The Effect of Impact Compliance, Velocity, and Location in Predicting Brain Trauma for Falls in Sport

R. Anna Oeur, T. Blaine Hoshizaki

Abstract Peak linear and angular acceleration are commonly used to measure concussion risk. In efforts to better understand the importance of loading factors affecting head dynamic response, the purpose of this study was to examine the effects of impact velocity, location, and compliance on head acceleration. A Hybrid III headform and modified neckform attached to a drop tower were subject to 4 impact velocities (1.5, 3.0, 4.5, 6.0 m/s) and 3 surfaces: steel, vinyl nitrile foam, and Rubatex R338 rubber foam to represent falls onto a hard surface, a helmeted fall (i.e. football helmet), and a gymnastics mat, respectively. Each combination of velocity and compliance were tested at front, front boss, side, and rear boss negative azimuth. Regression analysis were conducted for each impact location and revealed that compliance had a consistently stronger influence on head dynamic response than velocity.

Keywords dynamic Response, fall events in sport, Hybrid III headform, impact variables.

I. INTRODUCTION

Sports are a major source of concussion as a result of the number of people who participate as well as the frequency of this injury in the population [1-4]. In Canada, over 7.2 million people participated in sports in 2010 [5], contributing to nearly 1.5 million suffering short and long-term consequences [6]. In the US, sports-related accidents have been estimated to cause up to 3.8 million concussions [7]. Concussive injuries are characterised by clinical signs and symptoms that typically resolve within one week post-injury; however, in severe cases, prolonged symptoms can last from months to years [8-9]. In worst-case scenarios, the long-term effects of concussion cause disability that affects work and school, contributing to the overall economic burden [8].

Falls and collisions are among the most common causes of concussion in sport [2][10-12]. Concussions as a result of a fall present a unique element of risk as the head falls from a height and makes contact with the rigid, unyielding earth, where the majority of the impact energy is transferred to accelerate the head. In a collision, the two striking bodies can move relative to each other before and after the impact, which will affect the nature of the impact energy transferred to the head. The focus of this study will examine fall-type events. The rate of energy transfer can be modulated by adding compliance, either to the head by wearing a helmet, or putting foam or a mat on the ground [13]. The type of surface (i.e., turf, grass, ice, wooden gym floors) and whether the head is helmeted will affect impact energy attenuation, changing the nature of loading [14]. Therefore, the variables that influence head injury risk from falls include impact velocity (height of the fall) and impact compliance (overall stiffness of an impact). An unprotected head colliding with a rigid, stiff surface corresponds to a low compliance condition, whereas a head colliding with a soft surface would be a high compliance condition. Lastly, impact location on the head has been identified as a factor that influences head injury severity. [15] examined the effect of impact location on the duration of loss of consciousness of primates noting that side impacts were associated with a decrease tolerance to concussion. [16-17] attributed the directional sensitivity to skull and brain geometry and interior structures, like the falx and tentorium that play a role in overall brain trauma effects. They found that coronal motions resulted in the largest amounts of axonal damage to the primate brain.

R. A. Oeur (MSc) is a PhD student (613-562-5800 ext. 7207, anna.oeur@uottawa.ca) and T.B. Hoshizaki (PhD) is a Professor of Biomechanics in the School of Human Kinetics at University of Ottawa in Ottawa, Canada.

Commonly used kinematic variables to measure head injury risk include peak linear and angular acceleration. Early experimental research on cadaver and animals demonstrated that linear acceleration correlated highly with dynamic pressure responses in the brain [18]. These pressure gradients resulted in intracranial stresses that disrupt the neural tissue function responsible for concussive traumas [19]. In addition, angular acceleration was noted to be important for causing trauma due to the relative weakness of the brain to shear strain [20]. In a number of animal experiments conducted by [21] and [22], increasing levels of angular acceleration were linked with increased shear strain injuries in the brain [21-23].

Sports-related falls can occur under a variety of conditions: falling onto wooden gym floors such as in basketball or cheerleading, falling on grass with or without a helmet, as in soccer or American football, and onto mats or crash pads, common in wrestling and gymnastics [2]. The variety of impact surfaces or compliances possible as well as the various impact velocities and locations on the head contribute to the wide ranging potential conditions that may cause risk of injury. Previous research has demonstrated that impact velocity increases head acceleration, whereas compliance decreases peak values but in doing so, elongates the acceleration pulse [13][24]. Head impacts to different locations on a Hybrid III headform has been shown to result in unique acceleration responses, where non-centric vectors tend to result in relatively higher angular accelerations [25]. The majority of this research examined impact variables in isolation; however, the authors hypothesize that impact velocity, compliance, and location interact to influence head dynamic response. These interactions likely contribute to the challenges with predicting head injury risk using head acceleration. The purpose of this study was to describe the effects of impact velocity, compliance, and location on peak resultant linear and angular acceleration of the head, and to determine which factor has the strongest influence on response.

II. METHODS

Equipment

Head dynamic response was collected using a 50th percentile adult male Hybrid III headform instrumented with 9-uniaxial linear accelerometers (Endevco 7264C-2KTZ-2-300, California, USA) positioned in a 3-2-2-2 array, to capture linear and angular acceleration [26]. A Hybrid III headform (Michigan, USA) was used in conjunction with a customised neckform developed at the University of Ottawa (Ottawa, Canada), and composed of separate circular plates held together with a standard Hybrid III neck cable. The non-directional neck was comprised of four steel circular plates, one steel top plate, and four butyl-rubber disks weighing 1.30 kg. This neckform was used in place of the standard Hybrid III neckform (weight = 1.40 kg) to ensure uniform directional response. Acceleration signals were sampled at 20 kHz and filtered using a CFC class 1000 filter, specifying a low-pass filter with a 1650 Hz cutoff frequency [27].

A monorail drop tower was used to simulate simplified fall-type events and consists of a six-metre tower, a sliding carriage and motor, and a quick release anvil adapter attached to a concrete base. The headform and neckform were attached to a sliding carriage and glides over the tower rail via ball-bearings. The motor lifts the headform, neckform, and sliding carriage unit to a specified height and the unit is released in a guided drop to obtain the desired test velocity. A flat steel impact anvil was attached to the anvil adapter to serve as the test surface. Due to the numerous impact scenarios that are possible within the sports environment, the authors were tasked with selecting levels of impact variable that are representative of the possible range of conditions, but to also balance the number of permutations of each variable, and total number of impacts. Three impact surfaces were selected to represent a range of different types of compliance representative of causing concussion from falls in sports. Steel was selected to represent a non-compliant condition (5 ms duration event), such as unprotected falls onto rigid surfaces i.e., ice surface in figure skating. A 0.025m thick vinyl nitrile foam was used to obtain a 15ms duration response characteristic of protected or helmeted fall [28]. A 0.067m thick Rubatex R338 rubber foam was selected to represent falls onto well-padded surfaces, such as a gymnastics mat, lasting approximately 25ms in duration. Figure 1 demonstrates the impact surface (in solid lines) and the respective anvils used to match the linear acceleration response of each. Each compliance was subject to Shore A testing to determine the relative stiffness of each material (ASTM Standard D2240), where 3.0, 17.8 and 99.3 correspond to the Rubatex rubber foam, vinyl nitrile foam, and steel surface, respectively (Figure 1).

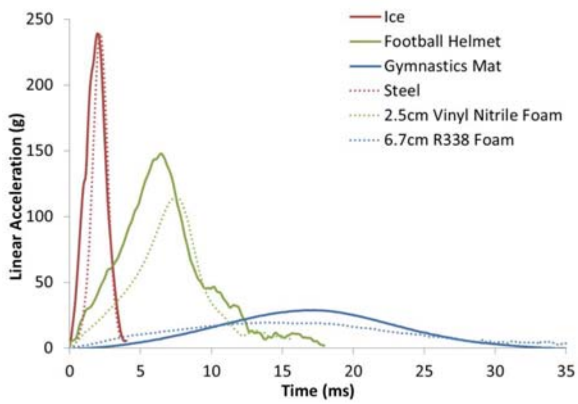

Fig 1. Linear acceleration-time (ms) for bare head falls onto ice, football helmet to ground, and head to gymnastics mat shown in solid lines. Dotted lines depict time histories for steel, vinyl nitrile foam, and R338 foam for similar pulse durations.

Fig 2. Impact locations demonstrated on steel anvil.

Experimental Testing

Four impact velocities: 1.5, 3.0, 4.5, 6.0 m/s were selected to capture the range of plausible falling events. These levels of velocity are consistent with those reported by [29] who investigated different falling scenarios using the full body Hybrid III anthropometric test dummy (ATD) for front, side, and backward falls (2.9 - 5.8 m/s). Four impact locations were selected to capture motion among different planes. Front centre gravity (FCG) and side centre gravity (SCG) have motions occurring in primarily the sagittal and frontal planes, respectively. The front boss centre gravity (FBCG) location will cause motion along a plane that is mid-way between the frontal and sagittal planes. Lastly, a non-centric impact condition was chosen to elicit a high rotational response; the rear boss negative azimuth (RBNA) has the rear boss impact site with a 45° rotation in the horizontal plane, aligning the vector in the same orientation as the SCG but occurring through the rear boss site (Fig 2). These conditions are consistent with previous research conducted by [25] eliciting low and high levels of injury risk based on linear and angular acceleration of the Hybrid III headform [30]. The Impact variables (velocity, compliance, and location) were selected to capture the range of dynamic response associated with concussive events, however, combinations of impact variables producing responses above 500g were avoided due to equipment constraints and represents risk well above concussion [30-32]. As a result, the steel condition (A 99.3) was only tested at 1.5 and 3.0 m/s and the vinyl nitrile (A 17.8) and Rubatex foams (A 3.0) were tested at four velocities for all locations. 40 unique conditions (A 99.3: 2x4; A 17.8: 4x4; and A 3.0: 4x4) were tested for three trials, for a total of 120 impacts. Stepwise multiple linear regression is a type of regression analysis where the sequence of independent variables input into the predictive models are based on the correlation matrix, where the independent variable with the highest Pearson product-moment correlation coefficient is input first, followed by the next highest correlated variable and so on. Unstandardized beta coefficients in each model provide an indication of the incremental effect of the independent variable in its original units (in this study, Shore A stiffness value or m/s) on dependent variable (linear or angular acceleration). These values are the slope coefficients in the formula for the predictive equation. The standardized beta coefficients convert the effects of the independent variable into *z*-score equivalents, allowing for a better comparison of the relative influence of each independent or predictor variable on dependent variable. For this part of the study, impact location was treated as a categorical variable, therefore, step-wise multiple linear regressions were run to determine the relative influence of compliance and velocity (examination of Beta coefficients) at each location for linear and angular acceleration ($p < 0.05$). To test for main effects and interactions, a 3-way ANOVA for velocity, location, and compliance were conducted on the dependent variables. Further analyses include running one-way ANOVAs and Tukey *post hoc* tests for location (FCG, FBCG, SCG and RBNA) at each level of velocity (1.5, 3.0, 4.5, 6.0 m/s; $p < 0.05$) under each compliance.

III. RESULTS

Regression Analysis

Significant multiple linear regressions were found at each impact location for linear and angular acceleration. The unstandardized beta coefficients (*B*), corresponding standard error values (SE *B*), and standardized beta coefficients (Beta) are reported in Table I and II along with F ratios and R^2 values. The addition of impact velocity as a predictor variable in the multiple regression analysis significantly improved the R^2 value, therefore the discussion will focus on *Model 2* parameters in the tables. The larger Beta coefficients for impact compliance relative to velocity, demonstrates that this factor has a stronger influence on linear and angular acceleration. These findings were consistent for all locations. Additionally, trends in the Beta values (across location) seem to suggest that compliance tends to be more influential on angular acceleration (0.823 – 0.991) than it is to linear acceleration (0.732-0.961). On the contrary, velocity tends to be slightly more influential on linear acceleration (0.535 – 0.748) than it is to angular acceleration (0.317 - 0.703).

TABLE I. MULTIPLE LINEAR REGRESSION RESULTS FOR LINEAR ACCELERATION

Location	Predictor Variable	Model 1			Model 2		
		B	*SE B*	*Beta*	*B*	*SE B*	*Beta*
Front	*Compliance*	1.887	0.467	0.606**	2.680	0.315	0.862**
	Velocity				45.940	6.852	0.678**
	R^2	0.368			0.763		
	F	16.290**			43.404**		
Front Boss	*Compliance*	1.707	0.502	0.540*	2.533	0.320	0.802**
	Velocity				50.667	7.013	0.732**
	R^2	0.292			0.759		
	F	11.539*			42.415**		
Side	*Compliance*	1.794	0.281	0.770**	2.239	0.191	0.961**
	Velocity				27.302	4.177	0.535**
	R^2	0.593			0.842		
	F	40.802*			72.158**		
Rear Boss	*Compliance*	1.134	0.379	0.492*	1.750	0.247	0.759**
	Velocity				37.88	5.420	0.748**
	R^2	0.242			0.729		
	F	8.935*			36.737**		

* $p \le 0.05$; ** $p \le 0.001$

Table II. multiple linear regression results for Angular acceleration

Location	Predictor Variable	Model 1			Model 2		
		B	*SE B*	*Beta*	*B*	*SE B*	*Beta*
Front	*Compliance*	142.99	23.878	0.748**	177.127	19.363	0.930**
	Velocity				2004.181	421.564	0.483**
	R^2	0.560			0.760		
	F	35.614**			42.846**		
Front Boss	*Compliance*	93.197	18.020	0.699*	123.238	11.161	0.924**
	Velocity				1842.073	244.669	0.630**
	R^2	0.489			0.835		
	F	26.748*			63.313**		
Side	*Compliance*	217.161	22.455	0.877**	245.204	19.277	0.991**
	Velocity				1719.583	422.587	0.317**
	R^2	0.770			0.857		
	F	93.530*			81.029**		
Rear Boss	*Compliance*	125.923	34.210	0.571*	181.379	22.366	0.823**
	Velocity				3400.576	490.302	0.703**
	R^2	0.326			0.758		
	F	13.549*			42.223**		

* $p \leq 0.05$; ** $p \leq 0.001$

Effect of Location

Significant main effects for impact velocity, compliance, and location as well as an interactions between each variable were found for linear acceleration (F (12,80) = 937.621 , p = <0.001) and angular acceleration (F (12,80) = 7.738, p = <0.001). Figures 3 and 4 are plots of peak resultant acceleration by velocity for each impact location. For linear acceleration (Fig 2), head impact results from [13] cadaver head impact tests are approximated and serve as a reference. Statistically significant results for steel (A 99.3) show that at 1.5 m/s, the side (133 g) had the highest values and rear boss the lowest (74g; p<0.05), however as impact velocity increased (3.0m/s), the front had the highest value (359 g) followed by front boss (343g), side (296g), and rear boss (240g; p < 0.05). For angular acceleration (Fig 3), the side consistently produced the highest responses (14 900 and 31 000 rad/s^2), followed by rear boss, front, and front boss with the lowest values at 3.0 m/s (p<0.05).

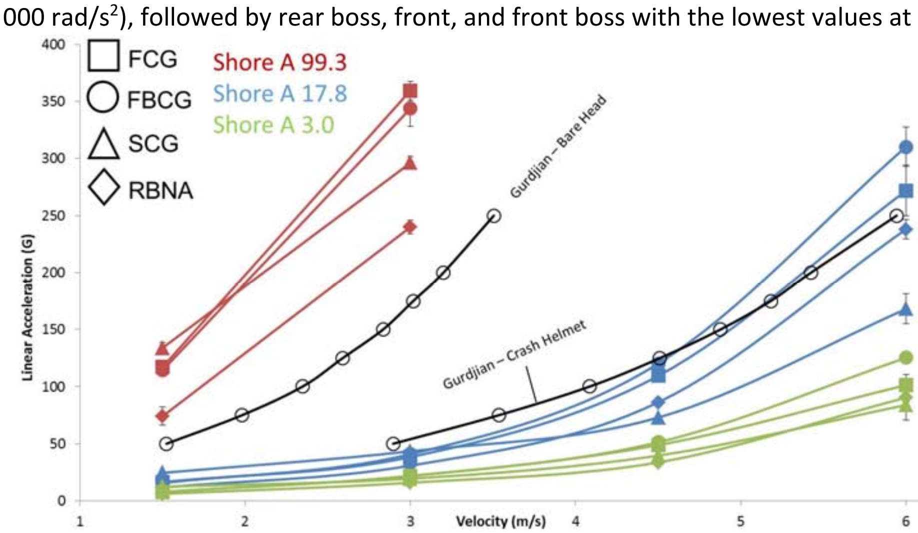

Figure 3. Peak resultant linear acceleration against velocity for low (A 99.3), medium (A 17.8), and high (A 3.0) compliance conditions plotted with [13] cadaver data.

For a more compliant surface (0.025m vinyl nitrile foam; Shore A 17.8), significant differences were noted at the lower velocities, however these differences are not meaningful until 4.5 m/s where all locations were significantly different from each other (73-120g), with the front boss having the highest results at 4.5 and 6.0 m/s ($p<0.05$). For angular acceleration, it was the rear boss location (non-centric condition) that displayed higher values across velocity ($p<0.05$).

At the highest compliance (0.067m Rubatex foam; A 3.0), significant differences in peak resultant linear acceleration were not observed until 4.5 m/s. The front boss had the highest values, followed by front, then side and rear with the lowest values ($p<0.05$). For angular acceleration, the rear boss had the highest values starting at 3.0 m/s ($p>0.05$)

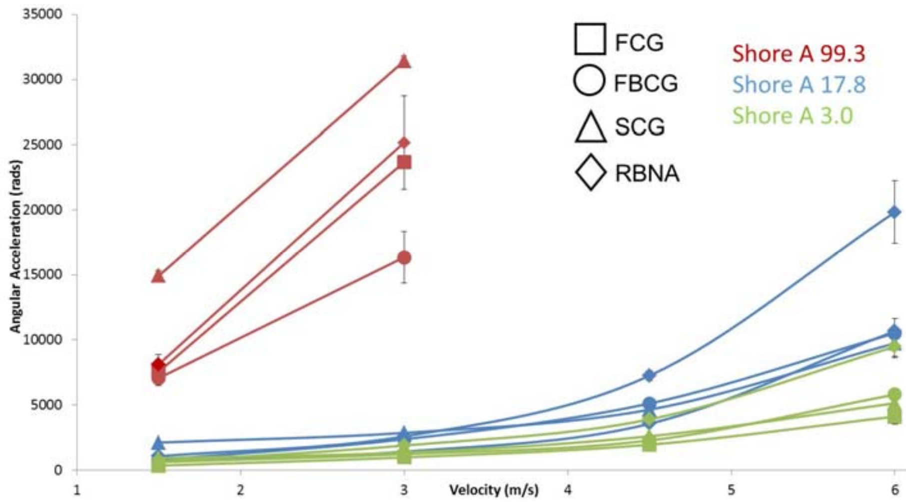

Figure 4. Peak resultant angular acceleration against velocity for low (A 99.3), medium (A 17.8), and high (A 3.0) compliance conditions.

IV. DISCUSSION

Peak resultant linear and angular acceleration are among the most commonly used kinematic measures of concussion risk. Injury reconstruction research involving simulations of head impact events in contact sports such as American football and rugby, as well as instrumented helmets used to measure head acceleration within games and practices, has provided valuable data sets with which researchers can interpret the levels of loading associated with injury probability [33-37]. The intention of this study was to supplement the understanding of concussion risk by describing *how* impact variables (compliance, velocity, and location) contribute to creating magnitudes of head acceleration.

[13] had conducted similar research in 1964, using cadavers subjected to full-body drop tests onto a rigid anvil with and without a crash helmet. They tested a number of velocities and measured peak linear acceleration. An approximation of their results are plotted in Figure 3, where adding a helmet (compliance) decreases the magnitudes of peak linear acceleration, thus higher velocities are required to achieve similar levels of acceleration. The headform drops onto steel tend to be stiffer in comparison to the cadaver test and is likely due to the lack of biofidelity of the Hybrid III headform at high energies, but the results from the crash helmet from [13]'s data fits within the medium compliance results selected to represent a helmeted condition. Adding compliance to the impact, results in lowering peak linear acceleration but extending the duration of the pulse. The trade-off between decreasing the magnitude by spreading out the force footprint, and elongating the acceleration pulse is no new concept to those designing head protection as this is precisely how

helmets work [14][38].

In an impact, velocity and compliance are two critical factors that influence the total amount of energy transferred to the head. What the authors were interested in evaluating was *which* of these impact variables contribute the most to creating peak head acceleration in environments that involve falling onto rigid and compliant surfaces common to sport environments. Overall, the regression analysis revealed that impact compliance, is a stronger contributor to both peak linear and angular acceleration than impact velocity, meaning that falling onto stiffer surfaces contributes to increasing head acceleration faster than velocity. In other words, efforts should prioritize decreases in the stiffness of an impact, over attempting to mitigate impact velocity in general falling scenarios, when risk is defined by peak head acceleration. While there is overlap in the Beta coefficients, data trends suggest that compliance tends to be more influential on angular acceleration than linear, however velocity may be more influential for linear than angular acceleration (that is, second to impact compliance). The relative contribution of compliance and velocity is dependent upon location however; the side was the most influenced by compliance (Beta values = 0.961 and 0.991 for linear and angular acceleration respectively) and the least influenced by velocity (0.535 and 0.317) of all locations.

The role of impact location on head injury risk has been demonstrated throughout the literature. Based on animal research, the brain tends to have a decreased tolerance to concussive effects from lateral loading, or side impacts [15][17]. This is consistent with findings from [37] and [39] who conducted video analysis of Australian rules football and rugby and found that impacts to temporal area were the most prevalent in resulting in concussion. On the contrary, impacts to the front were more prevalent for injury among high school American football players, whereas impacts to the side and rear were more common among elite players [40-41]. While it is difficult to conclude a location based human tolerance to head injury from the research conducted across various sports, it is not unreasonable to suggest that the head and brain have a directional sensitivity to injury [17][22][42].

The location based findings in this study do not necessarily support a directional injury tolerance but simply illustrate that location plays a role in the magnitude of head acceleration. While significant differences were found, many of these differences were not necessarily meaningful, especially at the lower velocities. Impacts to the front or front boss tend to result in relatively higher values of linear acceleration across compliance and velocity. This is likely a result of these impacts occurring through the centre of gravity of the head as well as contribution from the mass of the neck due to its slightly raised orientation in relation to the anvil at the base of the drop tower. The neckform used in this study was chosen to represent the mass of neck without any directional biases. The standard Hybrid III neckform has directional constraints (particularly in extension) and while these effects are appropriate to simulate neck flexion/extension, they are not well described under high dynamic loads from direct impacts. Therefore to remove any potential directional biases associated with the neck at higher impact energies, to maintain the neck mass, and to isolate the effects of the impact variables sought, the non-directional neckform was used.

For angular acceleration, it was hypothesized that the rear boss negative azimuth location would result in the highest values across all levels of compliance and velocity, since this vector is located outside the centre gravity. The side had considerably higher angular accelerations than all other locations for the steel. For compliant conditions, the rear boss negative azimuth produced the highest values which is consistent with previous research, demonstrating that non-centric impacts result in higher head rotations [16][25]. The high responses at the side are likely due to the close proximity of the accelerometers with respect to the impact, the tethering effect of the neckform, as well an interaction between the headform surface (including vinyl skin) and the anvil [43-44]. The headform is comprised of hollowed steel covered with a vinyl outer skin that takes on an ellipsoid shape. These physical characteristics of the Hybrid III headform as well as the local contact characteristics associated with the three different types of surfaces (steel, VN and Rubatek foams) interact in a manner that create unique acceleration-time traces. An example of the resultant linear acceleration-time traces for different locations are illustrated in Figure 5 for 3.0 m/s. For steel, impact energy is rapidly transferred to the head and therefore it is hypothesized that the local geometry of the head and the vinyl skin contribute to the unique responses observed. For the compliant conditions (A 17.8 and A 3.0), the foams increased the contact time between the head and anvil, likely allowing both the head and neck to rotate as a unit throughout entire pulse, especially at the non-centric rear boss NA location. Although the authors did not measure contact characteristics, such as peak force and force footprint in this study, these factors would help to explain the contact phenomena associated with each location and how the impact force is transferred from the anvil to the headform. In fall-type events, the bulk of the impact energy is translated into compressing and shearing of the compliant materials on headform (i.e., the vinyl skin) or anvil. The same types of processes likely occurs under collision-type events but to a different extent due to two relatively movable masses colliding. While surface geometries influence the contact mechanics during an impact, flat surfaces (steel and foam) were used in this study to simplify the number of conditions tested in a nearly fully-crossed research design.

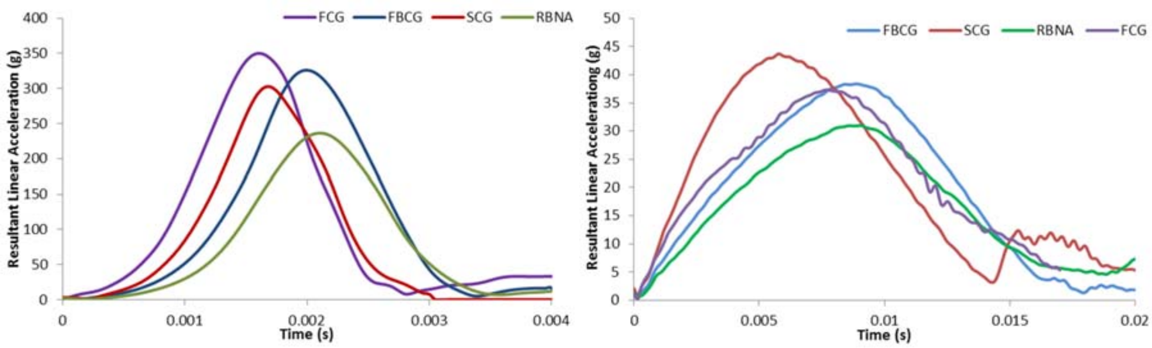

Figure 5. Exemplar resultant linear acceleration-time curves at 3.0 m/s for Shore A 99.3 (steel) on the left and Shore A 17.8 (VN foam) on the right.

An understanding of the types of effects impact variables have on head acceleration is an important aspect of developing strategies for head protection. While decreasing the overall stiffness of an impact during a fall should be prioritized in keeping head accelerations at a low level, interactions between compliance, velocity, and location contribute to the difficulty in predicting head injury risk. Historically, head protection has been designed to manage primarily linear acceleration, and while it has been rather successful [13][45], it is important that levels of angular acceleration are not increased, for example the rear boss NA in this study had the lowest levels of linear acceleration but high levels of angular at the medium compliance.

V. CONCLUSIONS

Compliance, or stiffness, was more influential than velocity on peak linear and angular acceleration for fall-type events in sport. Interactions between compliance, velocity, and location contribute to the challenges with predicting injury risk based on head impact events. Impact location alone contributes to unique magnitudes of linear and angular acceleration, as in the case with the rear boss NA (non-centric condition), characteristic of relatively low linear acceleration, but high angular for compliant surfaces. These findings suggest that effective strategies for managing injury risk from head acceleration during fall-type events in sport, should first consider mechanisms to decrease the stiffness of the impact. Secondly, when working within the confines of a level of compliance, strategies should include managing both linear and angular acceleration for specific impacts locations.

VI. ACKNOWLEDGEMENT

This research was supported by the Natural Sciences and Engineering Research Council of Canada (NSERC) Postgraduate Doctoral Scholarship Program.

VII. REFERENCES

[1] Bazarian, J.J., McClung, J., et al. Mild traumatic brain injury in the united states, 1998-2000. *Brain Injury*, 2005. **19**(2): p.85-91

[2] Daneshvar, D.H., Nowinski, C.J., McKee, A.C., Cantu, R.C. The epidemiology of sport-related concussion. *Clinics in Sports Medicine*, 2011. **30**(1): p.1-17, vii

[3] Gavett, B.E., Stern, R.A., McKee, A.C. Chronic traumatic encephalopathy: A potential late effect of sport-related concussive and subconcussive head trauma. *Clinics in Sports Medicine*, 2011. **30**(1): p.179-88, xi

[4] Tator, C. Sport concussion education and prevention. *Journal of Clinical Sport Psychology*, 2012. **6**(3): p.293-301

[5] C. Heritage. Sport participation 2010: Research paper. 2013, Statistics Canada: Canada. Accessed on March 16, 2016: http://publications.gc.ca/collections/collection_2013/pc-ch/CH24-1-2012-eng.pdf

[6] Billette, J.-M., Janz, T. Injuries in canada: Insights from the canadian community health survey. 2011, Statistics Canada: Canada. Accessed on March 16, 2016: http://www.statcan.gc.ca/pub/82-624-x/2011001/article/11506-eng.pdf

[7] Langlois, J.A., Rutland-Brown, W., Wald, M.M. The epidemiology and impact of traumatic brain injury: A brief overview. *The Journal of Head Trauma Rehabilitation*, 2006. **21**(5): p.375-8

[8] Rimel, R.W., Giordani, B.M.A., Barth, J.T., Boll, T.J., Jane, J.A. Disability caused by minor head injury. *Neurosurgery*, 1981. **9**(3): p.221-228

[9] Marshall, S., Bayley, M., McCullagh, S., Velikonja, D., Berrigan, L. Clinical practice guidelines for mild traumatic brain injury and persistent symptoms. *Canadian Family pPhysician*, 2012. **58**(3): p.257-267

[10] Hanlon, R.E., Demery, J.A., Martinovich, Z., Kelly, J.P. Effects of acute injury characteristics on neuropsychological status and vocational outcome following mild traumatic brain injury. *Brain Injury*, 1999. **13**(11): p.873-887

[11] Rimel, R.W., Giordani, B., Barth, J.T., Jane, J.A. Moderate head injury: Completing the clinical spectrum of brain trauma. *Neurosurgery*, 1982. **11**(3): p.344-351

[12] Cantu, R.C. Cerebral concussion in sport: Management and prevention *Sports Medicine*, 1992. **14**(1): p.64-74

[13] Gurdjian, E.S., Hodgson, V.R., Hardy, W.G., Patrick, L.M., Lissner, H.R. Evaluation of the protective characteristics of helmets in sports. *The Journal of Trauma*, 1964. **4**: p.309-24

[14] Hoshizaki, T.B., Brien, S.E. The science and design of head protection in sport. *Neurosurgery*, 2004. **55**(4): p.956-66

[15] Hodgson, V., Thomas, L., Khalil, T. The role of impact location in reversible cerebral concussion. *Proceedings of Twenty-Seventh Stapp Car Crash Conference* 1983. San Diego, California, USA.

[16] Gennarelli, T., Thibault, L.E., et al. Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration. *Proceedings of 31st Stapp Car Crash Conference*, 1987. New Orleans, LA, USA.

[17] Gennarelli, T.A., Thibault, L.E., et al. Diffuse axonal injury and traumatic coma in the primate. *Annals of Neurology*, 1982. **12**(6): p.564-74

[18] Gurdjian, E., Lissner, H., Patrick, L. (1963). Concussion: Mechanism and pathology. *Proceedings of 7th Stapp Car Crash Conference,* p.470-482.

[19] Gurdjian, E.S., Webster, J.E., Lissner, H.R. Mechanism of scalp and skull injuries, concussion, contusion and laceration. *Journal of Neurosurgery*, 1958. **15**(2): p.125-8

[20] Holbourn, A.H.S. Mechanics of head injuries. *The Lancet*, 1943. **242**(6267): p.438-441

[21] Ommaya, A.K., Gennarelli, T.A. Cerebral concussion and traumatic unconsciousness. Correlation of experimental and clinical observations of blunt head injuries. *Brain: A journal of Neurology*, 1974. **97**(4): p.633-54

[22] Gennarelli, T.A., Thibault, L.E., Graham, D.I. Diffuse axonal injury: An important form of traumatic brain damage. *The Neuroscientist*, 1998. **4**(3): p.202-215

[23] Gennarelli, T., Thibault, L. Acceleration damage to the brain. *Proceedings of Advisory Group for Aerospace Research and Development. AGARD-Conference Proceedings-322*, 1982. Cologne, Germany.

[24] Gimbel, G.M., Hoshizaki, T.B. Compressive properties of helmet materials subjected to dynamic impact loading of various energies. *European Journal of Sport Science*, 2008. **8**(6): p.341-349

[25] Walsh, E.S., Rousseau, P., Hoshizaki, T.B. The influence of impact location and angle on the dynamic impact response of a Hybrid III headform. *Sports Engineering*, 2011. **13**(3): p.135-143

[26] Padgaonkar, A.J., Krieger, K.W., King, A.I. Measurement of angular acceleration of a rigid body using linear accelerometers. *Journal of Applied Mechanics*, 1975. **42**(3): p.552-556

[27] Instrumentation for impact test 1995, SAE International: United States. Accessed on March 16, 2016: https://law.resource.org/pub/us/cfr/ibr/005/sae.j211-1.1995.pdf

[28] Hodgson, V.R. and Thomas, L.M. Effect of long-duration impact on head. *Proceedings of 16th Stapp Car Crash Conference*, 1972. Detroit, MI, USA

[29] Hajiaghamemar, M., Seidi, M., Ferguson, J., Caccese, V. Measurement of head impact due to standing fall in adults using anthropomorphic test dummies. *Annals of Biomedical Engineering*, 2015. **43**(9): p.2143-2152

[30] Zhang, L., Yang, K.H., King, A.I. A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering*, 2004. **126**(2): p.226 - 236

[31] Lissner, H.R., Lebow, M., Evans, F.G. Experimental studies on the relation between acceleration and intracranial pressure changes in man. *Surgery, Gynecology & Obstetrics*, 1960. **111**: p.329-38

[32] Oeur, R.A., Karton, C., et al. A comparison of head dynamic response and brain tissue stress and strain using accident reconstructions for concussion, concussion with persistent postconcussive symptoms, and subdural hematoma. *Journal of Neurosurgery*, 2015. **123**(2): p.415-22

[33] Rowson, S., Duma, S., et al. Rotational head kinematics in football impacts: An injury risk function for concussion. *Annals of Biomedical Engineering*, 2012. **40**(1): p.1-13

[34] Pellman, E.J., Viano, D.C., Tucker, A.M., Casson, I.R., Waeckerle, J.F. Concussion in professional football: Reconstruction of game impacts and injuries. *Neurosurgery*, 2003. **53**(4): p.799-814.

[35] Newman, J., Beusenberg, M., et al. A new biomechanical assessment of mild traumatic brain injury. Part I: Methodology. *Proceedings of the International Conference on the Biomechanics of Impact (IRCOBI),* 1999*,* Sitges, Spain.

[36] Patton, D.A., McIntosh, A.S., Kleiven, S. The biomechanical determinants of concussion: Finite element simulations to investigate brain tissue deformations during sporting impacts to the unprotected head. *Journal of applied biomechanics*, 2013. **29**(6): p.721-30.

[37] McIntosh, A.S., Patton, D.A., et al. The biomechanics of concussion in unhelmeted football players in australia: A case–control study. *BMJ open*, 2014. **4**(5).

[38] Newman, J.A., Biomechanics of Head Trauma: Head Protection, In "*Accidental injury*", pp. 303-323, Springer New York, 2002.

[39] McIntosh, A.S., McCrory, P., Comerford, J. The dynamics of concussive head impacts in rugby and australian rules football. *Med Sci Sports Exerc*, 2000. **32**(12): p.1980-4

[40] Kerr, Z.Y., Collins, C.L., et al. Impact locations and concussion outcomes in high school football player-to-player collisions. *Pediatrics*, 2014. **134**(3): p.489-96

[41] Pellman, E.J., Viano, D.C., Tucker, A.M., Casson, I.R. Concussion in professional football: Location and direction of helmet impacts-part 2. *Neurosurgery*, 2003. **53**(6): p.1328-40.

[42] Newman, J.A. The influence of time duration as a failure criterion in helmet evaluation. *Proceedings of International Off-Highway Meeting & Expo*, 1982. Milwaukee, WI, USA

[43] Bartsch, A., Benzel, E., Miele, V., Morr, D., Prakash, V. Hybrid III anthropomorphic test device (ATD) response to head impacts and potential implications for athletic headgear testing. *Accident Analysis & Prevention*, 2012. **48**: p.285-291
[44] Loyd, A.M., Nightingale, R.W., et al. Impact properties of adult and atd heads. *Proceedings of IRCOBI Conference,* 2012*, Dublin, Ireland*.
[45] Cantu, R.C., Mueller, F.O. Brain injury-related fatalities in american football, 1945-1999. *Neurosurgery*, 2003. **52**(4): p.846-52.