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Measuring the Influence of Neck Compliance on Brain Tissue Strain

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ABSTRACT – The aim of this study was to determine if a change in neck compliance would influence the risk of sustaining a TBI by quantifying its effects on maximum principal strain in the brain. This was done by impacting a Hybrid III headform with a 16.6 kg impactor arm at $5.5 \text{ m}\cdot\text{s}^{-1}$. Three different Hybrid III neckforms were used: 1) one 50th percentile male neck – standard neck; 2) one 50th percentile male neck plus 30 per cent compliance – soft neck; 3) one 50th percentile male neck minus 30 per cent compliance – stiff neck. The kinematic data obtained was then used to drive a finite element model developed by University College Dublin. The results showed that neck compliance did not have a significant effect on maximum principle strains in the white matter, grey matter and brain stem. The stiff neck, however, generated higher strain levels in the cerebellum than the medium and soft neck conditions. A comparison to proposed mild TBI thresholds revealed that collisions happening at $5 \text{ m}\cdot\text{s}^{-1}$ can disrupt the axons in the white and grey matter.

KEYWORDS – Impact biomechanics; brain injury; finite element; maximal principal strain.

INTRODUCTION

Athletes participating in contact sports are subject to violent collisions. Such events may cause irreparable trauma to the brain if head contact occurs. Although traumatic brain injuries (TBI) represent a small portion of all reported injuries in sports (**Powell & Barber-Foss, 1999; Covassin et al., 2003**), their cost to American society is approximately \$56.3 billion annually in direct and indirect cost (**Langlois et al., 2006**). Mild TBI, commonly referred to as concussions, are the most frequent brain injury and may occur more often than previously thought (**Delaney et al., 2002; Forero Rueda et al., 2010**).

The most common mechanism of mild TBI is a direct impact to the head during a collision with an opponent (**Delaney et al., 2006; Gerberich et al., 1983; Gerberich et al., 1987; McIntosh et al., 2000**). For this reason, skilled athletes have developed strategies to mitigate the severity of a collision. The most effective, and most instinctive, reaction to an incoming impact is to dodge the blow. By angling the head away from the point of impact, it is possible to reduce the energy transfer occurring upon contact (**Reid et al., 1975**). When evading the collision is not possible, athletes will then brace themselves for the incoming hit (**Reid et al., 1975**).

Rousseau and Hoshizaki (**2009**) evaluated both techniques by impacting a Hybrid III headform with a 16.6 kg pneumatic linear impactor. “Avoiding” was simulated by translating the headform laterally by increments of 3.875 cm (1/8th of the width of the headform). “Bracing” was simulated by manipulating the compliance of the Hybrid III neck. Impact deflection was shown to be effective at reducing peak linear acceleration and peak angular acceleration. The results supported the theory that a small lateral shift could reduce the energy transfer between the impactor and the headform sufficiently to significantly reduce the severity of the impact. The results were not so clear for neck compliance, where a decrease in neck compliance – mimicking neck muscle contraction – caused a significant decrease in angular acceleration but a significant increase in linear acceleration. When compared to published risk thresholds (**Zhang et al., 2004**), it was determined that the increase in linear acceleration was minimal compared to the decrease in angular acceleration meaning that strategies aimed at decreasing neck compliance could reduce the risk of sustaining a mild TBI.

Linear and angular accelerations have historically been used as TBI predictors. Early work by Gurdjian et al. (**1943; 1944; 1953**) associated accelerations present in head injuries to skull deformation and intracranial pressure. It was later reported that angular acceleration had a minimal contribution in concussive events; thus making linear acceleration a more important determinant (**Gurdjian et al., 1955**). This was supported by Ommaya et al. (**1966; 1971**) who demonstrated that higher levels of angular acceleration were needed to cause a concussion through whiplash than linear acceleration through a direct impact, making the latter a more likely cause for mild TBI. On the other hand, Holbourn (**1943**) speculated using the brain’s physical properties that angular acceleration was a better injury predictor for mild TBI. The brain’s high bulk modulus and low shear modulus make shear-stress a more likely injury mechanism. Considering that shear-stress and angular acceleration were correlated, it was concluded that angular acceleration was responsible for concussions and hemorrhages. This was supported by Gennarelli et al. (**1971; 1972; 1982; 1982**) who conducted a series of studies using monkeys and physical model. It was determined that angular acceleration played a predominant role in the mechanism of concussion, diffuse axonal injury and subdural hematoma when no direct impact to the head occurred.

Unfortunately, acceleration does not accurately depict brain deformation as both are outcomes of the force transfer occurring during a direct impact (**Newman, 1980**). Being composed of different types of tissue, the brain deforms in a heterogeneous manner creating shear stress and strains, which in turn cause lesions in the white matter (**Strich, 1961**). Rapid developments in finite element (FE) models have enabled the study of mechanical deformations occurring in the brain (**Zhou et al., 1995; Ruan et al., 1997; Zhang et al., 2001; Kleiven and Hardy, 2002; Horgan and Gilchrist, 2003**). Despite inherent limitations, FE models remain a valuable tool and offer information which can improve brain trauma prediction. Thus, the objective of this study was to determine if a change in neck compliance would influence the risk of sustaining a TBI by quantifying its effects on maximum principal strain in the brain.

Table 2 – Material properties for the human head finite element (brain).

Material	Density ($\text{kg}\cdot\text{m}^{-3}$)	Shear modulus (kPa)		Decay constant (s^{-1})	Bulk modulus (GPa)
		G_0	G_∞		
Grey matter	1 000	10	2	80	2.19
White matter	2 000	12.5	2.5	80	2.19
Brainstem	1 300	22.5	4.5	80	2.19
Cerebellum	1 130	10	2	80	2.19

METHODS

The influence of neck compliance was quantified using a FE model representing the human head complex. The kinematic data used to drive the model was obtained by impacting a 50th percentile Hybrid III headform with a 16.6 kg mass projected at a velocity of $5 \text{ m}\cdot\text{s}^{-1}$. The headform was attached to three different Hybrid III necks to mimic variations in compliance. Maximum principle strains were evaluated in the white matter, grey matter, brain stem and cerebellum. The analysis was done using ABAQUS 6.8-2 (SIMULIA, Providence RI, USA.).

Figure 1 – University College Dublin Brain Trauma Model, full head – left, and brain – right.

Finite Element Model

The head model (**Figure 1**) used in this study was developed at University College Dublin (**Horgan and Gilchrist, 2003; Horgan and Gilchrist, 2004**). The model includes the cerebrum (grey matter, white matter and ventricles), cerebellum and brainstem, intracranial membranes (falx and tentorium), pia, cerebrospinal fluid layer (CSF), dura, a varying thickness three-layered skull (cortical and trabecular bone layers), scalp and the facial bone. 7 318

hexahedral elements represent the brain and 2874 hexahedral elements represent the CSF layer with one element through the thickness. The model was validated against the pressure response of Nahum's (1977) cadaveric impact test and the displacement response of Hardy's (2001) high-speed x-ray cadaver impact test.

Material properties – Various constitutive models have been used to describe the mechanical behavior of brain tissue, including using an integral model in combination with hyperelasticity. In this present analysis, the neural tissue was simply characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus, while the compressive behavior was considered as elastic. The CSF layer was modeled using solid elements with a low shear modulus and a high bulk modulus made of the hybrid elements in ABAQUS. The CSF layer had a 1.3 mm depth. The remaining parts of the model were taken from the literature (**Horgan and Gilchrist, 2003; Horgan and Gilchrist, 2004**). All material properties can be found in **Tables 1 and 2**.

Impact Reconstruction

A linear impactor system consisting of a weighted, pneumatically driven impactor arm (**Figure 2**) and a Hybrid III head- and neckform attached to a sliding table (**Figure 3**) were used to produce the kinematic data used to drive the FE model. Impacts were to the forehead, $30 \pm 0.5 \text{ mm}$ above the intersection of the longitudinal plane and the reference plane.

Table 1 – Material properties for the human head finite element (scalp, skull and meninges).

Material	Density ($\text{kg}\cdot\text{m}^{-3}$)	Young's modulus (MPa)	Poisson's ratio
Scalp	1 000	16.7	0.42
Cortical bone	2 000	15 000	0.22
Trabecular bone	1 300	1 000	0.24
Dura	1 130	31.5	0.45
Pia	1 130	11.5	0.45
Falx and tentorium	1 130	31.5	0.45
Facial bone	2 100	5 000	0.23

Figure 2 – Linear impactor arm and steel frame.

To reproduce different neck compliances, three different Hybrid III neckforms were used. One 50th percentile male neck was identified as the standard compliance, while the remaining two were manufactured to correspond to softer (+ 30 per cent compliance) and stiffer (-30 per cent compliance) necks (Denton ATD, Milan OH, USA). The desired compliance was obtained by modifying the properties of the rubber discs. All three neckforms were calibrated following the standard protocol established by the National Highway Traffic Administration (FMVSS, part 572E).

Figure 3 – 50th percentile male Hybrid III head- and neckform mounted to a sliding table.

RESULTS

Maximum principal strain in the white matter, grey matter, brainstem and cerebellum can be found in **Table 3**. The results showed that neck compliance did not have a significant effect on maximum principle strains in the white matter ($F_{(2, 24)} = 0.362$, $p > 0.05$), grey matter ($F_{(2, 24)} = 2.922$, $p > 0.05$) and brain stem ($F_{(2, 24)} = 1.096$, $p > 0.05$). It did, however, have a significant effect in the cerebellum ($F_{(2, 24)} = 53.138$, $p < 0.001$). Further analysis using Tukey's

post hoc statistical method, revealed that the stiff neck condition generated higher strain levels in the cerebellum than the medium and soft neck conditions ($p < 0.001$).

DISCUSSION

A previous report by Rousseau and Hoshizaki (2009) showed that the three neck compliances tested influenced peak linear and angular accelerations. Although significant, the influence on linear acceleration was minimal. It was also found that a decrease in neck compliance was accompanied by an increase in linear acceleration. The influence was much larger on angular acceleration with a variation of approximately 20 per cent. In this case, a decrease in neck compliance was accompanied by a decrease in angular acceleration. The results led to the interpretation that athletes involved in unexpected collisions might have a greater risk of sustaining a head injury.

Angular acceleration has been linked to brain tissue strain (Viano et al., 2005; Kleiven, 2007); therefore, it was expected that strain would be higher when the neck was the most compliant. This was not the case as maximal principal strain values were not significantly different in the white matter, grey matter and brain stem.

The stiffer neck produced slightly higher maximum principal strains in the cerebellum; however, the values were below even the most conservative proposed injury thresholds. A study performed on giant squid axons found that a 12 per cent elongation suppressed nerve activity for three minutes, a 20 per cent elongation would prevent the axon from fully recovering and a 25 per cent elongation resulted in structural failure. Thus, ten per cent was determined as a conservative threshold for the onset of mild TBI (Galbraith et al., 1993; Thibault, 1993). A slightly higher threshold of 0.21 was established by Bain and Meaney (2000) following a study conducted on a guinea pig. A recent analysis made by Kleiven (2007) using 58 National Football League cases obtained a 50 per cent risk of mild TBI for a level of 0.21 and 0.26 in the corpus callosum and grey matter,

Table 3 – Maximum principal strain peaks (standard deviation) within the brain tissue for a front impact of a Hybrid III headform at 5 m·s⁻¹.

	Maximum principal strain			
	White matter	Grey matter	Brain Stem	Cerebellum
Soft neck	0.119 (0.019)	0.176 (0.007)	0.085 (0.009)	0.037 (0.001)
Median neck	0.114 (0.020)	0.176 (0.011)	0.081 (0.006)	0.036 (0.003)
Stiff neck	0.122 (0.013)	0.168 (0.003)	0.080 (0.003)	0.050 (0.004)

respectively.

Interestingly, results showed that regardless of an athlete's neck compliance, collisions happening at $5 \text{ m}\cdot\text{s}^{-1}$ can disrupt the axons in the white and grey matter. This means that even impacts at a lower velocity can be detrimental to the athletes' health.

CONCLUSION

The objective of this study was to determine the influence of neck compliance on maximum principal strain in the brain. Results showed no significant differences in the white matter, grey matter and brain stem. Maximal principal strains in the white and grey matter were above a mild TBI threshold which was established by Thibault (1993), while the strains in the brainstem remained below. The least compliant neck caused significantly higher strains in the cerebellum; however the values remained below threshold.

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