Comparison of Dynamic versus Static Head Impact Reconstruction Methodology by Means of Dynamic Impact Response and Brain Deformation Metrics.

Marshall Kendall, Andrew Post, Katrina Zanetti, Thomas Blaine Hoshizaki and Michael D. Gilchrist

Abstract Current reconstructions of head impacts in sports utilize systems which involve a static headform. Many impacts in sport, such as football and ice hockey, typically involve two bodies colliding at velocities in different directions. As a result, current reconstruction techniques using a static component may not fully represent the actual impact event and thus produce different headform responses. Therefore, the purpose of this study is to compare the dynamic impact response and brain deformation metrics from two distinct impact reconstruction methods: 1) static headform (no neckform) and 2) dynamic headform (no neckform). The results of this study show differences between the two impact methodologies in terms of the three-dimension dynamic response loading curves and peak MPS. These differences are likely due to differences in characteristics of dynamic response time-history curves driving the finite element model producing higher strain within the brain tissue. This study emphasizes the importance of accurate impact reconstructions to better assess head injury mechanisms. This study may provide useful insight into the development of specific impact reconstruction methodologies for more accurate predictions of brain injury risks.

Keywords impact reconstruction, brain injuries, brain deformation, dynamic response

I. INTRODUCTION

Head and brain injuries affect over 1.5 million people each year in the United States alone [1]. While these injuries can occur at anytime, many impacts in sport, such as football, rugby and ice hockey, have demonstrated increases in concussive-type injury over a number of years. Recent advances in helmet technologies have virtually removed traumatic brain injuries within high impact sports; however, mild tramatic brain injuries continue to rise. This is likely due to the limitation with regard to helmet technology where management of rotational accelerations is limited. Currently, injuries in these sports tend to occur when two bodies collide at different velocities in different directions [2]. Previous research methodologies used anthropometric headforms, such as a Hybrid III, to reconstruct the injurious event. Current reconstruction methodologies use either a moving headform hitting a static anvil or a moving anvil hitting a static headform [3-6]. Verifying closing velocities, impact angle and location from video, they attached one headform to a vertical drop rig and dropped it on to another static headform. Other methodologies utilised a similar set-up using a linear impactor [7]. This method, however, also involves a static component of the impact condition similar to previous reconstruction methods [3, 4]. While closing velocities are considered for these reconstructions, it is still unknown if this set-up truly represents the actual injurious event when both players are in motion. Recent research has demonstrated that changes in impact conditions can create differences in characteristics of the dynamic response loading curves, which may result in different brain deformation metrics [8, 9]. As a result, current reconstruction techniques using a static component may not fully represent the actual impact event and thus produce different headform responses. Thus, the purpose of this study is to compare the dynamic impact response and brain deformation metrics from two distinct impact recontruction methods: 1) Swinging (Dynamic) headform (DH) and 2) Static headform (SH). It is hypothesized that the dynamic response will differ between the two methodologies and result in higher brain tissue deformation for the swinging headform versus a static headform.

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II. METHODS

Five (5) impact reconstructions were performed where each reconstruction was comprised of one impact from each of the two (2) impact conditions for a total of ten (10) impacts. Each trial used a pendulum to impact a 50th percentile male Hybrid III headform instrumented with a 3-2-2-2 accelerometer array [10]. The accelerometers used were Endevco 7264C-2KTZ-2-300, were sampled at 20 kHz, and filtered using the SAE J211 class 1000 protocol. The data were collected and stored using a TDAS module from DTS (Calabasas, CA). The pendulum impactor had a mass of 10kg with an impactor cap, which consisted of a hemispherical nylon striker (diameter 0.132 +/- 0.01 m; mass 0.677 +/- 0.01 kg). Vinyl nitrile 602 foam, with a thickness of 0.0357 +/- 0.0.1 m, was used between the nylon pieces in order to simulate foam layers of a typical American football helmet.

Test conditions

The DH (Swinging) condition involved the Hybrid III headform (no neck attachment) hanging from a cable and moving at a velocity of approximately 1m/s at a 25˚ angle (relative to horizontal) in the direction of the pendulum impactor (Figure 1). The pendulum impactor velocity was set to 4.1 m/s and impacts occurred on the right side of the headform. The pendulum impactor velocity, headform velocity and impact location were monitored using a High Speed Imaging PCI-512 Fasctam running at 2 kHz for each of the five impacts. The impact location and headform velocity information were essential for comparison with the static headform condition (SH). Each individual Static Headform (SH) impact was reconstructed per the velocity and impact locations from the DH impact.

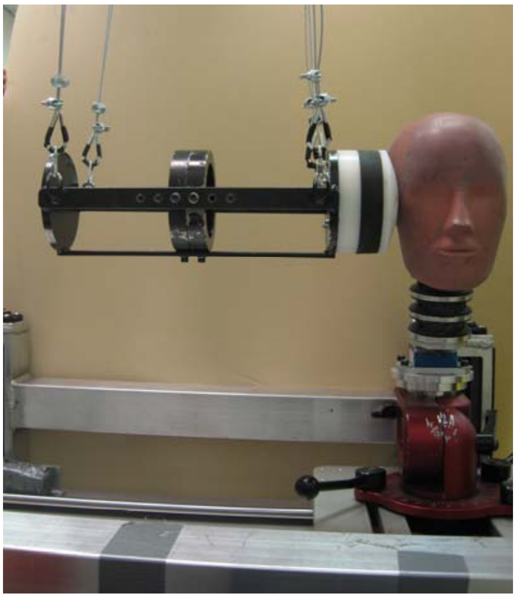
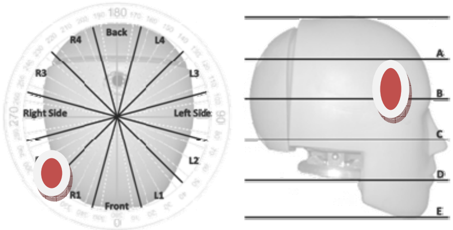
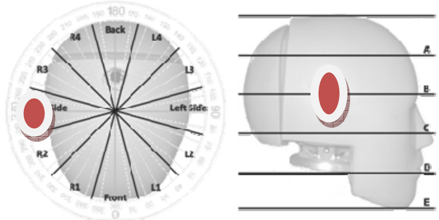
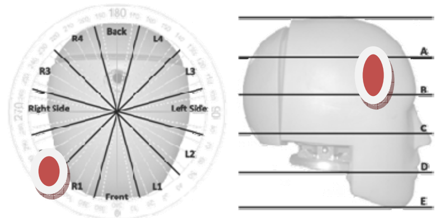
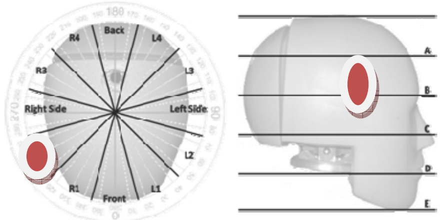
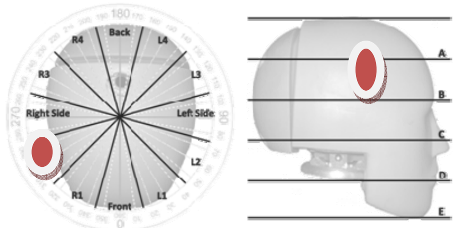


Figure 1. Shows the pendulum used for the impacts.

The SH involved the exact same set-up as the DH condition; however, the headform was not in motion. The headform was angled at 25 degrees (similar to the swinging head condition) and impacted on the right side of the headform by the pendulum at the calculated velocity displayed in Table 1 for each of the five impacts. The calculated velocity for the SH condition included the addition of the horizontal component of velocity calculated from the DH condition.

The locations impacted in this study, while not directly related to an injurious event per se, are high impact areas with potential risk of concussive injury within the sport of football or ice hockey [2, 4]. Three dimensional linear and angular acceleration time-history curves were collected and used as input into a finite element model of the brain, the University College Dublin Brain Trauma Model (UCDBTM), to obtain peak Maximal Principal Strain (pMPS) from nine regions within the cerebrum [8].

Table 1. Paired dynamic and static impact condition and their associated impact location and impact velocities.

Impact condition		Velocity (m/s)		Impact location on Headform
		Pendulum	***Headform***	
A	SH-1	5.4	0.0	
	DH-1	4.1	1.3	
B	SH-2	5.2	0.0	
	DH-2	4.1	1.1	
C	SH-3	5.5	0.0	
	DH-3	4.1	1.4	
D	SH-4	4.6	0.0	
	DH-4	4.1	0.5	
E	SH-5	5.0	0.0	
	DH-5	4.1	0.9	

Finite Element Analysis

Research using impact reconstructions has shown that Maximal Principal Strain (MPS) may be a good predictor for concussion [11-13]. This was the dependent variable used in this study as a measure to quantify brain tissue deformation and its relation to risk of concussive injury per reported thresholds [12, 13]. In order to obtain these values, the resulting linear and angular acceleration loading curves were input at the center of gravity of the University College Dublin Brain Trauma Model [14, 15]. This model was validated according to cadaveric pressure and brain motion data [16, 17] as well as reconstructions of traumatic brain injuries [18].

The material properties of the cerebrum, cerebellum and brain stem for the UCDTBM are described in Tables 2 and 3 [19-24]. The material model used for the brain was linearly viscoelastic, while yielding a large deformation theory [6]. The compressive nature of the brain was considered elastic. The shear modulus of viscoelastic brain tissue was described as [22, 24]:

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

where G_∞ is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor [14]. The brain skull interface was modeled with a low shear modulus that allows it to behave like water. The contact definition at the brain skull interface specified no separation and comprised a friction coefficient of 0.2 [25].

Table 2. Properties of the materials used in the finite element model

Material	Poisson's Ratio	Young's Modulus (Mpa)	Density (kg/m^3)
Falx	0.45	31.5	1140
White Matter	0.49	Hyperelastic	1060
Trabecular Bone	0.24	1000	1300
Grey Matter	0.49	Hyperelastic	1060
Pia	0.45	11.5	1130
CSF	0.5	Water	1000
Bridging veins	-	9.5	-
Dura	0.45	31.5	1130
Tentorium	0.45	31.5	1140

Table 3. Characteristics of the brain tissue

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

The cerebrum within the model was divided into nine specific regions, which have been shown to represent functional areas associated with symptomatology of concussion [26]. MPS values were obtained from the nine regions, and peak MPS (pMPS) was identified and compared for each impact condition for the two reconstruction methodologies.

III. RESULTS

The dynamic response time-history curves (X, Y, Z components) were visibly different for all impact conditions (Figures 3–7). Three-dimensional dynamic response components from each of the impact conditions using the swing and static headform methodology are found in Table 4.

Table 4. Three-dimensional components of the dynamic response for the two impact conditions (Swing vs Static methodologies).

		Peak Linear				Angular			
Condition		**X**	**Y**	**Z**	**Resultant**	**X**	**Y**	**Z**	**Resultant**
A	**Swing**	-39.8	60.0	-40.2	82.4	-7346.0	-2389.1	3699.6	8531.0
	Static	-18.7	64.2	-32.9	67.2	-6586.2	-1792.4	-2802.3	6870.7
B	**Swing**	-17.4	51.7	-23.9	59.5	-4061.8	-2549.1	-3683.0	5176.3
	Static	-9.0	71.5	-9.1	72.0	-6690.8	-1068.4	-2348.6	7120.9
C	**Swing**	-74.2	62.6	-45.1	106.9	-6531.9	-3986.7	3827.2	8500.4
	Static	-55.8	44.0	-36.4	79.7	-4968.6	-3279.6	3136.4	6668.5
D	**Swing**	-69.4	63.7	-51.4	107.3	-7087.8	-3822.1	3992.9	8895.3
	Static	-47.0	66.3	-30.4	86.6	-7793.4	-2279.3	5721.5	9926.2
E	**Swing**	-39.1	-63.1	-58.2	99.6	-10331.8	-2528.4	4800.1	11622.8
	Static	-33.4	66.0	-41.9	84.4	-9106.3	-1879.7	4779.5	10400.4

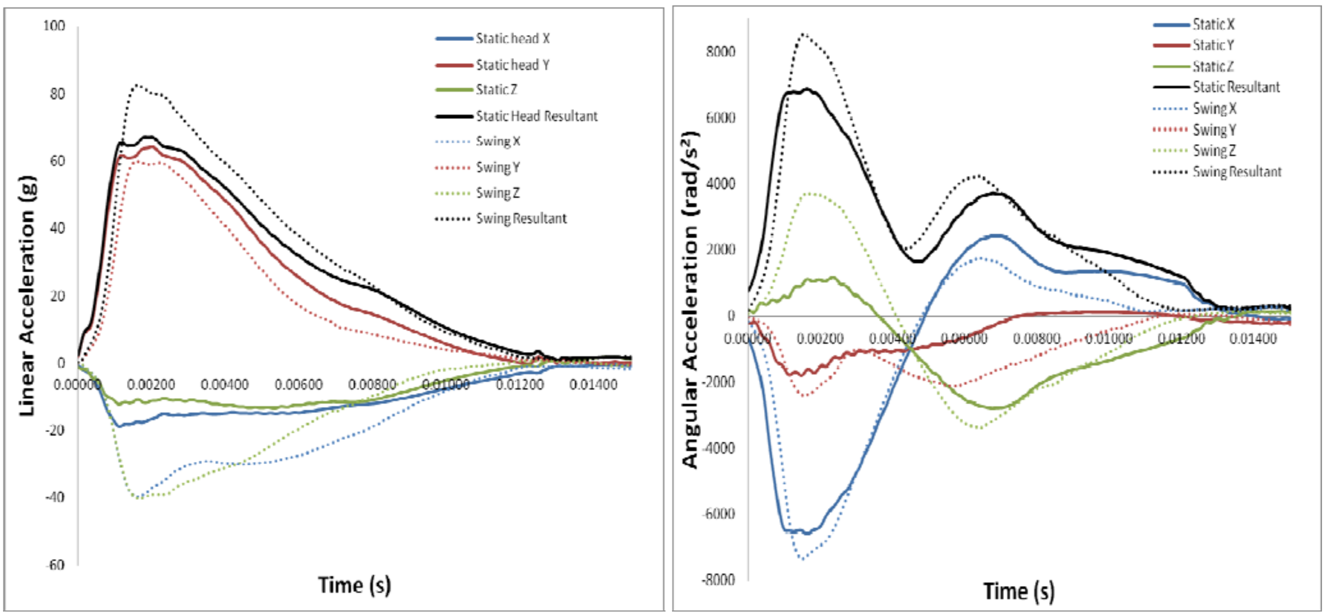


Figure 3. Dynamic response time-history curves for Static headform (Solid lines) and Swinging headform (dotted lines) conditions (impact condition A).

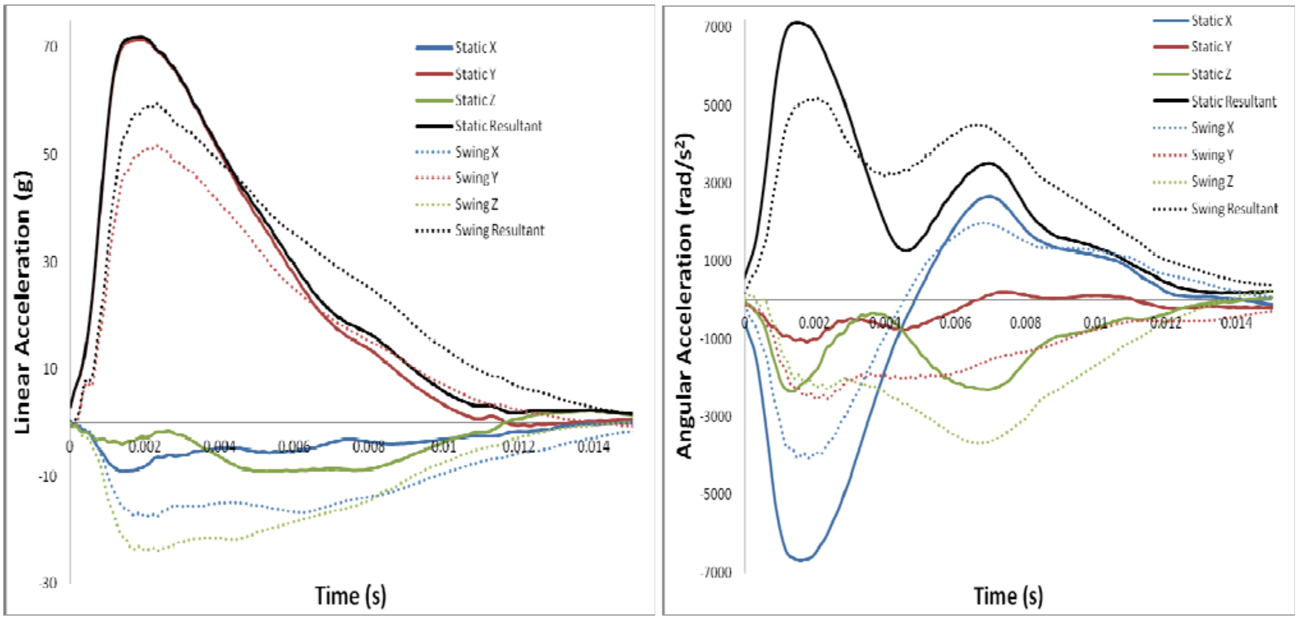


Figure 4. Dynamic response time-history curves for Static headform (Solid lines) and Swinging headform (dotted lines) conditions (impact condition B).

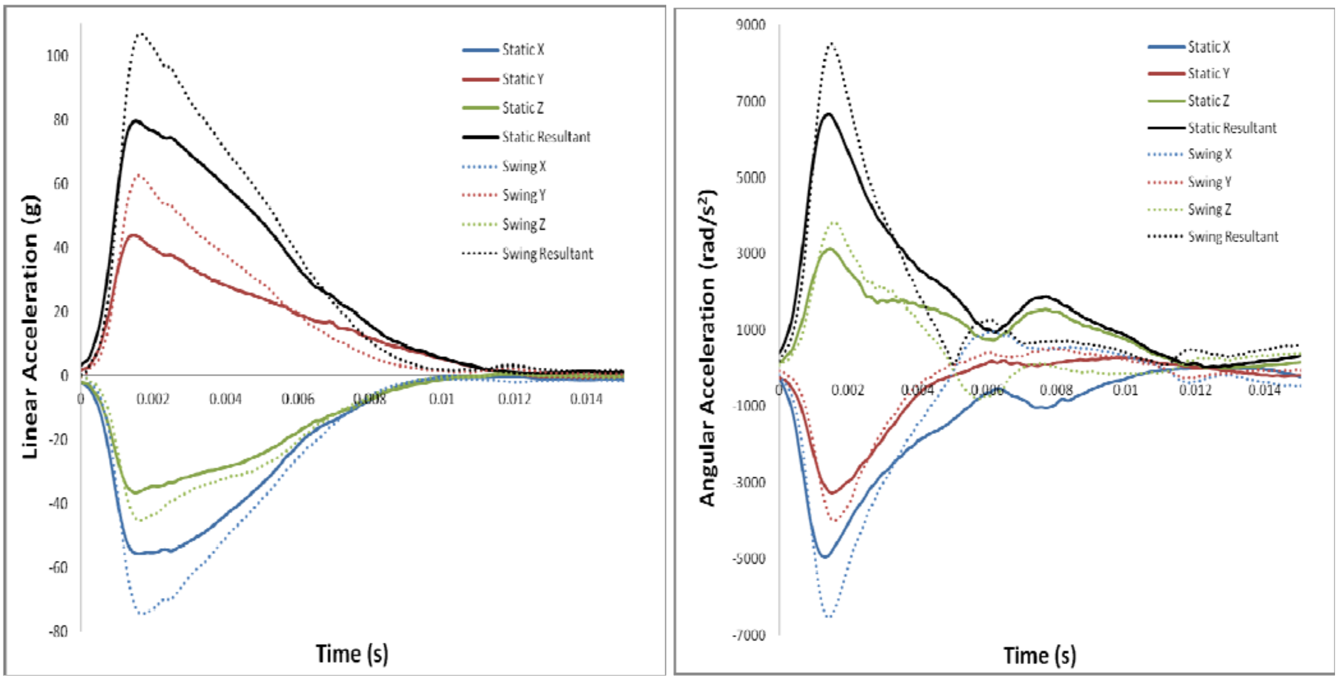


Figure 5. Dynamic response time-history curves for Static headform (Solid lines) and Swinging headform (dotted lines) conditions (impact condition C).

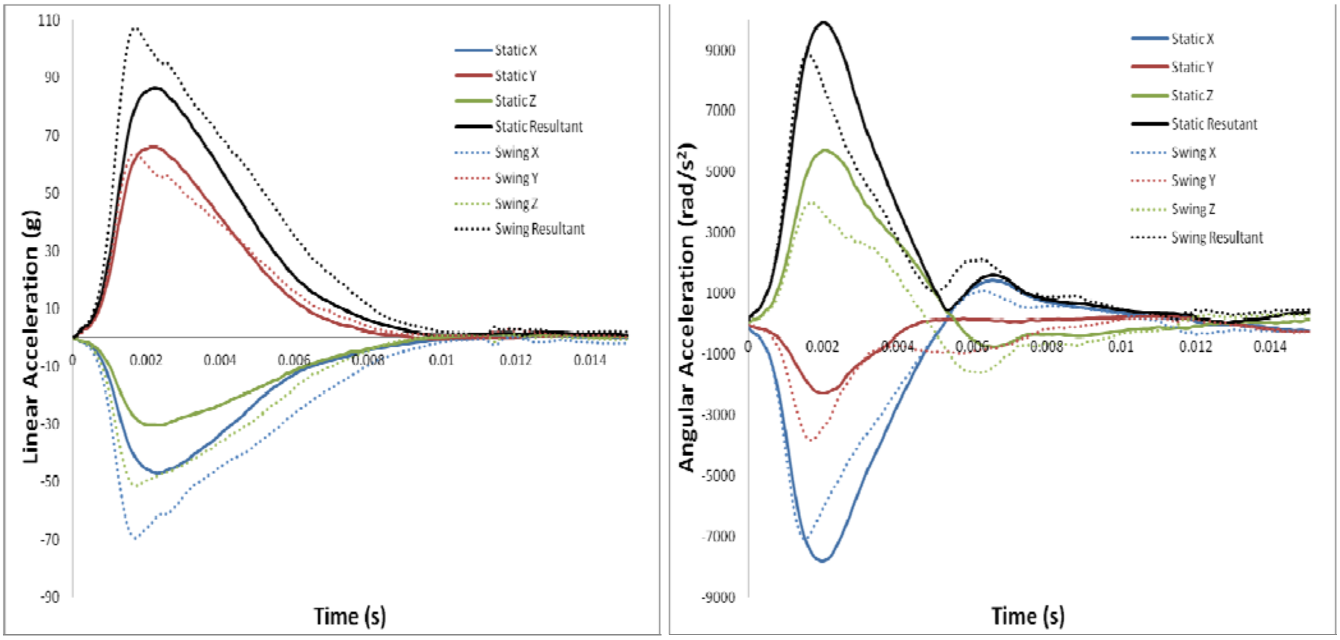


Figure 6. Dynamic response time-history curves for Static headform (Solid lines) and Swinging headform (dotted lines) conditions (impact condition D).

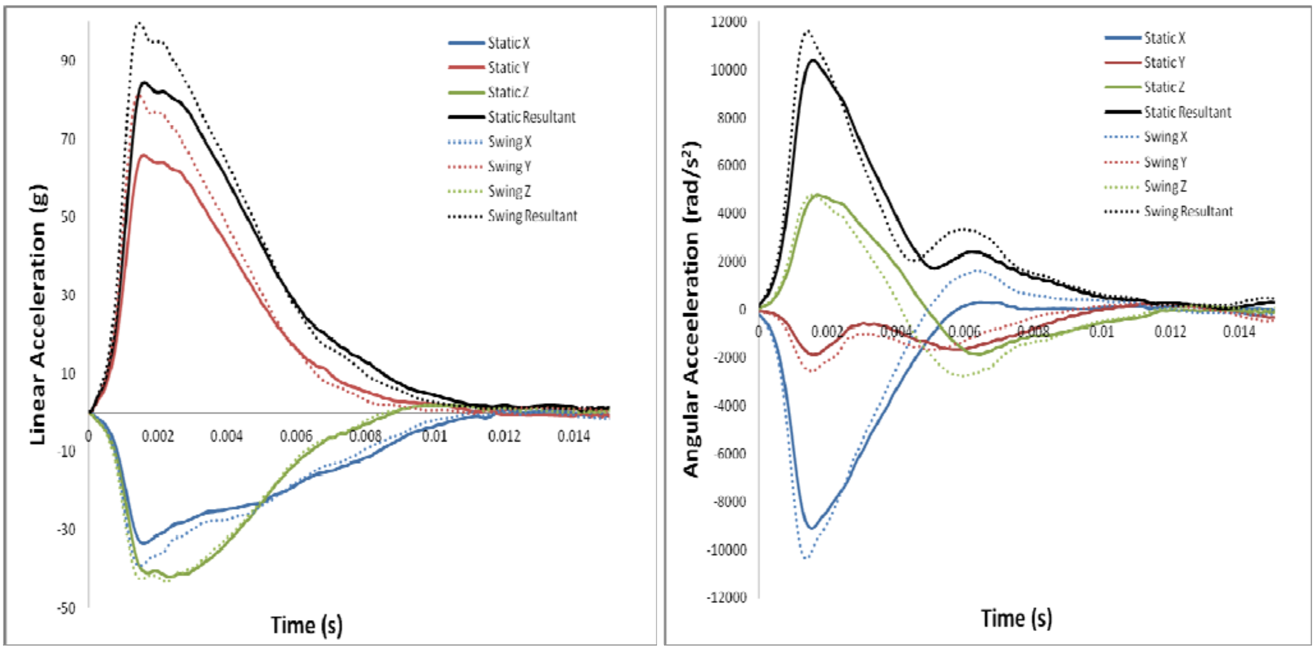


Figure 7. Dynamic response time-history curves for Static headform (Solid lines) and Swinging headform (dotted lines) conditions (impact condition E).

Peak Maximal Principal Strain (MPS) (nine regions of the cerebrum) for each of the five impact conditions comparing the swinging (DH) versus the static headform (SH) methodologies are shown in Table 5.

Table 5. Maximal principal strain (MPS) per region and peak MPS from the two impact conditions.

		MPS per Regions of Cerebrum									
Condition		**1**	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**Peak MPS**
A	**DH-1**	0.18	0.25	0.17	0.25	0.29	0.27	0.26	0.18	0.25	**0.29**
	SH-1	0.30	0.46	0.31	0.42	0.45	0.43	0.30	0.18	0.37	**0.46**
B	**DH-2**	0.23	0.26	0.23	0.31	0.35	0.26	0.27	0.12	0.25	**0.35**
	SH-2	0.20	0.28	0.22	0.28	0.29	0.28	0.27	0.18	0.24	**0.29**
C	**DH-3**	0.20	0.21	0.20	0.23	0.24	0.18	0.31	0.19	0.18	**0.31**
	SH-3	0.25	0.39	0.27	0.36	0.43	0.41	0.29	0.20	0.32	**0.43**
D	**DH-4**	0.20	0.23	0.20	0.22	0.27	0.20	0.30	0.19	0.20	**0.30**
	SH-4	0.21	0.30	0.22	0.30	0.35	0.33	0.22	0.15	0.30	**0.35**
E	**DH-5**	0.20	0.28	0.22	0.27	0.26	0.27	0.29	0.17	0.24	**0.29**
	SH-5	0.14	0.18	0.13	0.21	0.26	0.17	0.17	0.10	0.19	**0.26**

*1 – Prefrontal Cortex; 2 – Dorsolateral Prefrontal Cortex; 3 – Motor Association Cortex; 4 – Primary Motor Cortex; 5 – Primary Somatosensory Cortex; 6 – Sensory Association Area; 7 – Visual Association Area; 8 –Visual Cortex; 9 – Auditory Cortex

IV. DISCUSSION

Accuracy of impact reconstructions is important in order to obtain realistic dynamic impact response that is used for modeling accurate brain deformations which represent risk of injury responses. The purpose of this study was to determine if there are differences between two different methodologies for impact reconstruction. Current methods use a static headform set-up, which is not always a true representation of actual injurious events where two bodies are moving at the time of a collision. In theory, an impact reconstruction should reflect the actual event. Therefore, if two bodies are in motion during the injurious event, the injury reconstruction should also use two moving parts. The hypothesis is that this type of methodology will produce different dynamic response characteristics compared to current static set-up. This may result in differences in brain deformation metrics (MPS) which could result in increases in potential risk of concussive injury. The results of this study demonstrated that there are indeed differences with regards to the dynamic response components between the SH and DH impact methodologies.

The results of this study show that peak linear and peak angular acceleration values are different between the DH and SH reconstruction methodologies for each impact condition. While peak linear and angular accelerations have been associated with risk of concussive type injuries, neither of these two parameters are suitable, on their own, in predicting mTBI. This is evident in this study looking at impact condition A. The SH condition reports peak linear and angular accelerations of 67.2g and 6870.7 rad/s^2 (respectively), while the DH condition reported values of 82.4g and 8531.0 rad/^{2}. While current concussive injury risk would say the DH condition has a higher risk of injury, peak MPS values show the SH (0.46) as being more injurious than the DH condition (0.29). The low correlation between peak linear and angular acceleration as they relate to concussion is likely explained by the unique dynamic response of a specific impact on the brain tissue [27]. Maximal principal strain has been shown to be a good predictor of brain injury [12, 13], since analysis of the complete dynamic response time-history curve is used as input to the finite element model. The results of this study could speak to the different reconstruction methodologies which may be creating differences in the acceleration-time history curves which have been shown to have significant effects on brain deformation metrics values resulting from FE modeling [9]. The time-history curves for both linear and angular accelerations show that though the three-dimension component curves for DH and SH conditions have similar shapes, they differ in amplitude. Visual differences with regard to the loading curves can be seen between the two reconstruction methods. More specifically, trend differences related to acceleration peaks, impact duration, slope of the curve and integral. Thus these differences within the curves, (i.e. duration, time to peak, integral) could be the cause for the higher MPS values from the swinging headform set-up. One dynamic response variable that was consistenly higher for all impact conditions reporting the higher pMPS was the integral for the resultant and Z component of the angular acceleration. More research should focus on different characteristics of the the dynamic response curves to better understand their relation to brain deformation metrics.

The peak MPS values within the cerebrum varied with regard to region. These results may indicate that magnitude of MPS and location of the peak MPS within the cerebrum is sensitive to the methodological set-up for the impact reconstruction. Where it has been previously shown that a particular impact reconstruction method may have significant effects on the location of the peak brain deformation within the cerebrum [8], this study differs from this in that the same impacting method (with same mass) was used in both conditions. It is very likely that the DH impact methodology is creating a different impact mechanism than that of the SH condition, thus emphasising the importance of accurate reconstruction methods when conducting impact reconstructions. If reconstructing a concussive injury in ice hockey where two players moving at given velocities come together, the reconstruction methodology should be developed using two moving components.

The research presented should be considered according to limitations inherent in its methodologies. The Hybrid III headform is commonly used as a physical model in impact reconstruction due to its reliability. The headform being constructed of steel evidently does not provide a biofidelic response to impacts. Also, the Hybrid III is primarily used for impacts in the antero-posterior direction and thus impacts to other regions of the headform may create unknown error. This study did not use a neckform for either impact methodology, and thus should not be considered a true representation of an injurious event. The UCDBTM is one of few partially validated models available for this type of research. However, the finite element model makes assumptions surrounding the characteristics of the brain tissue and the interactions between different parts of the brain and skull. As a result, the comparisons made with the UCDBTM are meant to be representative of how the brain may deform under the loading scenarios and may not represent the exact motion of the brain resulting from impacts. Furthermore, brain deformation metric analysis was limited to the cerebrum since the brain stem has not yet been validated for finite element modelling.

V. CONCLUSIONS

Current reconstruction methodologies typically include one dynamic and static component. In sports, such as football and hockey, impacts usually involve two bodies moving at different velocities towards each other. The purpose of this study was to compare two impact reconstruction methodologies, a static headform and a dynamic headform, by means of dynamic impact response and brain deformation metrics. The results of this study showed that there are differences between an impactor hitting a static headform versus a headform in motion. While peak linear and angular accelerations did not conclusively distinguish between the two impacting methodologies, the brain deformation metrics showed important differences. These differences were likely due to specific characteristics within each method's loading curves. This study demonstrates the importance of ensuring accurate impact reconstructions of injurious events in order to accurately assess brain injury risk. This understanding could aid in the development of better impact reconstructions specific to each injury mechanisms.

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