341	0.758
	N	36	36	36	36
Rotational acceleration	Pearson Correlation	-0.27	1	.472(*)	.428(*)
	Sig. (2-tailed)	0.112	.	0.004	0.009
	N	36	36	36	36
average MPS	Pearson Correlation	-0.163	.472(*)	1	.919(*)
	Sig. (2-tailed)	0.341	0.004	.	0
	N	36	36	36	36
Peak MPS	Pearson Correlation	-0.053	.428(*)	.919(*)	1
	Sig. (2-tailed)	0.758	0.009	0	.
	N	36	36	36	36

IV. **Discussion**

The results indicate that the centric impacts show correlations between linear acceleration to rotational acceleration and brain deformation. This relationship no longer exists with non-centric impact conditions, where linear acceleration does not have significant correlations to either rotational acceleration or maximum principal strain. The results show that for methodologies which use centric impacts to evaluate risk of injury, the correlation of linear acceleration to brain deformation is moderately high and significant, and for rotational acceleration the correlation only exists for the average values for the cerebrum. For the non-centric condition, linear acceleration has no correlation to either rotational acceleration or brain deformation metrics, whereas rotational acceleration becomes correlated. These results concerning linear acceleration indicate why there is some research indicating linear acceleration as correlated to rotational acceleration [11]. It is likely the research in question used a primary centric impact condition, which would create such a relationship between these two injury metrics. The correlation of rotational acceleration to maximum principal strain is lower than that shown in other research [5]-[26]-[27], which may be due to the nature of the impacts for this protocol. When these results are applied to helmet development and design, they indicate that it is necessary to measure both linear and rotational acceleration when examining the performance of the technology. However, the correlations to brain deformations were relatively low, which may indicate that in the future helmets should be evaluated with measures of brain deformation which may be more closely correlated with concussion [3]-[6]-[7].

This research examined the relationships between linear and rotational acceleration and maximum principal strain. The differences in response (as shown in the Appendix) would be associated with the different energy absorbing liners, helmet shell design, and faceguard construction. While not exclusively the purpose of this study, the results do indicate that it is possible to influence the brain deformation from an impact through helmet design. This research has certain limitations surrounding the use of physical and finite element modelling. The physical model used was intended for use in car crash scenarios and as such may not produce responses accurate to sporting collisions. The Hybrid III headform in particular is constructed of steel and does not produce a biofidelic response to an impact and as a result affects the results. The Hybrid III neck is also not biofidelic and likely affects the resulting dynamic response used for the finite element modelling. Also, to date, there is a lack of validation concerning neckform performance for centric and non-centric impacts. This physical model is also only intended for impacts in the antero-porterior directions and thus the impacts along other axes may not be true to normal human responses. The UCDBTM is a three dimensional simulation of the human head and brain system. As such, the material characteristics and boundary conditions used are assumptions meant to simulate human anatomical responses and therefore produce results which may not be accurate to a live human being. The analysis of the UCDBTM does include the analysis of the deep grey and white matter. The brain stem and cerebellum was not included as these regions lack data to compare to for validation purposes. It would be beneficial to examine the brainstem in particular as it has strong associations with unconsciousness produced from concussive injury.

V. **Conclusions**

The results of this research indicate that the methodology used to simulate impacts to the head will influence the relationships between variables used to indicate risk of injury. Furthermore, these results confirm that it is important to consider both linear and rotational acceleration and brain deformation metrics when determining injury risks and developing protective technologies. This research shows the link between the nature of the impact condition and the output of the response. For this reason it would be necessary to consider these relationships for helmet standardization as well as impact reconstructions. For standards, this research shows the importance of evaluating helmets using centric and non-centric conditions and to use performance metrics in linear and rotational acceleration. When reconstructing injuries for brain threshold research, these results show the importance of considering whether or not the impact vector lies within or without the centre of gravity of the headform, as this can have influence over the stresses and strains which are known to be associated with brain injuries. Finally, when considering the results of the impacts on the football helmets (as shown in the Appendix), the differences that can be found between helmets at certain impact conditions indicate that it is possible to influence the strains incurred by the brain using design characteristics. These results show that using a centric and non-centric impact methodologies for the evaluation of helmet performance may produce unique design elements in both liner and shell technologies intended to reduce the rotational acceleration and brain deformations under these conditions.

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APPENDIX

Table 5. Dynamic response data for the centric sites. Helmet A had a VN liner, B a Skydex liner, and C a three dimensional structure liner.

Site	Helmet Designation	Peak Acceleration Linear	Peak Acceleration Rotational
Site 2	A	62.3 (2.8)	5979 (1292)
	B	60.9 (1.7)	4508 (171.6)
	C	60.7 (3.3)	2643 (460.4)
Site 3	A	71.5 (2.8)	4345 (468.5)
	B	65.4 (4.0)	4590 (872.6)
	C	73.0 (2.9)	4235 (262.3)
Site 5	A	68.4 (1.3)	4272 (177.9)
	B	66.4 (2.5)	4684 (387.5)
	C	80.0 (1.3)	4856 (145.6)
Site 7	A	70.3 (2.1)	4949 (374.7)
	B	72.8 (2.8)	6309 (495.8)
	C	79.2 (5.2)	7561 (868.4)
Site 9	A	81.0 (0.70)	3985 (230.8)
	B	84.7 (2.5)	4106 (132.4)
	C	76.1 (4.7)	6265 (274.3)

Table 6. Dynamic response data for the non-centric sites. Helmet A had a VN liner, B a Skydex liner, and C a three dimensional structure liner.

Site	Helmet Designation	Peak Acceleration Linear	Peak Acceleration Rotational
Site 1	A	65.5 (3.0)	5828 (1748)
	B	64.6 (1.9)	2869 (318.7)
	C	64.7 (3.0)	4072 (161.9)
Site 4	A	75.2 (1.8)	4761 (617.6)
	B	75.2 (4.9)	4681 (153.5)
	C	78.8 (1.3)	4317 (244.7)
Site 6	A	49.9 (1.5)	5131 (276.0)
	B	52.6 (1.9)	5519 (275.1)

	C	57.2 (1.7)	5049 (75.35)
Site 8	A	64.7 (0.96)	4483 (582.4)
	B	71.3 (2.7)	4748 (101.7)
	C	67.4 (0.50)	4696 (255.6)

Table 7. Finite element modeling results for the centric sites. Helmet A had a VN liner, B a Skydex liner, and C a three dimensional structure liner.

		Maximum Principal Strain	
Site	Helmet Designation	Peak	Average
Site 2	A	0.215 (0.084)	0.113 (0.017)
	B	0.162 (0.017)	0.106 (0.027)
	C	0.172 (0.027)	0.121 (0.006)
Site 3	A	0.375 (0.010)	0.235 (0.003)
	B	0.291 (0.067)	0.178 (0.019)
	C	0.375 (0.050)	0.212 (0.032)
Site 5	A	0.359 (0.011)	0.172 (0.005)
	B	0.377 (0.014)	0.183 (0.012)
	C	0.326 (0.008)	0.168 (0.005)
Site 7	A	0.337 (0.002)	0.228 (0.011)
	B	0.330 (0.020)	0.226 (0.014)
	C	0.413 (0.009)	0.284 (0.003)
Site 9	A	0.297 (0.027)	0.197 (0.014)
	B	0.277 (0.006)	0.188 (0.001)
	C	0.385 (0.019)	0.265 (0.013)

Table 8. Finite element modeling results for the non-centric sites. Helmet A had a VN liner, B a Skydex liner, and C a three dimensional structure liner.

		Maximum Principal Strain	
Site	Helmet Designation	Peak	Average
Site 1	A	0.169 (0.019)	0.112 (0.024)
	B	0.131 (0.004)	0.075 (0.003)
	C	0.139 (0.005)	0.095 (0.005)
Site 4	A	0.448 (0.069)	0.233 (0.010)
	B	0.327 (0.069)	0.223 (0.015)

	C	0.405 (0.063)	0.221 (0.026)
Site 6	A	0.399 (0.021)	0.273 (0.007)
	B	0.451 (0.034)	0.280 (0.008)
	C	0.384 (0.008)	0.250 (0.016)
Site 8	A	0.363 (0.049)	0.253 (0.024)
	B	0.332 (0.003)	0.252 (0.003)
	C	0.365 (0.017)	0.264 (0.017)