The Effects of Reconstructed Head Impact Event Parameters on Risk of Sport Related Concussions

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Abstract

Falls and collisions are the most common types of events leading to sports-related concussions where impacts to the head play an important role on the onset of traumatic brain injury. Each event can be described by impact parameters that define the loading conditions on the head and brain and are necessary for accurate accident reconstruction employing physical impact tests, anthropometric headforms, and finite element (FE) modelling. It was the purpose of this research to describe the effects and interactions of impact velocity, compliance, mass and impact location on head acceleration and brain tissue strain measures associated with risk of concussions in sports.

Impact parameters were varied to capture responses from no-injury up to concussive levels. Study one examined the effect of impact parameters on fall events simulated using a monorail drop tower. Impact mass was varied using three different headforms representing child, adolescent, and adult sizes measuring peak linear and angular acceleration and maximum principal strain. Regression analysis revealed that impact compliance was the most influential on peak linear and angular acceleration measures, meanwhile FE strain was most affected by changes in impact velocity. Smaller headforms tend to produce higher acceleration and strain values, supporting the need for age and size related mechanical definitions of risk.

Study two examined the effect of impact parameters for collision events simulated using a multi-mass pendulum to represent common striking masses in sport measuring peak linear and angular acceleration and strain. Study three provided further insight into collision impacts by evaluating the distribution of peak strains in different brain lobes and the volume of the brain experiencing strains passed a critical level. Results show that compliance was similarly the most influential on peak head acceleration whereas peak strain and volume were most affected by impact velocity. Mass-velocity interactions had effects where a 9 kg mass had greater response than 15 kg, but similar to 21 kg. The temporal lobe consistently contained the highest strains with the rear boss non-centric impact location producing the largest values. Interacting impact parameters illustrate the challenges with predicting associated risk of concussion from head collisions in sport and supports the need to identify effective performance ranges of protective materials.

Preface

This dissertation is organized into three parts with multiple chapters in each section:

I. *Opening*, which introduces the background, research statement, literature review, and research design,

II. *Effect of Impact Parameters*, which describes the role of mechanical impact parameters on head acceleration and brain tissue strain for falls and collisions in sport,

III. *Closing*, which includes the points of discussion, limitations, and concluding remarks.

The dissertation is comprised of three original research articles, which I was lead author. I was fully involved with: 1) the conception and design of the studies; 2) data acquisition, analysis and interpretation; 3) drafting, revising, and submission of the articles for peer review in scholarly journals.

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I Opening

1 Introduction

Statement | Rationale | Objectives

1.0 Statement

In Canada, it has been estimated that over 11 million children, youth, and adults participate regularly in sport (Statistics Canada, 2010; Cusimano et al., 2013; Canadian Heritage, 2013; Canadian Fitness & Lifestyle Research Institute, 2011). While regular physical activity helps prevent chronic health diseases including diabetes, obesity, and depression, sport participation presents a significant risk of concussion (Bazarian et al., 2005; Canadian Institute for Health Information, 2006; Warburton et al., 2006). Concussion is a type of brain injury 'induced by biomechanical forces' resulting in a 'rapid onset of short-lived impairment of neurological function' (McCrory et al., 2017). Impacts to the head are the primary cause of concussion in sport, with the top 5 symptoms reported by athletes are headache, problems concentrating, feeling slowed down, dizziness, and nausea (Lovell et al., 2006). The majority of these problems resolve within the first 10 days post injury, however 10-15% suffer persistent symptoms (Makdissi et al., 2013), decreasing the quality of life for many Canadians from an inability to resume a normal lifestyle and a loss of productivity in school and earning wages (Cantu, 1992; Rimel et al., 1982; Rimel et al., 1981). In addition, repeat concussive injuries and head impacts have been identified as a primary risk factor for chronic traumatic encephalopathy, a long-term neurodegenerative disease associated with increased irritability, suicidality, poor memory, movement and speech impairments later on in an athlete's life (Gavett et al., 2011; McKee et al., 2009).

In 2015, nearly 70 000 emergency room visits to Canadian hospitals were from sport participation resulting in $187 million in direct and indirect costs associated with injury (Parachute, 2015). These figures largely underestimate the true rate and economic burden of concussion as they only reflect registered patients seeking medical care, and a number of cases go un-reported due the seemingly minor effects of concussion in comparison to more fatal injuries (Roozenbeek et al., 2013). In the U.S, it has been estimated that up to 1.6 to 3.8 million concussions occur each year from sport and recreation (Langlois et al., 2006). The Canadian government has acknowledged that concussions are an important public health concern and with efforts to encourage more Canadians to reap the benefits of physical activity, improving the safety in sport is a current priority. The provincial, federal, and territorial ministers have aligned efforts in the awareness, surveillance, detection, management, and prevention of concussions in Canada (Canadian Intergovernmental Conference Secretariat, 2016).

1.1 Research Question

Ice hockey, football, and soccer are among the top sports with the highest rates of concussion for children, high school, and collegiate athletes (Cusimano et al., 2013; Daneshvar et al., 2011; Tommasone & Valovich McLeod, 2006). The most common events leading to injury are player-to-player collisions involving an opponent's helmet, shoulder, or other body parts accounting for 60-90% of injuries, and falls onto playing surfaces, i.e., ice, grass, turf, and boards, as the next frequent cause with 7-35% (Cusimano et al., 2013; Daneshvar et al., 2011; Delaney et al., 2006; Gessel et al., 2007; Hutchison et al., 2013). In a collision, the head contacts a moveable mass where both bodies are free to move post-impact. In contrast to a fall, the head contacts a rigid, non-yielding surface resulting in a near-complete transfer of energy to the head. Impact parameters that modulate the nature of this energy transfer and the mechanical effects on the head and brain are impact velocity, striking mass, surface compliance (stiffness), and impact location (Denny-Brown & Russell, 1941; Gurdjian et al., 1964; Gurdjian et al., 1963; Holbourn, 1943; Ommaya, 1966). The relationship between the external forces and head and brain injury response are linked, where changing the nature of the input parameters will have an effect on the output (Gennarelli et al., 1987; Gurdjian et al., 1964; Gurdjian & Webster, 1947; Meaney & Smith, 2011). Ultimately, the goal of injury biomechanics is to establish a cause-and-effect relationship between impact parameters (velocity, mass, location and compliance) and injury outcomes in order to better predict risk (Goldsmith, 2001; Viano et al., 1989). Effects of head impact parameters have been described in the literature, where increases in impact velocity, mass, and low compliance conditions resulted in increased peak linear and angular accelerations (Gurdjian et al., 1966; Karton et al., 2014). Interactions between these parameters have been noted with bicycle helmet foam samples where the rate of bottoming out (point of increased energy transfer to the head) was a function of the impact mass, striking velocity, and foam density (Gimbel & Hoshizaki, 2008). It is hypothesized that interactions between impact parameters exist within sport creating conditions associated with increased risk of concussion. It is also hypothesized that these interactions are unique to the impact event as head impacts from falls and collisions transfer energy differently. Therefore, the purpose of this research was to address the question, *what are the main effects and interactions of mechanical impact parameters on modulating head acceleration and brain tissue strain responses associated with concussion risk?*

1.2 Background & Rationale

1.2.1 Head Injury Research Paradigm

A paradigm for head injury research as proposed by Ommaya et al. (1994) based on the group's previous experimental work with animal and tissue levels models served as a guide for this dissertation. Figure 1-1 is a depiction of the conceptual framework reflecting the relationship between mechanical loads, head motions, intracranial strains, and injury. The specific application of this paradigm adopted for studying sports related concussions is summarized in Figure 1-2. Mechanical loads from sport impacts are defined by event type: falls and collisions with loading durations up to 50 ms (Hodgson & Thomas, 1972; Hoshizaki et al., 2016; Stalhammar, 1986). Falls causing sports related concussions are typically defined by the mass of the head and/or helmet contacting rigid surfaces, such as ice or grass and can occur up to impact velocities of 6 m/s (Hoshizaki et al., 2014). In contrast, collisions causing concussions are defined by the head coming into contact with helmets, shoulders, and elbow pads, comprised of striking masses ranging from 3-16 kg traveling at velocities from 4-11 m/s (Hoshizaki et al., 2014). These events are described by specific levels of velocity, mass, compliance, and location, which influence the relative amounts of head linear and angular accelerations and patterns of focal and diffuse strains on the brain associated with cerebral concussion (Denny-Brown & Russell, 1941; Gennarelli et al., 1982).

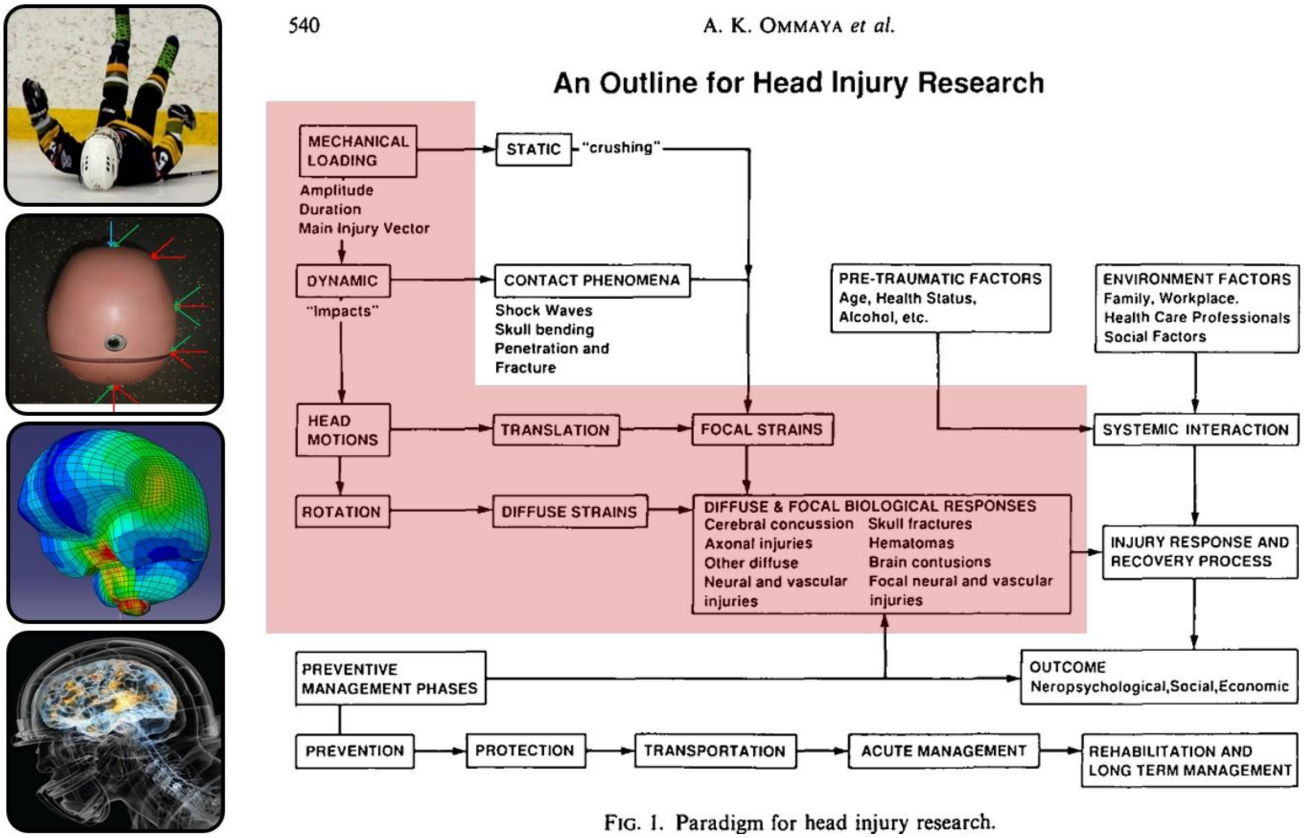

Figure 1-1. Head injury paradigm proposed by Ommaya et al. (1994).

Previous experimental investigations on the effects of impact parameters have reported increases for mass (Karton et al., 2014) and velocity (Gurdjian et al., 1964), or a decrease in compliance (stiffer surface) produced higher peak head accelerations and strains as the trends of these conditions transfer greater amounts of energy (Gurdjian et al., 1964). Impacts to the side of the head in experimental animal research caused prolonged loss of consciousness and increased severity of axonal damage, highlighting the role of restricting structures, i.e., the tentorium on intracranial deformations (Hodgson et al., 1983). These reported effects were studied under limited conditions, manipulating one or two parameters under a narrow range. The interaction of parameters have been demonstrated to result in increased responses under a specific range of conditions. Gimbel and Hoshizaki (2008) manipulated the mass of a magnesium headform, dropped at 1 to 5 m/s onto samples of bicycle helmet foams of different stiffness (compliance). The point at which the foam was no longer effective at attenuating impacts, causing a spike in peak linear acceleration response (inflection point), was a function of the combined effects of mass and velocity (Gimbel & Hoshizaki, 2008).

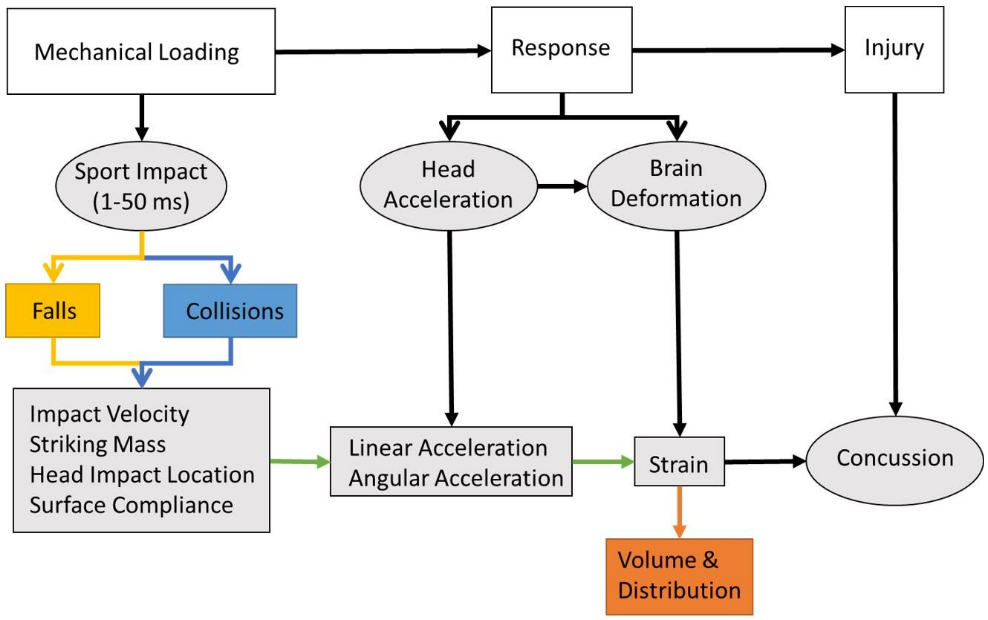

Figure 1-2. Conceptual framework guiding this dissertation.

While general trends have been established for main effects of impact parameters, these data span different domains and methodologies that used cadavers, animal and physical models, often measuring a single dependent or response variable, which limits the full characterization and understanding of interaction effects. Under the diverse loading conditions in sport, interacting parameters likely create environments associated with increased risk, which has limited the complete understanding of this phenomenon and predicting the relationship between mechanical loads (input) and injury risk and outcome. Delineating these phenomena will lead to improvements in sport safety by identifying the specific impact conditions that lead to increased risk. The implications of this data will inform researchers, policy makers, and helmet manufacturers on the necessary energy requirements needed to develop adequate intervention strategies through protective devices, coaching and game changes to reduce the incidence of concussion.

1.3 Objectives

The objectives of this research was to identify interactions between impact parameters for fall and collision events in sport. Levels of impact velocity, surface compliance, and head impact locations covering motions occurring in different planes, were varied to capture sub-concussive to concussive level responses characteristic of sport loading conditions (Hoshizaki et al., 2016; Pellman et al., 2003; Zhang et al., 2004). Levels of impact parameters were kept consistent for both events except mass, where headform size was used to reflect levels of mass for fall impacts and a range of striking masses were used to reflect typical collision impacts (Rousseau & Hoshizaki, 2015; Walilko et al., 2005). The specific aims of this dissertation include:

1. Describe the main effects and interactions of impact parameters (mass, velocity, compliance, and location) on peak head acceleration and peak strains for falls in sport
2. Describe the main effects and interactions of impact parameters (mass, velocity, compliance, and location) on peak head acceleration and peak strains for sport collisions
3. Determine the influence of interacting parameters on strain in four brain lobes and the volume of brain experiencing 0.10, 0.15, 0.20, and 0.25 strains for sport collisions

2 Biomechanics of Head Injuries in Sport

Literature Review

2.1 Types of Head Injuries in Sport

Head injuries in sport can affect the external structures such as the skull and scalp and the internal brain and its coverings (Goldsmith, 2001; Gurdjian et al., 1968). Scalp and skull injuries include bruises to the scalp from leakages of blood from damaged vessels into tissue beneath the skin. Abrasions are scrapes of the upper layers of skin from a sliding object or surface and lacerations are a cutting or opening of the skin from a sharp object. Injuries to the skull are fractures and the different types are described by the nature of the broken bone (Goldsmith, 2001). Linear skull fractures are usually from a thin line break from blunt impact and occurs at points remote from the impact site. Depressed fractures are when the bone is displaced inwards onto the brain and is typically caused by an object perforating the skull (Yoganandan et al., 1995). The remainder of this chapter will review the spectrum of possible brain injuries in sport with a focus on concussion.

2.1.1 Traumatic Brain Injuries

In sport, there are two types of traumatic brain injury (TBI): acute and chronic. Acute TBI are those with outcomes that are evident immediately after the injury and are further categorized into focal and diffuse brain injuries depending on the nature of tissue damage. Chronic TBIs are those where there is a period of time between the injury event and neurological effects (Jordan, 2013). Examples of chronic TBIs and neurodegenerative disorders associated with contact sport are dementia, motor neuron diseases, and chronic traumatic encephalopathy (CTE). These diseases have been identified in athletes participating in professional levels of ice hockey, American football, and boxing (Bailes et al., 2013; Blennow et al., 2012; McKee et al., 2009). These types of brain injuries are an active area of research with scientists attempting to establish the relationship between frequency and magnitude of brain trauma and the onset of neurodegenerative effects (McKee et al., 2009).

Focal Brain Injuries

Focal brain injuries are those that are localized to a single area in the brain and include lesions to blood vessels and brain tissue that can lead to death (Blennow et al., 2012). Epidural hematomas are from ruptured blood vessels creating clots between the skull and dura, usually from skull deformation. Subdural hematomas result from ruptured blood vessels situated between the dura and brain and can result from relative brain-skull motion from angular acceleration (Depreitere et al., 2006). Intercerebral hematomas result from the tearing of blood vessels within the brain, and lacerations are from a tearing of the brain tissues. Contusions are bruises on the surface of the brain from the breakage of small vessels with the seeping of blood into adjacent areas (Löwenhielm, 1975).

Diffuse Brain Injuries

Diffuse brain injuries (DBI) are a group of injuries that encompass the spectrum of axonal traumas from functional injuries with no structural axonal damage to observable axonal damage. The severity of DBI will depend on the degree of axonal traumas and the specific regions of damage in the brain with injury severity increasing from mild concussion, concussion with loss of consciousness, diffuse axonal injury (DAI), and death (Gennarelli et al., 1998). Early evidence of axonal damage in human subjects have been reported by Strich (1969) who observed diffuse damage in the brains of patients who had suffered minor head injuries, and later died of other causes. Similarly, Jane et al. (1985) observed the presence of axonal damage in animals suffering a loss of consciousness but did not display any long-term deficits from the traumatic loading. These early studies sparked on-going research efforts in the understanding of biomechanics of concussion using accident reconstruction (Wright et al., 2013), animal testing (Margulies & Thibault, 1992), and computer modelling (Giordano & Kleiven, 2014), as well as tissue and cell culture tests linking levels of head motion to localized axonal stretch that disrupt cell membranes and function (Barkhoudarian et al., 2011; LaPlaca et al., 2007).

In 1974, Ommaya & Gennarelli developed the ‘Centripetal Theory of Concussion’ based on studying head injury in primates. These researchers observed patterns of axonal damage with severity of injury outcome based on the notion that strain deformation was the main cause of axonal trauma. These researchers proposed that severity, ranging from transient concussion to severe diffuse axonal injury (DAI) depends on the distribution and extent of axonal damage (Ommaya & Gennarelli, 1974). Specifically, less severe injury occurs when only the cortical-subcortical areas are affected; however, more severe injury occurs as the level of trauma engages more of the brain tissues (increased volume of axonal injury) and extends inwards towards the core of the brain and brainstem. Therefore, these researchers proposed that the severity of DBI ranging from reversible axonal injury, concussions, DAI, and death is a result of increased strain magnitudes, volumes of brain tissue affected, and if specific, vulnerable brain regions are affected, i.e., the brainstem (LaPlaca et al., 2007; Ommaya & Gennarelli, 1974). Concussions are a mild form of DBI best characterized by physiological effects as opposed to mechanical effects, due to the 'negative' diagnostic results based on conventional imaging techniques (Bailes, 2009). Generally, the severity of concussion is determined based on the type and duration of symptoms and deficits however, severity is not often accurately determined until after the injury resolves (Ommaya, 1963). Characterizations of the levels of head acceleration and brain tissue strain associated with human concussion have been informed from head injury reconstruction techniques and will be described later.

2.2 Biomechanics of Concussion

The mechanical effects of impacts loads on the head result in four processes which have been studied as possible mechanisms of concussion in the literature: 1) skull deformation, 2) wave propagation, 3) linear acceleration, and 4) angular acceleration (Stalhammar, 1986).

2.2.1 Skull Deformation

Skull deformation from impact was first proposed by Gurdjian and colleagues after the observation of local and transient skull bending at the impact site (Gurdjian et al., 1964). The authors conducted a stress-coat analysis on cadavers by applying a lacquer on the surface of the skull, hitting the head, and observed stress cracks around the impact site indicative of localized and rapid in-bending of the skull. From these observations, the authors hypothesized that an impact results in localized and transient 'contact phenomena', where the skull deforms and 'snaps-back' when the load is removed. It had been proposed that skull deformation at the impact site creates an area of high pressure underneath the impact site that initiates a pressure wave throughout the cranium that could explain the transient nature of concussive effects (Gurdjian et al., 1954). When the skull deforms past a critical level, skull fracture results (Gurdjian et al., 1964; Thomas et al., 1966; Yoganandan et al., 1995).

2.2.2 Wave Propagation

Pressure waves as a mechanism of trauma was first proposed by Walker et al. (1944) who measured the electrical brain activity and intracranial pressures of animals subjected to head impact. Traditional definitions of concussion required loss of consciousness as a necessary marker for injury however, since this measure was difficult to observe in the anesthetized animal, researchers relied on a perturbation in physiological responses (Walker et al., 1944). They found that an impact to the head caused a spike in intracranial pressure that occurred at the same time as a change in electrical activity of the brain. The authors proposed that transient pressure waves caused a breakdown in the cellular membranes of axons that resulted in a traumatic excitation of the nervous tissue. They proposed that this uncontrolled excitation of the neurons was the main cause of dysfunction responsible for concussive trauma (Walker et al., 1944).

Prominent investigations led by Gurdjian and his laboratory further characterized the severity of concussion with pressure (Gurdjian et al., 1954). Pressure transducers placed throughout the brain in canines were used to map changes to the intracranial responses at impact and demonstrated an impact caused a peak positive pressure directly beneath the impact site, named the ‘coup’ pressure, and a peak negative pressure opposite the impact site, called the ‘contre-coup’ pressure (Gurdjian et al., 1964; Gurdjian et al., 1953; Gurdjian et al., 1963; Gurdjian et al., 1954; Gurdjian & Webster, 1947). This group of researchers hypothesized that concussion was a result of this imbalance of intracranial pressures, which created dynamic stresses responsible for injury (Gurdjian & Lissner, 1944; Gurdjian et al., 1963).

2.2.3 Translation

The link between translation and concussion was first purposed by Gurdjian and colleagues who found that linear acceleration correlated highly with peak pressures in anesthetized animals (Gurdjian et al., 1954; Gurdjian et al., 1955). These researchers used linear acceleration as a proxy variable because measuring intracranial pressures directly was not always feasible. Therefore, linear acceleration from an impact was deduced to be responsible for the intracranial pressure gradients (Gurdjian et al., 1953; Gurdjian et al., 1963). At high levels of linear acceleration causing severe damage, skull fractures were observed from excessive skull deformation and brain contusions were found at the coup and contre-coup sites consistent with the skull deformation and wave propagation theories previously proposed (Gross, 1958; Gurdjian, 1975; Hodgson & Thomas, 1972; Ono et al., 1980; Sano et al., 1972; Unterharnscheidt, 1972).

Linear Acceleration

Linear acceleration measures the rate of change in translational velocity of the head following an applied load and is measured as a unit of gravity g, (9.81m/s^2). Based on clinical observations, 80% of patients with skull fracture also presented with moderate concussion and this led researchers to establish skull fracture tolerances as a level for the onset of severe concussion (Gurdjian & Webster, 1947; Hodgson & Thomas, 1972). Together, skull fracture and intracranial pressure studies were used to derive the Wayne State Tolerance Curve for cerebral concussion, which was later adapted for head injury criteria, such as the Gadd Severity Index (GSI) and Head Injury Criteria (HIC) that govern current standards for head protection.

Wayne State Tolerance Curve

The Wayne State Tolerance Curve (WSTC) was originally designed to establish a relationship between the magnitude and duration of linear acceleration and onset of concussion (Lissner et al., 1960). The curve is composed of three data sets capturing linear acceleration points that covered impact durations from 1-6ms, 6-20 ms, and > 20 ms using cadaver, animal, and volunteer studies (Gurdjian & Lissner, 1944; Gurdjian et al., 1955). The WSTC as well as the data sources used to establish this curve are presented in Figure 2-1, with the red area referring to the skull fracture tests, blue area referring to research conducted on canines, and the green area referring to non-injurious volunteer research. The basis of this curve proposed that humans could tolerate high magnitudes of acceleration for very short durations, as well as low magnitudes for long durations and any responses above the line will result in concussion (Gurdjian et al., 1966; Hodgson & Thomas, 1972; Lissner et al., 1960).

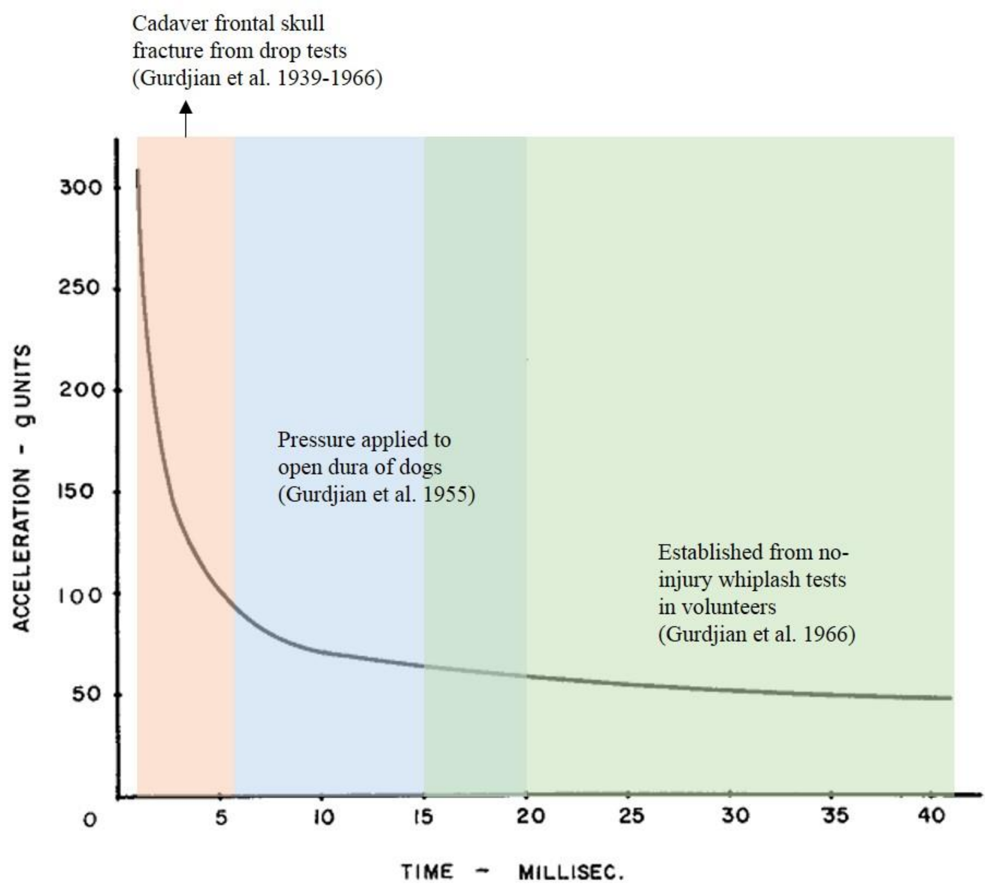

Figure 2-1 Modified linear acceleration-time concussion tolerance curve for the human head to demonstrate source of data used to derive curve. Figure taken from Gurdjian et al. 1966. Area defined in red refers to skull fracture tests, blue refers to research conducted on canines, and green refers to research conducted on volunteers.

Traditional definitions for experimental concussion required a loss of consciousness (LOC) in the animal, however in the case of human concussion, the most common injuries result in cognitive symptoms and deficits, which do not necessarily result in LOC (Ommaya, 1963). Therefore, concussion thresholds derived from skull fracture and animal LOC may not be representative of the level required to cause cognitive dysfunction in humans. Furthermore, the extrapolation of pressure pulses applied to the open dura of anesthetized dogs to the human case is inappropriate, simply because concussion does not occur in humans this way. Additionally, the cadaver skull fracture tests were from a drop test system that is characteristic falls events, however, a large portion of concussions have been attributed to collisions (Daneshvar et al., 2011). Other important factors that the WSTC neglects are impact location and angular acceleration which have been demonstrated to play an important role in concussion severity

(Hodgson et al., 1983). Despite the criticisms of the WSTC and its derivatives, it serves as an important reference point for head injury criteria and highlights that the peak and duration characteristics of the acceleration pulse are important in measuring the risk of brain trauma (Hodgson & Thomas, 1972).

2.2.4 Rotation

Holbourn (1943) was among the first to propose angular acceleration as the primary cause of concussive injuries. He deduced that the brain was more susceptible to shearing than it was to compression, based on the relative ease of deformation of the brain to shear loads. Therefore, Holbourn (1943) hypothesized that concussive injuries result from shear strains caused by angular acceleration, and that linear accelerations were less important. Holbourn tested his hypothesis using a physical model composed of a gel-filled container subjected to pure rotation in comparison with pure translation, where he observed diffuse shear strain tracts in the gel from rotation. The observations of stretching and shearing in the physical model were thought to be analogous to axonal strains under rotational loading, which was proposed as the main mechanism of injury for diffuse brain injuries, including concussion (Holbourn, 1943).

Angular Acceleration

Angular acceleration describes the rate of rotation experienced by the head and is measured in radians/second^2. The spectrum of DBI from concussion to diffuse axonal injury in animals and physical models attributes angular acceleration as the main mechanism of injury from rotationally induced shear strain damage to axons (Gennarelli, 1983; Holbourn, 1943; Jane et al., 1985; Margulies & Thibault, 1992; Ommaya & Gennarelli, 1974). Despite rotation being identified as the main contributing factor for concussion, proposed levels of angular accelerations associated with injury from animal research has been primarily studied from non-contact (impulsive) loading (Hirsch & Ommaya, 1970; Ommaya & Hirsch, 1971). Hirsch and Ommaya (1970) studied the effects of concussion in monkeys from impacts and impulsive loads to the head and reported that injuries as a result of head impact had a lower threshold to concussion, and thereby injured monkeys at lower levels as compared to impulsive loading. Their work reported that impacts and impulsive loads to the head have different thresholds for injury.

2.2.5 Strain

The link between angular acceleration, strain and axonal injury prompted experimental work to examine the mechanical effects of strain on animal white matter axons. At low and non-injurious levels of strain, the neurometabolic cascade of concussion causes an influx of ions into the cells which initiates a release of neurotransmitters. These processes disrupt cellular homeostasis of neurons, which require increased glucose metabolism by cell membrane pumps to restore ion levels to recover axonal function (Barkhoudarian et al., 2011). Galbraith et al. (1993) varied levels of strain by stretching live but isolated giant squid axons and measured the duration of electrical dysfunction of the tissue. The authors reported that strains at 0.2 resulted in transient effects of the membrane potential, and at 0.25, morphological damage resulted. Bain and Meaney (2000) conducted experiments by stretching the *in vivo* optic nerve of live guinea pigs. They hypothesized that isolated tissues would respond differently than if they were tested in the position and within the environmental conditions with which they normally function. The researchers measured the levels of disruption of the action potential across the optic nerve as an indicator for functional impairment. These researchers concluded that strains of 0.18 and 0.21 resulted in reversible injury and morphological damage of the optic nerve, respectively (Bain & Meaney, 2000). While white matter tissue injury has been reported to occur over a range of strains (0.18-0.25), the animal species as well as methods of experimentation likely contribute to the differences in the observed results. A summary of strain levels associated with transient, functional, and structural injury is presented in Table 2-1 to demonstrate the graded effects of strain. Strain rate has also been shown to play a role in the recovery of white matter axons, with higher strain rates associated with worse outcomes, however there has been less agreement of specific levels of strain rate associated with dysfunction (Galbraith et al., 1993; Maxwell et al., 1997).

Table 2-1. Summary of strain levels and the associated functional and structural effects from axonal studies.

Strain Level	5%	10%	15%	20%	25%	Author(s)
Summary of Axonal Tissues	Transient depolarization	Focal axonal transport loss	Loss of axonal transport	Axonal bulbs		Maxwell et al. 1997
Squid Axon	Spontaneous recovery	Longer axonal recovery	Residual ion deficit	Irreversible injury	Structural failure	Thibault et al. 1990
Squid Axon	Membrane potential recoverable			Axon never recovers	Structural failure	Galbraith et al. 1993
Guinea Pig Optic Nerve			Functional injury	Morphological injury		Bain & Meaney 2000

While external forces and head dynamic response have been shown to influence injury outcome, brain trauma is caused by localized tissue stresses and strains (LaPlaca et al., 2007). Due to the challenges with measuring tissue deformation *in vivo* during impact, alternative methods have been proposed to study the injury phenomena from brain deformation. To estimate the amount of tissue deformation from loading, experimental work on animal nerve tissue and finite element analysis of the human brain have been conducted. Strain is a measure of the amount of deformation of a material, and is measured by taking the ratio of the change in length of the tissue, to its original length. Research investigating human cases of concussion have used accident reconstruction with finite element analysis as a tool to derive values of brain tissue strains associated with injury (Kleiven, 2007; Patton et al., 2013; Zhang et al., 2004).

2.2.6 Cumulative Strain Damage Measure (CSDM)

Cumulative strain damage measure (CSDM) is another injury measure employed in finite element analysis to obtain the volume of the brain that experiences strain above a defined level (Bandak, 1995; Takhounts et al., 2003). CSDM has been proposed as measure of severity of diffuse brain injuries, from concussion to severe DAI based on Ommaya & Gennarelli's 'Centripetal Theory of Concussion' where larger CSDM values are expected for more severe injuries (1974). Reconstructions of concussive impacts in American football reporting CSDM values have found that ranges between 18% to 60% of the brain experiencing at least a 0.1 strain are associated with a 50% risk of concussion, and that a 55% volume of at least 0.15 strain is associated with a risk of more severe DAI (Kimpara & Iwamoto, 2012; Kleiven, 2007; Takhounts et al., 2003).

2.3 Response Levels Associated with Concussion

The WSTC illustrates the duration dependence of the brain to loading. Injury reconstructions of unprotected falls, mixed-martial arts (MMA), American football and ice hockey impacts leading to concussion similarly displayed a duration dependent relationship for peak linear and angular acceleration in more recent work by Hoshizaki and colleagues (2016). A similar relationship has also been proposed for contusions and subdural hematomas at higher levels of angular acceleration (Depreitere et al., 2005; Van Lierde et al., 2005). Injury reconstructions of sport concussions provide further information regarding the magnitudes of peak linear and angular acceleration and their durations, as well as peak strains associated with injuries. A summary of head acceleration values as a function of durations from the literature are plotted according to the WSTC and Hoshizaki et al. (2016) tolerance curve shapes to demonstrate the trend in the responses (Figure 2-2).

Reconstructions of falls onto rigid surfaces and MMA impacts (punches and kicks) account for pulse durations of approximately 1-10 ms; helmeted impacts in ice hockey and American football, as well as head-body impacts in Australian rules football account for roughly 10-20 ms in duration; and shoulder collisions in ice hockey (well-padded conditions) last for approximately 20-30 ms (Brolinson et al., 2006; Cournoyer et al., 2014; Duma et al., 2005; Frechede & McIntosh, 2009; Kendall, 2016; McIntosh et al., 2014; Oeur et al., 2015; Patton et al., 2013; Pellman et al., 2003; Post et al., 2015; Rousseau, 2014; Rowson et al., 2012; Schnebel et al., 2007; Zhang et al., 2004). Injury reconstruction combined with finite element analysis of a human brain model provides an opportunity to evaluate the severity of three-dimensional head acceleration loads and their time courses reflected using peak strain measures that take into account the brain material properties and geometry (Bandak, 1995). Reports of strain levels from injury reconstructions of concussive impacts from Australian rules rugby, American football, and unprotected falls gives peak strains from 0.15 - 0.48 for various brain regions (Table 2-2).

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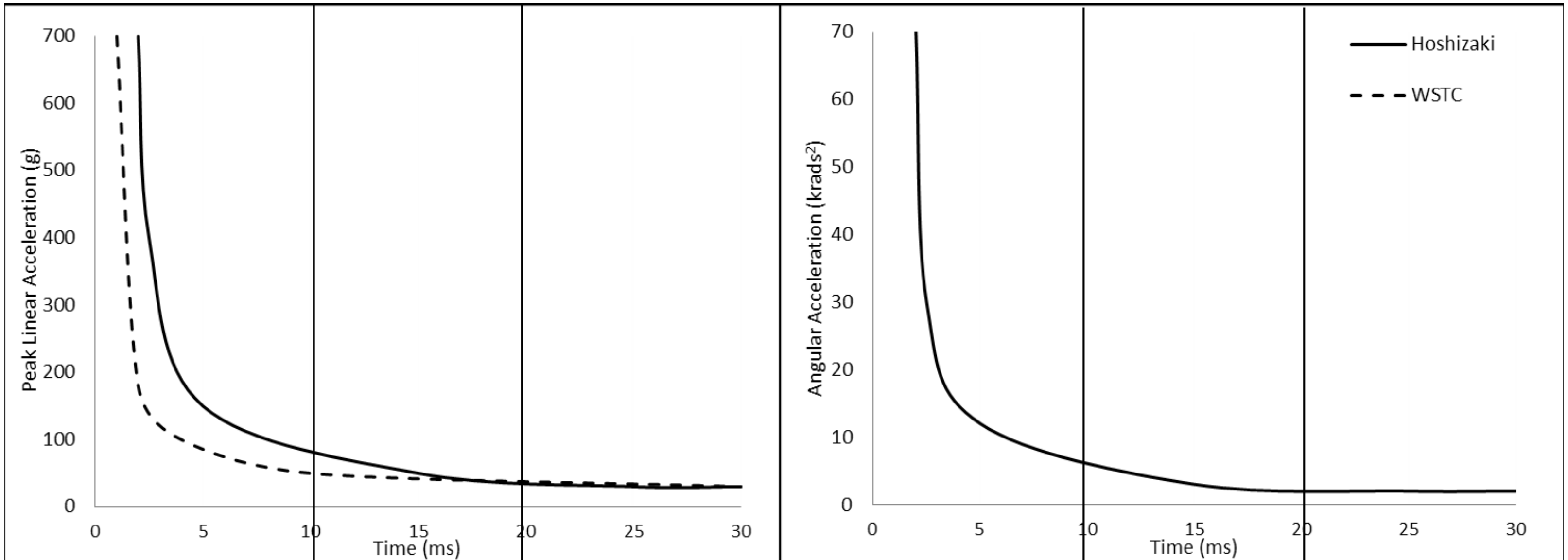


g	57-277 g	81-149g	23-30g	krad/s^2	9.9-19.6	4.1-12.2	2.9-3.2
Data set	MMA; unprotected falls	Hockey; Australia rules football; American football (sensors data and reconstructions)	Hockey shoulder collisions	Data set	MMA; unprotected falls;	Hockey; Australia rules football; American football (sensor data and reconstruction)	Hockey shoulder collisions
Author(s)	Post 2015; Kendall 2016	Cournoyer 2014; Oeur 2015; Frechede 2009; McIntosh 2014; Patton 2013; Schnebel 2007; Rowson 2012; Zhang 2004; Kendall 2016; Brolinson 2006; Pellman 2003; Duma 2005	Rousseau 2014; Kendall 2016	Author(s)	Post 2015; Kendall 2016	Cournoyer 2014; Oeur 2015; Frechede 2009; McIntosh 2014; Patton 2013; Kendall 2016; Rowson 2012; Pellman 2003; Zhang 2004	Rousseau 2014; Kendall 2016

Figure 2-2. A summary of peak linear and angular acceleration responses from the literature plotted according to the magnitude-duration relationship established from the WSTC (Lissner et al. 1960) and Hoshizaki et al. (2016)

A large range of overlapping values for head accelerations and strains are reported in the literature for sport related concussions. While this is depicted in Figure 2-2, an overall trend demonstrating a magnitude-duration relationship for concussive injury can be observed. Large differences in peak response values associated with injury can be attributed to measurement and laboratory variance as well as natural human variance. Laboratory variance encompasses the various test headforms, test equipment, computational models, selection of material properties and geometries used to approximate the complex human head and brain. Human variance encompasses individual characteristics, such as injury history, genetics, sex, and age that influence injury tolerance (Kutcher & Eckner, 2010). Additionally, concussion has been described as a complex functional injury associated with a wide-ranging list of signs and symptoms (McCrory et al., 2017). A symptom centric concept of concussion has been proposed, in which symptoms are reflective of the various brain regions they affect (e.g., vestibular vs auditory concussion) (Gennarelli, 2015). Understanding the limitations of this data is an important aspect for the application of this information. Levels of peak linear and angular acceleration, and strain responses associated with concussion can be used to establish a set of reference values in the literature for concussive injuries for duration-specific conditions and play an important role for informing target values that could be used to evaluate the performance of head protection that are more reflective of the injury causing events.

Table 2-2. Summary of reported strain values associated with concussion from injury reconstructions in the literature.

Value	**Strain Variable**	**Region**	**Author(s)**
0.15 (50% risk)	Max. Principal Strain	Midbrain	Patton et al. (2013)
0.15 (50% risk)	Max. Principal Strain	Corpus Callosum	Patton et al. (2013)
0.19 (50% risk)	Shear Strain	Midbrain	Zhang et al. (2004)
0.21 (50% risk)	Max. Principal Strain	Corpus Callosum	Kleiven (2007)
0.26 (50% risk)	Max. Principal Strain	Gray Matter	Kleiven (2007)
0.27 (50% risk)	Max. Principal Strain	Gray Matter	Patton et al. (2013)
0.28 (Avg)	Max. Principal Strain	Corpus Callosum	McAllister et al. (2012)
0.31 (50% risk mild DAI)	Max. Principal Strain	-	Deck and Willinger (2008)
0.32 (Avg)	Max. Principal Strain	Cerebrum	Viano et al. (2005)

0.32 (50% risk)	Max. Principal Strain	-	Kimpara and Iwamoto (2012)
0.32 (50% risk)	Shear Strain	-	Kimpara and Iwamoto (2012)
0.34 (Avg)	Max. Principal Strain	Midbrain	Viano et al. (2005)
0.38 (Avg)	Max. Principal Strain	White Matter	Post et al. (2015)
0.44 (Avg)	Max. Principal Strain	-	Oeur et al. (2015)
0.48 (Avg)	Max. Principal Strain	Gray Matter	Post et al. (2015)
0.6 (50% risk)	CSDM10		Kleiven (2007)
0.18 (50% risk)	CSDM10		Kimpara and Iwamoto (2012)
0.55 (50% risk of DAI)	CSDM15		Takhounts et al. (2003)

2.4 Head & Brain Injury Surrogate Models

Animal, cadaver, anthropometric test devices (ATD), and finite element models have been used to study the mechanisms of head injury to describe the mechanical forces associated with injury processes (Gurdjian et al., 1954; Holbourn, 1943; Pellman et al., 2003; Willinger & Baumgartner, 2003). Animal subjects are advantageous as they allow the real-time, physiological manifestations of injury to be studied, however an unknown scaling factor needs to applied to make it useful for human head injury (Gennarelli, 1994; Ueno et al., 1995). While cadavers have the same geometry and weight distribution as the living human head, the effects of embalming and tissue decomposition produce inconsistent impact response (van Dommelen et al., 2009). ATDs were developed to study head injury by best approximating the complex human response with a simplified model (Patrick, 1973). ATDs mimic the mass distributions and geometries of the human head and are tuned to provide a human-like response under similar test conditions. Test headforms are commonly developed from solid materials such as steel and rubber, giving robust and reliable results when subject to repeated testing, but in doing so, biofidelic aspects are sacrificed. Finite element (FE) models of the human brain are a tool that permit an analysis of the intracranial deformations from loading, taking into account the geometry and material properties of the head and brain (Yang et al., 2006). Combining ATD physical impacts with FE brain analysis provides an opportunity to understand the characteristics and subtleties of impact loading in a systematic and controlled environment and the risk for head injury.

2.4.1 Anthropometric Test Devices (ATD)

Two important considerations for an ATD are that they must represent human size and shape and produce representative responses under similar loading conditions. Some degrees of freedom are sacrificed in order to obtain repeatability in the ATD however, since human variability contributes to a wide range of responses, if the ATD response falls within this range, the information is valuable to represent certain groups of the population (Patrick, 1973). Three head and neckform sizes were used to modulate impact mass for fall events; the Hybrid III 6 Year Old Child Headform (H-III6C), the Hybrid III 5th Percentile Female Headform (H-III5F), and the Hybrid III 50th Percentile Adult Male Headform (H-III50M). The range of headform sizes used are consistent with the range of head circumferences for child, adolescents, and adult sizes and are used to represent these age groups in sport (Roche et al., 1987). For collision investigations, only the adult male headform was used to examine modulating impact parameters and striking mass was a reflection of the collision condition.

The Hybrid III ATD family are composed of a steel headform that is covered with a vinyl layer to simulate the soft tissues of the skin. These devices were originally developed for study of injuries in frontal automobile crash tests and are a widely accepted standard for crash injury investigations involving passenger safety, air bags and seat-belt restraint systems (Mertz et al., 1997; National Highway Traffic Safety Administration, 2008; Prasad, 2015). Early stage development and validation consisted of a comparison of available test headforms with cadaver data and these included the adult male Hybrid III, a part 572 headform, repeatable Pete, and 2 Wayne State University (WSU) headforms (Mertz, 1985). These headforms were tested at 0.254, 0.635, 0.762 m drop heights. It was concluded that the Hybrid III headform produced the best comparison with the cadaver data. As a result, the child and female headforms and neckforms were developed after the adult male headform and are scaled versions best representing the size and weight of US children and small adults (Irwin & Mertz, 1997; Mertz et al., 1989). These scaling factors were also applied to the biomechanical impact response for the head and neck defining size appropriate response corridors (Mertz et al., 1989).

The standard Hybrid III neckforms have a directional constraint built into their design with slits at the front of the neck and a continuously joined rear side. This construction allows for increased extension and stiffer flexion in the response of the neck during whiplash (non-contact) high energy car crash scenarios (>50ms in loading duration). However, the appropriateness of these constraints and their effects are unknown for direct head impacts in sport loading conditions (<50ms). In order to isolate and understand the effects of impact parameters on head dynamic response, a set of non-directional neckforms have been developed at the University of Ottawa (Ottawa, Canada) matching mass and geometrical dimensions of the standard necks. The directional constraints are removed and replaced with uniform circular steel and rubber plates that are held together with the standard neckform Hybrid III cable tightened with a bolt as per the manufacturer's specifications. Physical characteristics of the head and neckforms used in this research as summarized in Table 2-3 and shown in Figure 2-3.

Table 2-3. Head and neckform dimensions.

		Child	Adolescent	Adult
Headform	Mass (kg)	3.47	3.73	4.54
	Circumference (m)	0.52	0.54	0.58
Standard Neckform	Mass (kg)	0.54	0.91	1.54
	Height (m)	0.08	0.11	0.12
	Diameter (m)	0.06	0.07	0.08
	Cable Torque (Nm)	0.22	1.36	1.40
Non-Directional Neckform	Mass(kg)	0.42	0.80	1.30
	Height (m)	0.08	0.12	0.13
	Diameter (m)	0.06	0.07	0.08

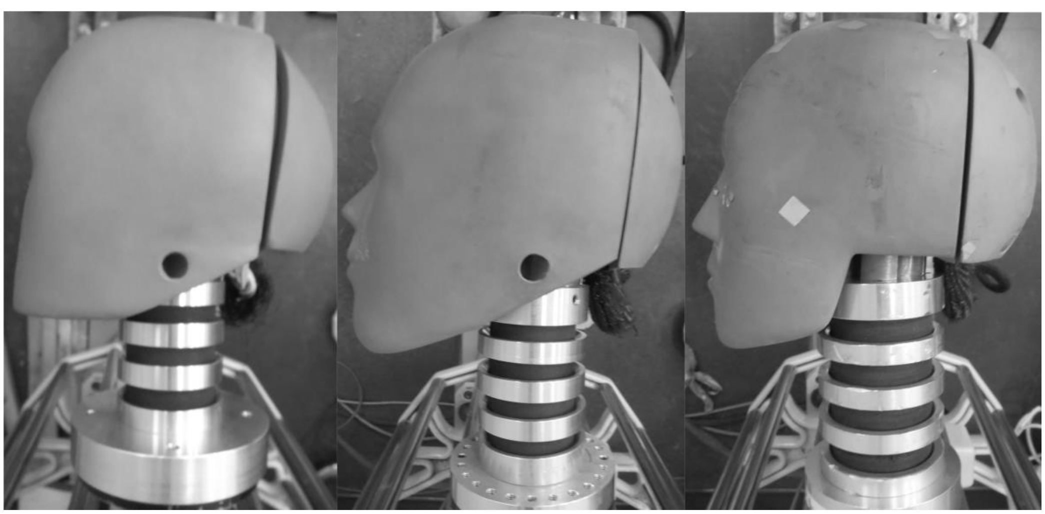


Figure 2-3. Child (left), adolescent (middle), and adult (right) Hybrid III headforms and non-directional neckforms.

The adult headform was validated with comparisons against a series of cadaver drop tests and is considered valid up to skull fracture levels (Mertz, 1985). Test conditions specify a 0.376 m drop onto a flat, rigid surface with an acceptable headform peak resultant linear acceleration tolerance range for the representative child, female, and adult headforms are presented in Table 2-4. The standard Hybrid III neckforms are required to respond within corridors of neck moments (Nm) for flexion and extension from non-impact, deceleration types of loads, typical of car crash loading environments (Mertz et al., 1989; Mertz & Patrick, 1971). Initial comparisons between the adult non-directional neckform and the standard neckform in conjunction with the Hybrid III head was conducted by Walsh et al. (2012), who delivered headform impacts at 5.5 m/s using a pneumatic linear driven impactor equipped with a stiff modular elastomer programmer (MEP) pad for centre gravity and non-centric impacts. The effect of the neckform was compared using peak linear and angular head acceleration, and while the standard neck produced significantly higher head angular accelerations (28.5 krad/s^2) than the non-directional neck (26.3 krad/s^2), these differences were not meaningful differences in the production of overall head injury risk.

Table 2-4. Peak linear acceleration requirements for the child, adolescent, and adult headforms.

		Child	Adolescent	Adult
Peak Resultant Head Acceleration Requirements (G)	Lower Limit	245	240	225
	Upper Limit	300	295	275

A comparison of the standard neckform and the non-directional neckform on head impact response under the loading conditions explored in this research was conducted and presented in Appendix B. Peak linear and angular acceleration responses were comparable between neckform types except for the front boss under steel at 3.0 m/s (high-energy impact scenario). The standard Hybrid III neck and headform (27.7 krad/s^2) had nearly twice as high angular accelerations than the non-directional neck and Hybrid III headform (16.3 krad/s^2). Due to the high-energy impact scenario induced by the steel impact surface, it is possible that the directional constraints in the Hybrid III neckform (frontal slits) in a combination with the oblique nature of the impact location (mid-point between the front and side impact locations) contributed to stiffening up the head-neck as a unit causing increased impact response. These tests provide a reference for comparing the standard and non-directional neckforms. The non-directional neckforms were used in this research to minimize any other potential directional effects to best isolate the phenomena of modulating impact parameters on head dynamic response.

2.4.2 Finite Element Analysis

The finite element method is a numerical modeling approach to studying how mechanical loads affect a system. The external forces, the geometry of the object (composed of groups of elements), and the material properties of the system are known so that variables such as strain can be computed (Fish & Belytschko, 2007). For human head injury, the application of FE analysis sets out to study the response of the head and brain system from adverse mechanical loading. The finite element model used to approximate intracranial response in this research was the University College Dublin Brain Trauma Model (UCDBTM) developed by Horgan and Gilchrist (2003, 2004).

Geometrical accuracy of the UCDBTM was guided using computed tomography (CT) scans, magnetic resonance images (MRI), and sliced colour images of a male and female human cadaver (Horgan & Gilchrist, 2003). The anatomical features include the scalp, a three layered skull (diploe, outer and inner tables), dura, pia, falx, tentorium, cerebrospinal fluid, cerebrum, cerebellum and brain stem, totalling over 26 000 elements. The brain is a biological soft tissue composed of 75-80% water and has been described as a gel-like material exhibiting non-linear, time dependent (viscoelastic) properties (Ommaya, 1968; Prange & Margulies, 2002). The challenge with modeling an organ like the brain is that there is uncertainty in the material properties and that the macro- and microscopic structures lead to a diverse range of reported tissue characteristics (Chatelin et al., 2010; Viceconti et al., 2005). One notable characteristic of the brain, is its relatively large compressive strength (bulk modulus) in comparison to its shear strength (shear modulus) therefore; the brain is less resistant to shear deformation (shear strain injury) than compressive deformation (Holbourn, 1943; Ommaya, 1968). Brain tissue compressibility in the UCDBTM was adopted from finite element models developed by Ruan (1994) and Ward (1982), who proposed a Poisson’s ratio between 0.48-0.499 as reasonable to model the intracranial contents, taking into account the pressure relief systems of the foramen magnum, the lateral ventricles allowing some movement of the CSF, and blood vessels present in and around the brain (Horgan & Gilchrist, 2003, 2004). The shear properties of the brain model were taken from Zhou et al. (1995) who deduced a mathematical relationship from brain tissue tests by Shuck and Advani (1972). To take into account non-linear and viscoelastic properties, a material model developed specifically for finite element codes proposed by Mendis et al. (1995) was implemented in the UCDBTM, which has also been employed by other modellers (Kleiven

& von Holst, 2002). A summary of the material properties of each component in the UCDBTM is shown in Table 2-5.

Table 2-5. Material properties deformation characteristics defined in the UCDBTM.

Material	Shear Modulus (kPa)		Decay Constant	Bulk Modulus
	G_0	G_∞	(s^{-1})	GPa
Grey Matter	10	2.0	80	2.19
White Matter	12.5	2.5	80	2.19
Brain Stem	22.5	4.5	80	2.19
Cerebellum	10	2.0	80	2.19
	Young's Modulus (MPa)		Poisson's Ratio	Density (kg/m^3)
Scalp	16.7		0.42	1000
Cortical Bone	15 000		0.22	2000
Trabecular Bone	1000		0.24	1300
Dura	31.5		0.45	1130
Pia	11.5		0.45	1130
Falx	31.5		0.45	1130
Tentorium	31.5		0.45	1130
CSF	-		0.5	1000
White Matter	Hyper elastic		0.499981	1040
Grey Matter	Hyper elastic		0.499981	1040

The response of the UCDBTM was compared with cadaver head impact tests by Nahum et al. (1977), Trosseille et al. (1992), and Hardy et al. (2001) for pressure, displacement, and brain acceleration are detailed in Table 2-6. The input parameters had impact velocities ranging from 2-10 m/s with peak acceleration responses ranging from 24-204 g and 1800-7600 rad/s^2 for short and long pulse durations (6.5-20 ms). These impact characteristics are within range of concussive level impacts in sport (Figure 2-2). Due to the lack of *in vivo* human response data at injurious levels of loading, these data provide an end-point comparison to ensure the model accurately predicts brain impact phenomena. Parametric tests for low and high values for brain shear and bulk moduli were varied in the UCDBTM, as well as material properties employed in other FE models by Ruan (1994), Zhou et al. (1995), and Kang et al. (1997) were used and compared against cadaveric responses (Horgan & Gilchrist, 2003). The material properties proposed by Zhou et al. (1995) gave the best correlation with experimental results.

Table 2-6. Summary of cadaver tests used for comparisons with the UCDBTM.

Author (s)	*Velocity (m/s)*	*Mass (kg)*	*Acceleration Response*	*Pulse Duration*	*Brain Response*
Hardy 2001	2	Head/neck	24 g; 1813 rads/^{2}	20 ms	Displacement

Trosseille 1992	7	23.4	102 g; 7605 rad/s^2	15.8 ms	Acceleration
Nahum 1977	9.94	5.59	204 g	6.5 ms	Pressure

Since its development, the UCDBTM has been employed in traumatic brain injury investigations involving accident reconstructions. Doorly and Gilchrist (2006) and Post (2013) reconstructed fall related cases resulting in skull fractures, subdural hematomas, and contusions demonstrating consistent peak tissue stresses and strains as with the literature (Kleiven, 2007; Willinger & Baumgartner, 2003). The UCDBTM has also been used in accident reconstructions of transient and persistent concussions involving adolescents (Dawson, 2016) and adults (Kendall, 2016; Oeur et al., 2015; Rousseau, 2014), and while there is no end-point comparison with observable damage on diagnostic images, the peak brain strains have been consistent with other accident reconstruction efforts examining sports related concussions employing other FE models (Kleiven, 2007; Patton et al., 2013; Zhang et al., 2004). Furthermore, a comparison of peak brain tissue responses from transient concussions and persistent concussions demonstrated trends by which more severe outcomes displayed higher peak strains (Oeur et al., 2015), consistent with the theory of diffuse brain injuries where greater peak strains cause more severe injury (Gennarelli et al., 1998; Ommaya & Gennarelli, 1974).

2.5 Summary

Physical human head and neck surrogates play an important role in head impact investigations and are used by researchers as tools for injury reconstruction and performance evaluations of protective equipment in sport, including helmets (Bartsch et al., 2012; Newman et al., 1999; Oeur et al., 2014; Pellman et al., 2003). The extensive design and development of the Hybrid III headform has made it a desirable and robust tool for studying head impact in sport. While the headform lacks biofidelity in comparison to the more compliant human tissues, it provides repeatable dynamic impact response data for a large range of impact energies. The advantage of using this headform is that it allows for linear and angular acceleration data to be captured, providing a full three-dimensional kinematic analysis of the impact event. The UCDBTM uses the three dimensional linear and angular accelerations collected from the Hybrid III headform and calculates the resulting brain tissue strains from these loads, taking into consider geometrical and material properties of the model. Combining physical impact tests on a headform with finite element modeling has allowed for an analysis of the relationship between the mechanical event and impact response. Accident investigations linking the impact event and injury outcome has provided an opportunity to explore gross environmental characteristics, effects on head dynamic response, resulting brain tissue strains, and hypotheses about the processes leading to injury and injury outcomes.

3 Study Design

Impact Parameter Selection | Research Design

3.1 Head Impact Parameters

This chapter reviews the effects of mechanical impact parameters on head and brain impact response as reported in the literature and presents the levels of velocity, mass, compliance, and impact location selected for parametric study to identify interaction effects for head impact conditions in sport.

3.1.1 Impact Velocity

Early experimental work by Gurdjian et al. (1964) investigated the effect of velocity, mass, and compliance on impacts delivered to cadavers measuring peak resultant linear acceleration. Gurdjian et al. (1964) conducted bare head (un-filled triangles) and helmeted (filled triangles) forehead impacts from drop tests (Figure 3-1). The impacts were delivered by raising the cadaver via a drop test system and allowing the cadaver forehead to contact a heavy steel block secured to the concrete floor. In a separate set of tests, a rotatory weighted hammer of 1 kg (un-filled markers) and 6 kg (filled markers) was used to impact the side location of a helmeted (squares) and bare (circles) cadaver head (Figure 3-2). The head was situated on a freely movable platform to replicate collision impacts. The researchers recorded impact velocity (m/s) and linear acceleration (g) of the head in both tests. In Figure 3-1 and Figure 3-2, a clear trend is observed in the data where a positive relationship exists between impact velocity and head linear acceleration (Gurdjian et al., 1964). These findings are limited as they neglect the effects on angular acceleration and were tested for limited impact locations (front and side).

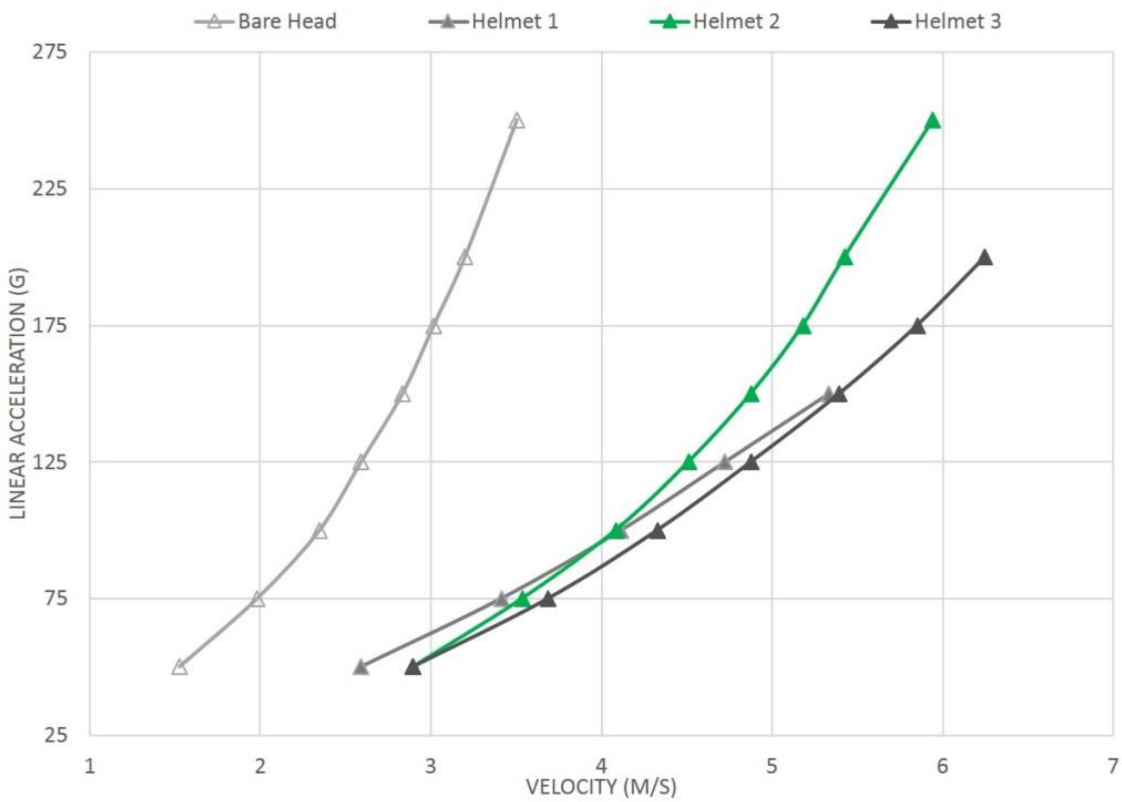


Figure 3-1. Experimental results from Gurdjian et al. (1964) of cadaver drop tests without a helmet (bare) and with helmets.

Impact velocities associated with concussion from falls and collisions in football, hockey, soccer, boxing and rugby have reported ranges between 1-9 m/s for falls and 1-15 m/s for collisions (Atha et al., 1985; McIntosh et al., 2000; Pellman et al., 2003; Smith et al., 2000; Stojsih, 2012; Walilko et al., 2005; Withnall et al., 2005). Four levels of impact velocity are used in this research to capture a range of responses associated with sub-concussion up to concussion levels: 1.5 m/s, 3.0 m/s, 4.5 m/s, 6.0 m/s. Impacts beyond 6.0 m/s result in responses associated with a high risk of concussion and a risk for more severe TBIs, such as skull fracture and subdural hematomas (Hoshizaki et al., 2016). It was in the interest of this research to examine interaction effects from sub-concussion to concussive levels as higher impact velocities are associated with more severe injuries.

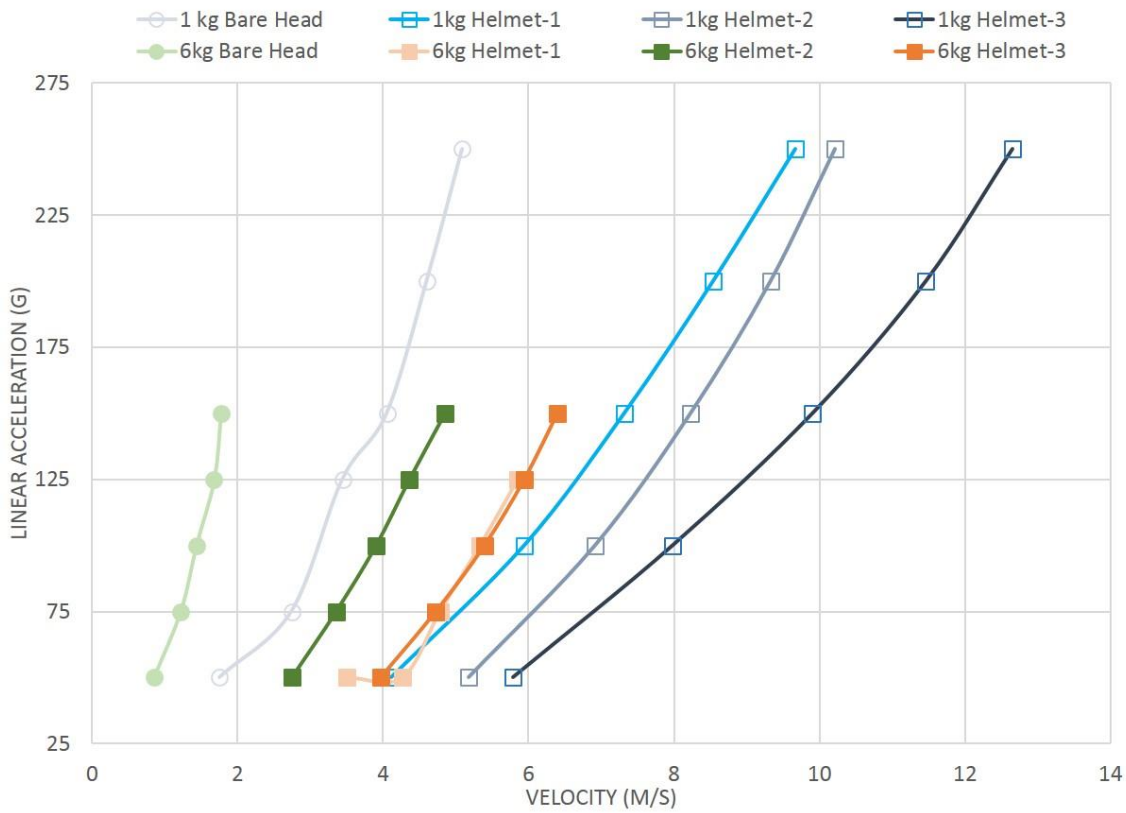


Figure 3-2. Experimental results from collisions tests with 1 and 6kg weights conducted by Gurdjian et al. (1964a) on helmeted and non-helmeted cadavers.

3.1.2 Impact Compliance

Impact compliance defines the stiffness of the materials involved in the impact event. The impact compliance (soft or stiff) influences the rate of energy transferred to the head, which directly affects the magnitude and duration of the acceleration pulse (Gurdjian et al., 1964). For non-compliant (stiff) surfaces, a high force will be transferred to the head at a high rate, causing a high magnitude and short duration acceleration pulse. This type of condition is associated with increased risk of injury. For compliant surfaces, some of the impact force will be attenuated by the material and the head experiences a lower magnitude and longer acceleration pulse (Gurdjian et al., 1964; Hoshizaki & Brien, 2004). To demonstrate the effect of compliance, adding a helmet lowers the magnitude of peak linear acceleration in comparison with bare head impacts (Figure 3-1and Figure 3-2). In sport, the majority of impacts can be grouped into three distinct compliance conditions. The first condition is where the head contacts a hard object or surface, where no protection is worn and can be described as low compliance. These types of events result in high magnitude accelerations over approximately 5 ms in duration (Doorly et al., 2005; Oeur et al., 2015; Post et al., 2014). In contrast with medium compliance, some level of protection is worn, such as a helmet on the head that causes a relatively lower magnitude but longer duration acceleration pulse lasting approximately 15 ms. High compliance conditions are where the head contacts well-padded and/or protected objects such as a shoulder pad, further lowering the peak acceleration magnitude and increasing the pulse duration to approximately 30 ms. Representative impacts for a bare head onto ice (low), football helmet impact (medium), and a headform impact onto a well-padded surface (high), such as a gymnastic mat, consistent with the compliance of a shoulder pad collision is illustrated in Figure 3-3 (Mills et al., 2006; Oeur & Hoshizaki, 2016; Rousseau, 2014; Zhang et al., 2004). Surrogate surfaces used to match the impact loading characteristics of these distinct conditions are steel, 2.5 cm vinyl nitrile foam, and 6.7 cm Rubatek Rubber R338 foam (Rubatek LLC, Virginia, USA), with ASTM Shore A stiffness values of 99.3, 17.8, and 3.0, respectively. These three impact compliances were used throughout the dissertation to represent unique sport compliances.

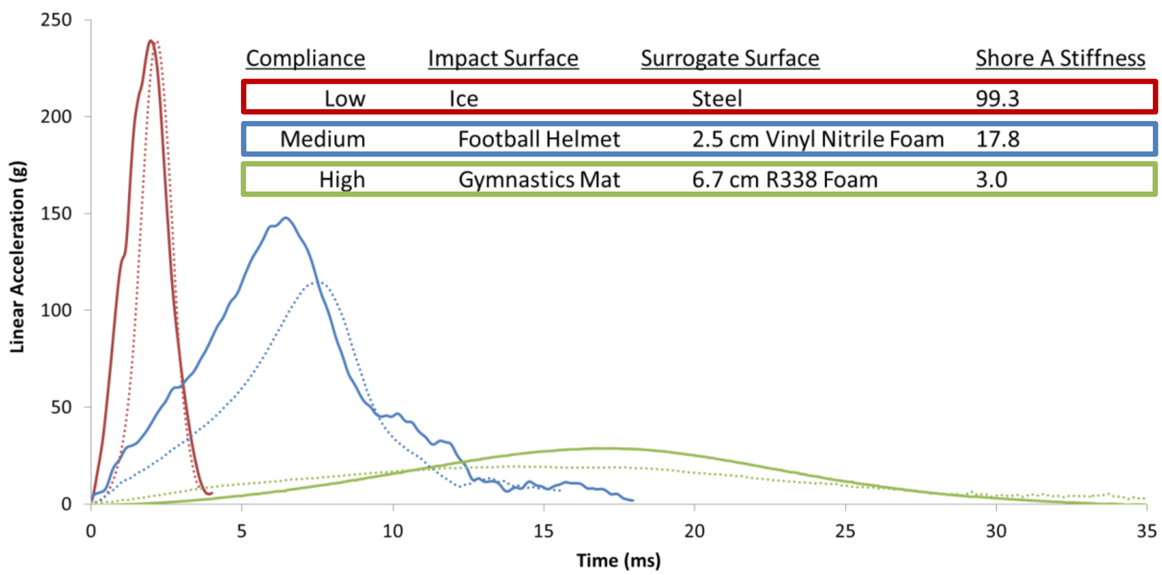


Figure 3-3. Linear acceleration histories of low, medium, and high compliance conditions representing a bare head impact onto a stiff ice surface (red), a football helmet impact (blue), and a gymnastics mat impact (green), respectively (solid lines). Surrogate surfaces used to replicate these loading conditions are steel, 2.5 cm vinyl nitrile and 6.7 cm R338 foams (dotted lines). Relative stiffness measures are presented accordingly.

3.1.3 Impact Mass

Figure 3-2 also displays the effects of striking mass for collisions (1 and 6 kg), where an increase in mass resulted in an increase in cadaver response (Gurdjian et al., 1964). The effect of striking mass on bare headform impacts similarly increase peak linear and angular acceleration (Karton et al., 2013), however these effects were not as large in comparison to velocity (Figure 3-4). Karton et al. (2013) isolated the effect of impact mass (4.3 – 14.3 kg) on linear and angular acceleration, by keeping all other parameters constant: impact velocity at 4 m/s, impact compliance consisted of a bare headform and a stiff modular elastomer programmer (MEP), and the impact direction occurred along the frontal plane. For the conditions studied, impact mass seemed to have some effect on linear acceleration with values ranging from 159 – 226 g, however, mass seemed to have less of an effect on angular acceleration with values ranging from 18.4 - 19.9 krad/s^2. The author noted a stabilization in the dynamic response data at 10.3 kg and hypothesized that a saturation of the system compliance caused little changes in head acceleration values (Karton et al., 2013). This is likely due to the stiff MEP impact surface used during the tests which masked the effect of striking mass on head acceleration.

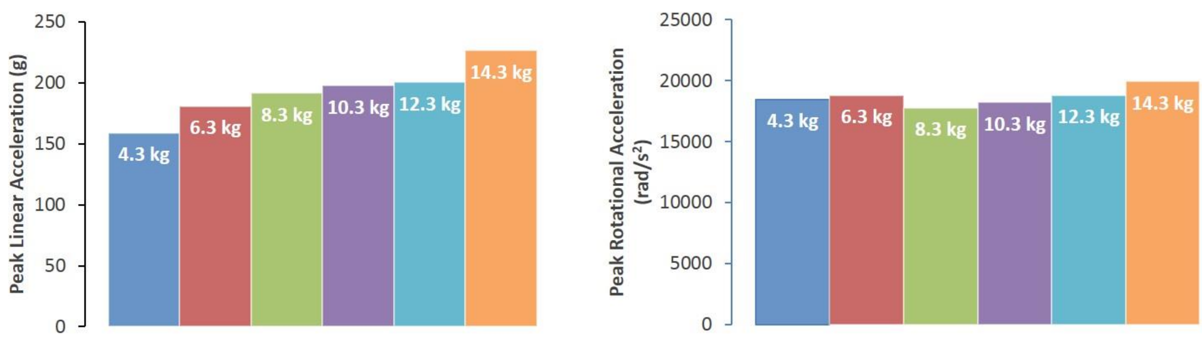


Figure 3-4. The effect of impact mass for collision type impacts on head dynamic response (adapted from Karton, 2012).

For this research, the levels of impact mass were specific to the impact event. For falls, impact mass was varied using three different sized headforms to capture children, adolescent, and adult head weight and geometries. The 6 Year Old Hybrid III Child Headform (H-III6C), 5th Percentile Female Headform (H-III5F), and 50th Percentile Adult Male Headform (H-III50M) with head circumferences of 0.52, 0.55, and 0.57 m respectively, represent children ages 5-8, adolescents ages 11-16, and adults ages 18 and over in Figure 3-5 (Roche et al., 1987).

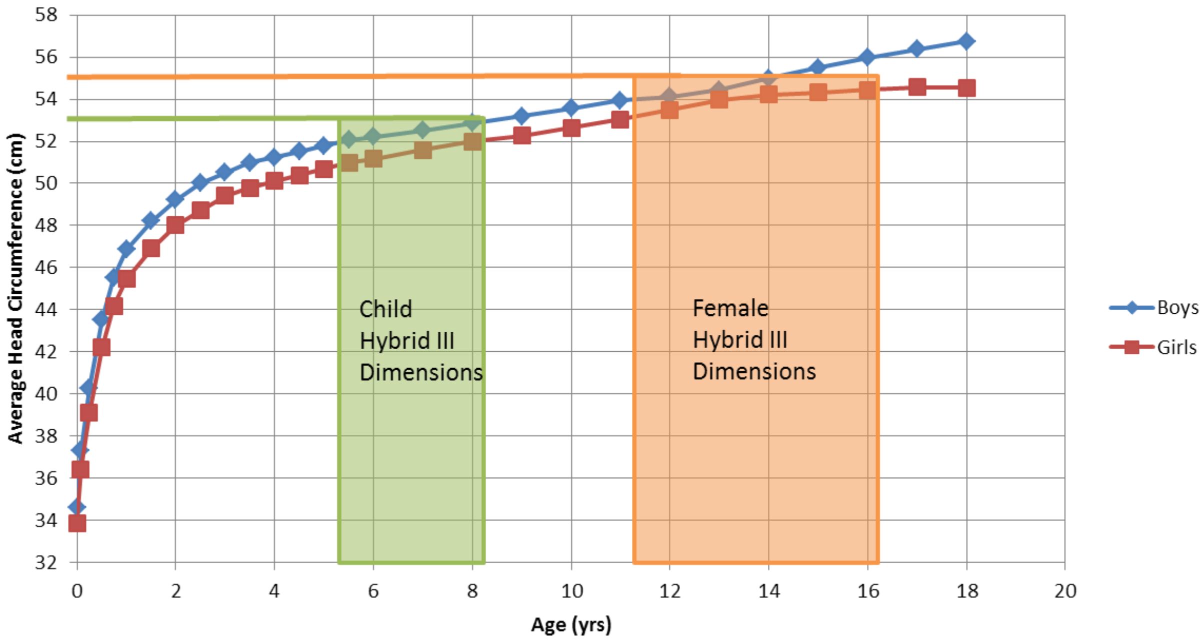


Figure 3-5. Average head circumference for boys and girls taken from Roche et al. (1987) with Hybrid III headform dimensions plotted.

In a collision, impact mass varies with the striking object. The range of reported striking masses and associated impact velocities reported in the literature for sport collisions are summarized in Figure 3-6 and are presented according to the relative levels of compliance. For low compliant impact surfaces, consisting of an unprotected body part, low impact masses constitute elbow to head strikes in ice hockey (1-5 kg), head to head collisions in soccer, and mixed martial arts (MMA) punches. These transition to medium and high mass conditions with knees and kicks to the head (MMA), and full head-body collisions (de Leva, 1996; Fife et al., 2012; Katis & Kellis, 2010; Kendall, 2016; McIntosh et al., 2000; Rousseau & Hoshizaki, 2015; Seidi et al., 2014; Walilko et al., 2005). For medium compliance, where the impacting surface is protected with a helmet or padding, striking mass ranges from tucked-in elbow strikes in ice hockey, where the player engages more of the torso mass in the collision (Bartsch et al., 2012; Rousseau & Hoshizaki, 2015), punches from elite boxers, helmet-to-helmet, and body-helmet collisions (Atha et al., 1985; Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Smith et al., 2000; Stojsih, 2012; Walilko et al., 2005; Withnall et al., 2005; Zanetti et al., 2013). For high compliance, where the body part is well protected, extended elbow strikes and shoulder collisions in sports like ice hockey constitute the range from low to high mass conditions (Kendall, 2016; Rousseau, 2014). To capture the range of possible striking masses characteristic of sport collisions, four masses were used in this research, 3, 9, 15, and 21 kg to best represents the conditions in Figure 3-5.

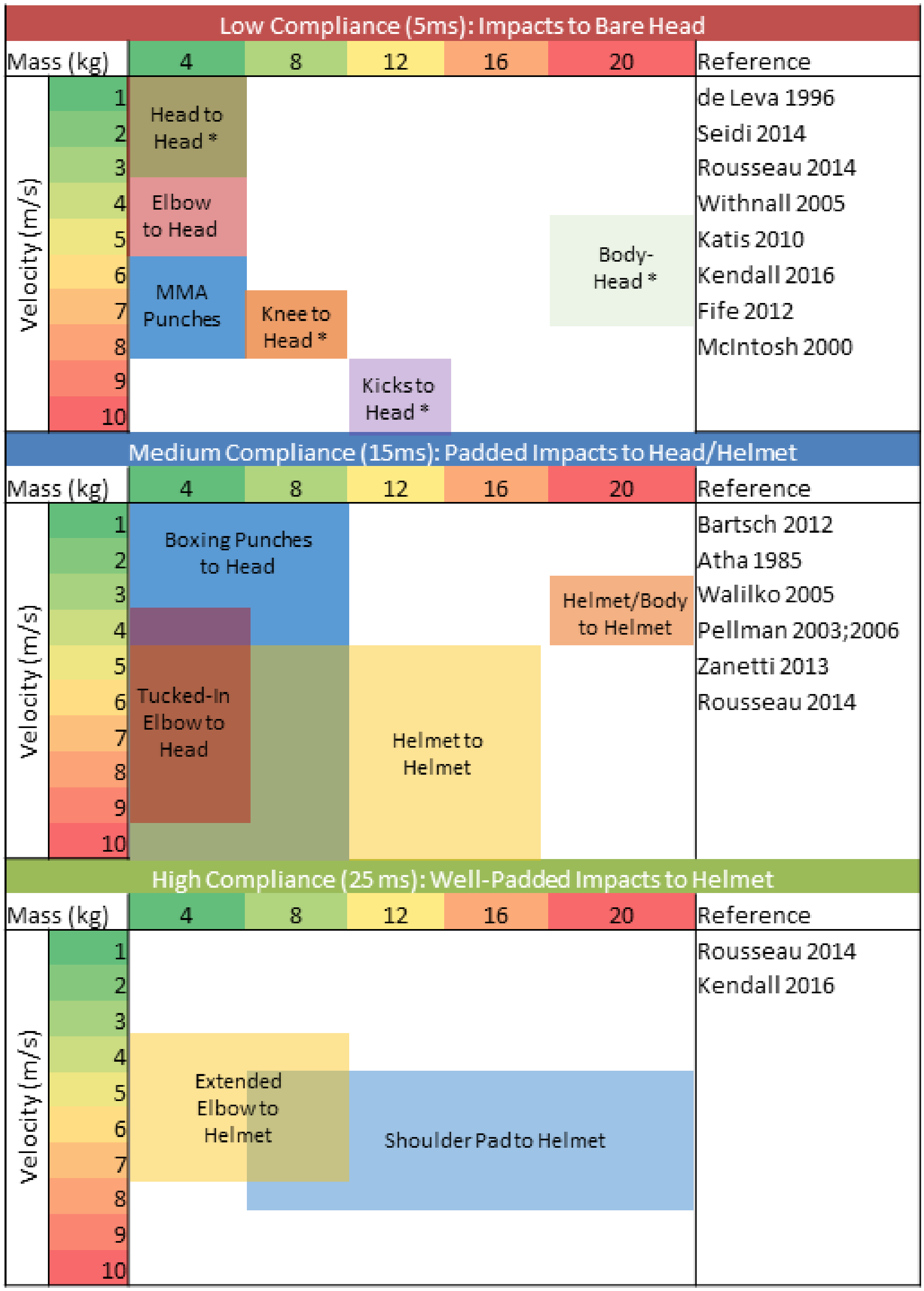


Figure 3-6. Striking masses and velocities of typical collision events separated for low, medium, and high compliance conditions.

3.1.4 Impact Location

The effect of impact location on risk and severity of concussion was first investigated by (Hodgson et al., 1983), using animal models. This group of researchers used compressed air to impact helmeted anesthetized monkeys at the front, side, rear, and top locations, with the head and neck freely movable after impact. The results from this study demonstrated that the side impact location, motion about the coronal plane, was associated with a decreased tolerance to concussion as compared to the other impact locations (Hodgson et al., 1983). The increased risk of brain trauma has been hypothesized to be a result of the relatively rigid falx partition between the right and left hemispheres that creates increased strains on the adjacent softer cerebral tissues from coronal motion (Gennarelli et al., 1987; Gurdjian et al., 1963; Smith & Meaney, 2000). Directional sensitivity of the head has also been supported by other work on physical impacts to anthropometric dummies, and finite element simulations, where side impacts were found to result in increased head accelerations and brain tissue strain (Gennarelli et al., 1982; Hodgson et al., 1983; Walsh et al., 2011; Zhang et al., 2001). Whether the head is impacted through the centre of gravity, or outside the centre of gravity influences the relative magnitudes of linear and angular acceleration of the head (Gennarelli et al., 1987; Ommaya, 1966). A series of centric and non-centric impacts delivered to a headform demonstrates the range of linear and angular acceleration (Figure 3-7) across different combinations of five impact locations (front, front boss, side, rear boss, and rear) and four angles (centre gravity, positive elevation, positive/negative azimuth) (Walsh et al., 2011). Based on Walsh et al.'s data, four impact locations covering primary motions in the sagittal, frontal, and combined axes were selected and used in this dissertation. A non-centric impact condition associated with a rotationally dominant response was also included. These impact locations are front CG, front boss CG, side CG, and rear boss negative azimuth (NA), which are identified in Figure 3-7 and result in distinct peak resultant linear and angular acceleration responses.

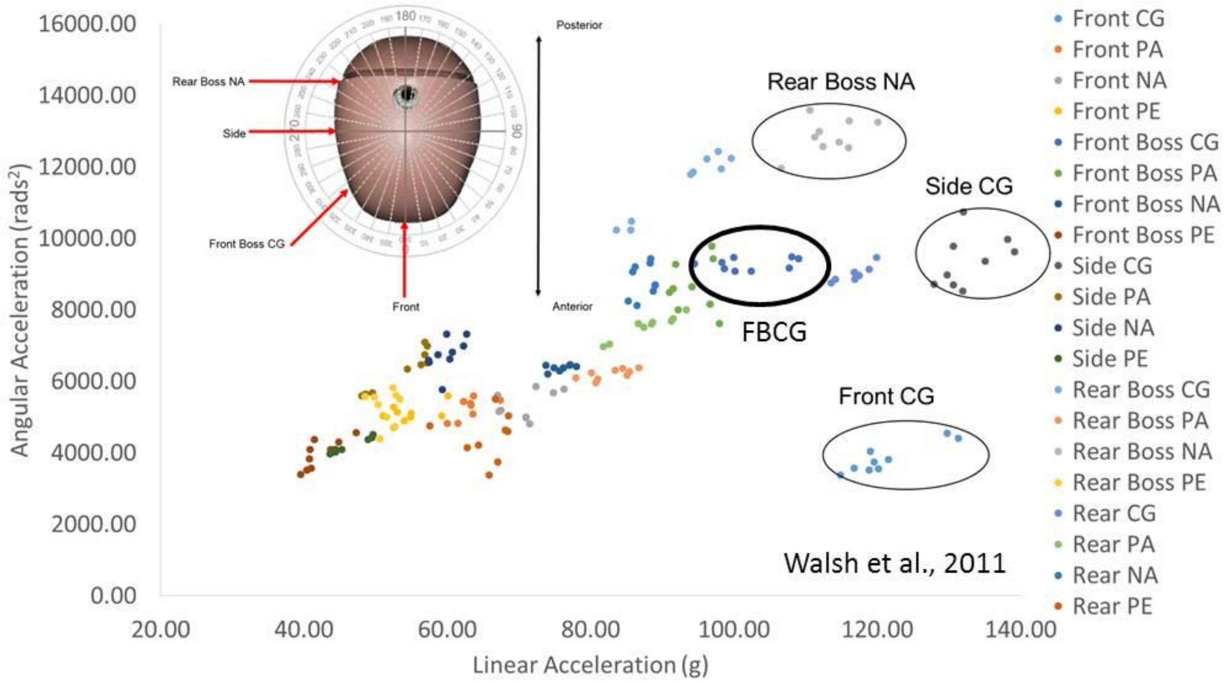


Figure 3-7. The effect of impact location on head dynamic response (adapted from Walsh et al. 2011). FB: Front Boss, S: Side, RB: Rear Boss, R: Rear, CG: Centre gravity, PA: Positive azimuth, NA: Negative Azimutth.

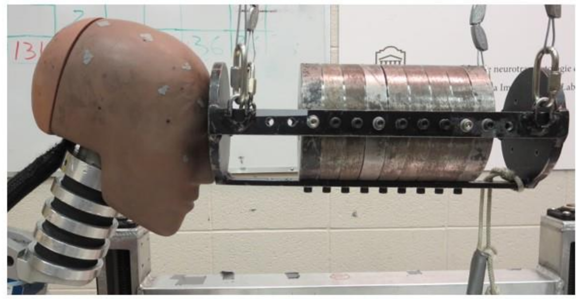
3.2 Summary

A summary of the levels of impact velocity, compliance, mass, and locations for falls and collisions used in this dissertation are illustrated in Figure 3-8. Each impact condition was tested for 3 consecutive trials resulting in 360 impacts collected for the fall test set up and 528 impacts for the collision test set up, totalling 888 impacts collected in this dissertation.

Falls

Velocity	1.5, 3.0, 4.5, 6.0 m/s
Compliance	99.3, 17.8, 3.0
Mass	3.5, 3.7, 4.5 kg
Location	Front, Front Boss, Side, Rear Boss

Collisions



Velocity	1.5, 3.0, 4.5, 6.0 m/s
Compliance	99.3, 17.8, 3.0
Mass	3, 9, 15, 21 kg
Location	Front, Front Boss, Side, Rear Boss

Figure 3-8. Summary of fall (left) and collision (right) impact velocities, compliances, masses, and locations examined in this research.

3.2.1 Research Design – Falls

Independent Variables (4):

A. Impact Location:
   - A_1 = Front PE
   - A_2 = Side CG
   - A_3 =Front Boss CG
   - A_4 =Rear Boss NA

B. Impact Velocity
   - B_1 =1.5 m/s
   - B_2 =3 m/s
   - B_3 =4.5 m/s
   - B_4 =6m/s

C. Impact Mass
   - C_1 = 3.5 kg Child Headform
   - C_2 = 3.7kg 5th Female Headform
   - C_3 = 4.5 kg 50th Male Headform

D. Impact Compliance
   - D_1 = Low (5ms)
   - D_2 = Medium (15ms)

- D_3 = High (25ms)

Dependent Variables (3)

- Peak resultant linear acceleration (g)
- Peak resultant angular acceleration (krad/s^2)
- Maximum principal strain

Research Design per Compliance:

[$AxBxCxD_1$]:

	C_1D_1	C_2D_1	C_3D_1
A_1B_1	$A_1B_1C_1D_1$	$A_1B_1C_2D_1$	$A_1B_1C_3D_1$
A_2B_1	$A_2B_1C_1D_1$	$A_2B_1\ C_2D_1$	$A_2B_1C_3D_1$
$A_3\ B_1$	$A_3\ B_1C_1D_1$	$A_3\ B_1\ C_2D_1$	$A_3\ B_1C_3D_1$
$A_4\ B_1$	$A_4\ B_1C_1D_1$	$A_4\ B_1\ C_2D_1$	$A_4\ B_1C_3D_1$
A_1B_2	$A_1B_2C_1D_1$	$A_1B_2\ C_2D_1$	$A_1B_2C_3D_1$
A_2B_2	$A_2B_2C_1D_1$	$A_2B_2\ C_2D_1$	$A_2B_2C_3D_1$
$A_3\ B_2$	$A_3\ B_2C_1D_1$	$A_3\ B_2\ C_2D_1$	$A_3\ B_2C_3D_1$
$A_4\ B_2$	$A_4\ B_2C_1D_1$	$A_4\ B_2\ C_2D_1$	$A_4\ B_2C_3D_1$

*** B3 and B4 excluded from D1 compliance as these conditions present a high risk of concussion and risk for more severe TBI including skull fracture and hematomas (Willinger & Baumgartner, 2003).**

[$AxBxCxD_2$]:

	C_1D_2	C_2D_2	C_3D_2
A_1B_1	$A_1B_1C_1D_2$	$A_1B_1C_2D_2$	$A_1B_1C_3D_2$
A_2B_1	$A_2B_1C_1D_2$	$A_2B_1\ C_2D_2$	$A_2B_1C_3D_2$
$A_3\ B_1$	$A_3\ B_1C_1D_2$	$A_3\ B_1\ C_2D_2$	$A_3\ B_1C_3D_2$
$A_4\ B_1$	$A_4\ B_1C_1D_2$	$A_4\ B_1\ C_2D_2$	$A_4\ B_1C_3D_2$
A_1B_2	$A_1B_2C_1D_2$	$A_1B_2\ C_2D_2$	$A_1B_2C_3D_2$
A_2B_2	$A_2B_2C_1D_2$	$A_2B_2\ C_2D_2$	$A_2B_2C_3D_2$
$A_3\ B_2$	$A_3\ B_2C_1D_2$	$A_3\ B_2\ C_2D_2$	$A_3\ B_2C_3D_2$
$A_4\ B_2$	$A_4\ B_2C_1D_2$	$A_4\ B_2\ C_2D_2$	$A_4\ B_2C_3D_2$
$A_1\ B_3$	$A_1\ B_3C_1D_2$	$A_1\ B_3\ C_2D_2$	$A_1\ B_3C_3D_2$
$A_2\ B_3$	$A_2\ B_3C_1D_2$	$A_2\ B_3\ C_2D_2$	$A_2\ B_3C_3D_2$
$A_3\ B_3$	$A_3\ B_3C_1D_2$	$A_3\ B_3\ C_2D_2$	$A_3\ B_3C_3D_2$
$A_4\ B_3$	$A_4\ B_3C_1D_2$	$A_4\ B_3\ C_2D_2$	$A_4\ B_3C_3D_2$
$A_1\ B_4$	$A_1\ B_4C_1D_2$	$A_1\ B_4\ C_2D_2$	$A_1\ B_4C_3D_2$
$A_2\ B_4$	$A_2\ B_4C_1D_2$	$A_2\ B_4\ C_2D_2$	$A_2\ B_4C_3D_2$
$A_3\ B_4$	$A_3\ B_4C_1D_2$	$A_3\ B_4\ C_2D_2$	$A_3\ B_4C_3D_2$
$A_4\ B_4$	$A_4\ B_4C_1D_2$	$A_4\ B_4\ C_2D_2$	$A_4\ B_4C_3D_2$

[$AxBxCxD_3$]:

	C_1D_3	C_2D_3	C_3D_3
A_1B_1	$A_1B_1C_1D_3$	$A_1B_1C_2D_3$	$A_1B_1C_3D_3$

A_2B_1	$A_2B_1C_1D_3$	$A_2B_1\ C_2D_3$	$A_2B_1C_3D_3$
$A_3\ B_1$	$A_3\ B_1C_1D_3$	$A_3\ B_1\ C_2D_3$	$A_3\ B_1C_3D_3$
$A_4\ B_1$	$A_4\ B_1C_1D_3$	$A_4\ B_1\ C_2D_3$	$A_4\ B_1C_3D_3$
A_1B_2	$A_1B_2C_1D_3$	$A_1B_2\ C_2D_3$	$A_1B_2C_3D_3$
A_2B_2	$A_2B_2C_1D_3$	$A_2B_2\ C_2D_3$	$A_2B_2C_3D_3$
$A_3\ B_2$	$A_3\ B_2C_1D_3$	$A_3\ B_2\ C_2D_3$	$A_3\ B_2C_3D_3$
$A_4\ B_2$	$A_4\ B_2C_1D_3$	$A_4\ B_2\ C_2D_3$	$A_4\ B_2C_3D_3$
$A_1\ B_3$	$A_1\ B_3C_1D_3$	$A_1\ B_3\ C_2D_3$	$A_1\ B_3C_3D_3$
$A_2\ B_3$	$A_2\ B_3C_1D_3$	$A_2\ B_3\ C_2D_3$	$A_2\ B_3C_3D_3$
$A_3\ B_3$	$A_3\ B_3C_1D_3$	$A_3\ B_3\ C_2D_3$	$A_3\ B_3C_3D_3$
$A_4\ B_3$	$A_4\ B_3C_1D_3$	$A_4\ B_3\ C_2D_3$	$A_4\ B_3C_3D_3$
$A_1\ B_4$	$A_1\ B_4C_1D_3$	$A_1\ B_4\ C_2D_3$	$A_1\ B_4C_3D_3$
$A_2\ B_4$	$A_2\ B_4C_1D_2$	$A_2\ B_4\ C_2D_2$	$A_2\ B_4C_3D_3$
$A_3\ B_4$	$A_3\ B_4C_1D_2$	$A_3\ B_4\ C_2D_2$	$A_3\ B_4C_3D_3$
$A_4\ B_4$	$A_4\ B_4C_1D_2$	$A_4\ B_4\ C_2D_2$	$A_4\ B_4C_3D_3$

D1:24 conditions x 3 trials = 72
D2:48 conditions x 3 trials = 144
D3:48 conditions x3 trials = 144
<u>Total Impacts: 360</u>

3.2.2 Research Design – Collisions

Independent Variables (4):

A. Impact Location:
    - A_1 = Front PE
    - A_2 = Side CG
    - A_3 =Front Boss CG
    - A_4 =Rear Boss NA
B. Impact Velocity
    - B_1 =1.5 m/s
    - B_2 =3 m/s
    - B_3 =4.5 m/s
    - B_4 =6m/s
C. Impact Mass
    - C_1 = 3kg
    - C_2 = 9kg
    - C_3 = 15kg
    - C_4 = 21kg
D. Impact Compliance
    - D_1 = Low (5ms)
    - D_2 = Medium (15ms)
    - D_3 = High (25ms)

Dependent Variables (3)

- Peak resultant linear acceleration (g)
- Peak resultant angular acceleration (krad/s^2)
- Maximum principal strain

Research Design per Compliance:

[AxBxCxD$_1$]:

	C_1D_1	C_2D_1	C_3D_1	C_4D_1
A_1B_1	$A_1B_1C_1D_1$	$A_1B_1C_2D_1$	$A_1B_1C_3D_1$	$A_1B_1C_4D_1$
A_2B_1	$A_2B_1C_1D_1$	$A_2B_1\ C_2D_1$	$A_2B_1C_3D_1$	$A_2B_1C_4D_1$
$A_3\ B_1$	$A_3\ B_1C_1D_1$	$A_3\ B_1\ C_2D_1$	$A_3\ B_1C_3D_1$	$A_3\ B_1C_4D_1$
$A_4\ B_1$	$A_4\ B_1C_1D_1$	$A_4\ B_1\ C_2D_1$	$A_4\ B_1C_3D_1$	$A_4\ B_1C_4D_1$
A_1B_2	$A_1B_2C_1D_1$	$A_1B_2\ C_2D_1$	$A_1B_2C_3D_1$	$A_1B_2C_4D_1$
A_2B_2	$A_2B_2C_1D_1$	$A_2B_2\ C_2D_1$	$A_2B_2C_3D_1$	$A_2B_2C_4D_1$
$A_3\ B_2$	$A_3\ B_2C_1D_1$	$A_3\ B_2\ C_2D_1$	$A_3\ B_2C_3D_1$	$A_3\ B_2C_4D_1$
$A_4\ B_2$	$A_4\ B_2C_1D_1$	$A_4\ B_2\ C_2D_1$	$A_4\ B_2C_3D_1$	$A_4\ B_2C_4D_1$
$A_1\ B_3$	$A_1\ B_3C_1D_1$	$A_1\ B_3\ C_2D_1$	$A_1\ B_3C_3D_1$	$A_1\ B_3C_4D_1$
$A_2\ B_3$	$A_2\ B_3C_1D_1$	$A_2\ B_3\ C_2D_1$	$A_2\ B_3C_3D_1$	$A_2\ B_3C_4D_1$
$A_3\ B_3$	$A_3\ B_3C_1D_1$	$A_3\ B_3\ C_2D_1$	$A_3\ B_3C_3D_1$	$A_3\ B_3C_4D_1$
$A_4\ B_3$	$A_4\ B_3C_1D_1$	$A_4\ B_3\ C_2D_1$	$A_4\ B_3C_3D_1$	$A_4\ B_3C_4D_1$

*** B4 excluded from D1 compliance as these conditions present a high risk of concussion and risk for more severe TBI including skull fracture and hematomas (Willinger & Baumgartner, 2003).**

[AxBxCxD$_2$]:

	C_1D_2	C_2D_2	C_3D_2	C_4D_2
A_1B_1	$A_1B_1C_1D_2$	$A_1B_1C_2D_2$	$A_1B_1C_3D_2$	$A_1B_1C_4D_2$
A_2B_1	$A_2B_1C_1D_2$	$A_2B_1\ C_2D_2$	$A_2B_1C_3D_2$	$A_2B_1C_4D_2$
$A_3\ B_1$	$A_3\ B_1C_1D_2$	$A_3\ B_1\ C_2D_2$	$A_3\ B_1C_3D_2$	$A_3\ B_1C_4D_2$
$A_4\ B_1$	$A_4\ B_1C_1D_2$	$A_4\ B_1\ C_2D_2$	$A_4\ B_1C_3D_2$	$A_4\ B_1C_4D_2$
A_1B_2	$A_1B_2C_1D_2$	$A_1B_2\ C_2D_2$	$A_1B_2C_3D_2$	$A_1B_2C_4D_2$
A_2B_2	$A_2B_2C_1D_2$	$A_2B_2\ C_2D_2$	$A_2B_2C_3D_2$	$A_2B_2C_4D_2$
$A_3\ B_2$	$A_3\ B_2C_1D_2$	$A_3\ B_2\ C_2D_2$	$A_3\ B_2C_3D_2$	$A_3\ B_2C_4D_2$
$A_4\ B_2$	$A_4\ B_2C_1D_2$	$A_4\ B_2\ C_2D_2$	$A_4\ B_2C_3D_2$	$A_4\ B_2C_4D_2$
$A_1\ B_3$	$A_1\ B_3C_1D_2$	$A_1\ B_3\ C_2D_2$	$A_1\ B_3C_3D_2$	$A_1\ B_3C_4D_2$
$A_2\ B_3$	$A_2\ B_3C_1D_2$	$A_2\ B_3\ C_2D_2$	$A_2\ B_3C_3D_2$	$A_2\ B_3C_4D_2$
$A_3\ B_3$	$A_3\ B_3C_1D_2$	$A_3\ B_3\ C_2D_2$	$A_3\ B_3C_3D_2$	$A_3\ B_3C_4D_2$
$A_4\ B_3$	$A_4\ B_3C_1D_2$	$A_4\ B_3\ C_2D_2$	$A_4\ B_3C_3D_2$	$A_4\ B_3C_4D_2$
$A_1\ B_4$	$A_1\ B_4C_1D_2$	$A_1\ B_4\ C_2D_2$	$A_1\ B_4C_3D_2$	$A_1\ B_4C_4D_2$
$A_2\ B_4$	$A_2\ B_4C_1D_2$	$A_2\ B_4\ C_2D_2$	$A_2\ B_4C_3D_2$	$A_2\ B_4C_4D_2$
$A_3\ B_4$	$A_3\ B_4C_1D_2$	$A_3\ B_4\ C_2D_2$	$A_3\ B_4C_3D_2$	$A_3\ B_4C_4D_2$
$A_4\ B_4$	$A_4\ B_4C_1D_2$	$A_4\ B_4\ C_2D_2$	$A_4\ B_4C_3D_2$	$A_4\ B_4C_4D_2$

[$AxBxCxD_3$]:

	C_1D_3	C_2D_3	C_3D_3	C_4D_3
A_1B_1	$A_1B_1C_1D_3$	$A_1B_1C_2D_3$	$A_1B_1C_3D_3$	$A_1B_1C_4D_3$
A_2B_1	$A_2B_1C_1D_3$	$A_2B_1 C_2D_3$	$A_2B_1C_3D_3$	$A_2B_1C_4D_3$
$A_3 B_1$	$A_3 B_1C_1D_3$	$A_3 B_1 C_2D_3$	$A_3 B_1C_3D_3$	$A_3 B_1C_4D_3$
$A_4 B_1$	$A_4 B_1C_1D_3$	$A_4 B_1 C_2D_3$	$A_4 B_1C_3D_3$	$A_4 B_1C_4D_3$
A_1B_2	$A_1B_2C_1D_3$	$A_1B_2 C_2D_3$	$A_1B_2C_3D_3$	$A_1B_2C_4D_3$
A_2B_2	$A_2B_2C_1D_3$	$A_2B_2 C_2D_3$	$A_2B_2C_3D_3$	$A_2B_2C_4D_3$
$A_3 B_2$	$A_3 B_2C_1D_3$	$A_3 B_2 C_2D_3$	$A_3 B_2C_3D_3$	$A_3 B_2C_4D_3$
$A_4 B_2$	$A_4 B_2C_1D_3$	$A_4 B_2 C_2D_3$	$A_4 B_2C_3D_3$	$A_4 B_2C_4D_3$
$A_1 B_3$	$A_1 B_3C_1D_3$	$A_1 B_3 C_2D_3$	$A_1 B_3C_3D_3$	$A_1 B_3C_4D_3$
$A_2 B_3$	$A_2 B_3C_1D_3$	$A_2 B_3 C_2D_3$	$A_2 B_3C_3D_3$	$A_2 B_3C_4D_3$
$A_3 B_3$	$A_3 B_3C_1D_3$	$A_3 B_3 C_2D_3$	$A_3 B_3C_3D_3$	$A_3 B_3C_4D_3$
$A_4 B_3$	$A_4 B_3C_1D_3$	$A_4 B_3 C_2D_3$	$A_4 B_3C_3D_3$	$A_4 B_3C_4D_3$
$A_1 B_4$	$A_1 B_4C_1D_3$	$A_1 B_4 C_2D_3$	$A_1 B_4C_3D_3$	$A_1 B_4C_4D_3$
$A_2 B_4$	$A_2 B_4C_1D_3$	$A_2 B_4 C_2D_3$	$A_2 B_4C_3D_3$	$A_2 B_4C_4D_3$
$A_3 B_4$	$A_3 B_4C_1D_3$	$A_3 B_4 C_2D_3$	$A_3 B_4C_3D_3$	$A_3 B_4C_4D_3$
$A_4 B_4$	$A_4 B_4C_1D_3$	$A_4 B_4 C_2D_3$	$A_4 B_4C_3D_3$	$A_4 B_4C_4D_3$

D1:48 conditions x 3 trials = 144
D2:64 conditions x 3 trials = 192
D3:64 conditions x3 trials = 192
Total Impacts: 528

II Effect of Impact Parameters

4 Mechanical Impact Parameters Causing Concussive Trauma to the Brain from Falls in Sport

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4.0 Abstract

Background: Falling accidents contribute to a large number of head injuries from impacts with rigid, immovable surfaces. In these events, a large amount of impact energy is transferred to the head. Factors that work to modulate the nature of this transfer include impact velocity, surface stiffness, impact location, and head size/mass. The purpose of this study was to describe how these mechanical parameters interact to cause concussion risk.

Methods: Mechanical parameters were informed from injury reconstructions and selected based on responses characteristic of sub-concussive and concussive injuries. These included impact velocities (1.5, 3.0, 4.5, 6.0 m/s), surfaces representing un-protected (bare head), protected (helmeted), and well-padded surface conditions (mat). Four impact sites were selected to cover frontal, sagittal, and combined-plane motions, as well as a non-centric rotationally dominant motion. Three Hybrid III headforms representing children, adolescent and adult head sizes were subjected to fall impacts using a monorail drop tower. Head accelerations were measured from the drop tower, then input into a finite element model, scaled to the appropriate sized brain, to obtain intracranial strains.

Results: Step-wise multiple linear regressions revealed that surface stiffness was the strongest predictor for head acceleration; while impact velocity was predictive of intracranial strain. Head size/mass was least influential to overall increases in response, however smaller headforms were associated with higher values of acceleration and strain. The data was subject to risk assessment guided by injury level responses determined from the literature and design limitations for sport related falls are proposed. Head acceleration-strain relationships indicate that linear acceleration best predicted strain under low and medium compliance, and angular acceleration for high compliance.

Conclusions: The most effective strategies to reduce concussive trauma from falls in sport should prioritize softening impact surfaces and minimizing impact velocities by adding protective layers and teaching safe falling techniques. Smaller headforms produced higher responses demonstrating the need for age and size appropriate definitions of risk for sport falls. Design limits in sport environments and relevant acceleration variables used to characterize risk are proposed.

4.1 Introduction

Sport related falls are a major contributor to the number of hospital visits for children, adolescents, and adults, most of which are preventable if principles of biomechanics are appropriately developed to guide safety design parameters (Cusimano et al., 2013; Daneshvar et al., 2011; Love et al., 2009; Ommaya et al., 1994; Thompson et al., 2006; Thurman et al., 1998). In a fall, the head collides with an immovable surface, resulting in a bulk energy transfer to the head and brain. Mechanical impact parameters that modulate this energy transfer include impact velocity, surface compliance, head mass, and impact location (Denny-Brown & Russell, 1941). Automatic reflexes, such as putting out an arm, or bending the knees and hips are often used as strategies to break the fall to minimize the impact velocity, but are only effective if the fall is anticipated and the individual has enough time to react.

During an impact, the head experiences linear and angular acceleration. Linear acceleration has been correlated with the onset of intracranial pressures proposed to result in dynamic stresses and strains responsible for concussion signs and symptoms (Gross, 1958; Gurdjian et al., 1953; Gurdjian et al., 1963; Gurdjian et al., 1954). At damaging levels of linear acceleration, skull fracture and contusions, have been well documented in experimental animal research, noting the focal nature of these types of injuries (Unterharnscheidt, 1972). Angular acceleration has been identified as an important predictor in the spectrum of diffuse brain injuries, from concussion to diffuse axonal injury, and has been proposed as the primary mechanism of brain tissue injury via shear strain modes on the neural tissues (Gennarelli et al., 1998; Holbourn, 1943; Margulies & Thibault, 1992; Ommaya & Gennarelli, 1974a). The link between linear and angular acceleration with tissue stresses and strains and clinical outcomes (transient concussion, persistent symptoms, loss of consciousness) have played an important role in setting injury criterion limits to guide injury prevention efforts (Gennarelli & Thibault, 1982; Kleiven, 2007; Margulies & Thibault, 1992; Oeur et al., 2015; Ommaya & Gennarelli, 1974b; Patton et al., 2015; Rowson et al., 2012; Willinger & Baumgartner, 2003; Zhang et al., 2004). Connecting these injury magnitudes with specific levels of mechanical impact parameters provides a road map to identifying at what levels these parameters contribute to risk and when. This supports the setting of safety limits within sport environments that reduce injury risk. The goal of this research was to map out the effects and interactions of impact velocity, head size, impact location, and surface compliance on linear and angular accelerations, and strain, to identify which variables are the most influential for increasing concussion risk. A secondary objective was to identify which aspect of kinematic response (peak or duration) was most influential on strain.

4.2 Materials & Methods

Impact variables were selected to capture the range of possible conditions and responses characteristic of concussive injuries for falls in sport (Oeur & Hoshizaki, 2016). Simplified fall events were conducted using a monorail drop tower using a test-set up consistent with previous reconstruction studies examining fall injuries (Oeur et al., 2015; Post et al., 2015). To vary impact mass, three different sized Hybrid III headforms were used in the tests (Humanetics Innovative Solutions, Michigan, USA): 1) 6-year old child, 2) 5th percentile female, and 3) 50th percentile adult male to represent head circumferences for children (5-8 years old), adolescents (11-16 years old), and adults (>18 years old) (Roche et al., 1987). These headforms were tested in a combination with non-directional neckforms (University of Ottawa, Ottawa, Canada) to limit the unknown effects of standard neckforms designed for high energy car crash research and comprised of alternating steel and butyl rubber discs held together using a standard Hybrid III neck cable (Kang et al., 2005; Walsh et al., 2012). Headform and neckform dimensions are presented in Table 2-3 and Figure 2-3.

4.1.1 Impact Variables

Impact velocities were selected to cover the range of fall injuries reported in the literature from injury reconstructions and full-body anthropomorphic test device (ATD) drop tests: 1.5, 3.0, 4.5, and 6.0 m/s (Doorly et al., 2005; Hajiaghamemar et al., 2015; Pellman et al., 2003). These velocities cover impact responses from no injury (sub-concussion) to clinically defined concussive injuries (Oeur et al., 2015). The lowest velocity represented the best-case scenario where the subject was able to break the fall; whereas the highest velocity represented the worst-case scenario where the subject was unable to protect themselves. Compliance describes the overall stiffness of the impact, with low compliant, rigid materials transferring impact energy rapidly; and high compliant materials able to attenuate energy (Gurdjian et al., 1966). Three levels of compliance were tested and compared: low, to represent unprotected (bare head); medium, to represent protected conditions (i.e a helmet); and high, to represent well-padded conditions (i.e. gym mat). Figure 3-3 illustrates resultant linear acceleration histories of a bare head impact onto ice, a helmeted fall, and impact onto a gymnastic mat. Surrogate surfaces used were steel (low compliance), 0.025 m vinyl nitrile foam (medium compliance), and 0.067 m Rubatek Rubber foam (high compliance) with Shore-A ASTM stiffness values of 99.3, 17.8, and 3.0, respectively. Four unique head impact locations were selected based on a study conducted by Walsh et al. (2011), to cover motion about different axes: front (sagittal plane), front boss (combined sagittal and frontal planes), side (frontal plane), and rear boss location with a negative 45° rotation, causing a rotationally dominant response (Figure 4-1).

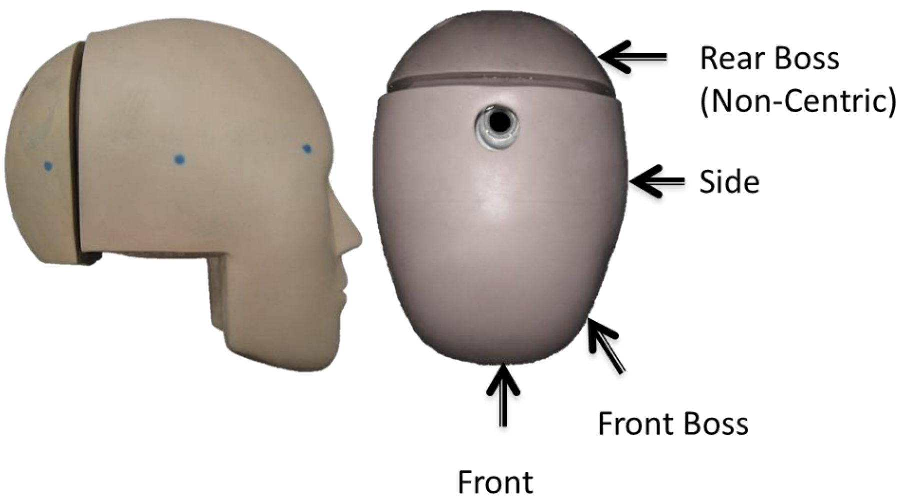


Figure 4-1. Head impact locations.

4.1.2 Data Collection

Headforms were equipped with nine Endevco 7264C-2KTZ-2-300 linear accelerometers (Endevco 7264C-2KTZ-2-300, San Juan Capistrano, CA) arranged in a 3×2×2×2 array to capture linear and angular accelerations according to the following (Padgaonkar et al., 1975):

$$\alpha_x = \frac{a_{zS} - a_{zC}}{2S} - \frac{a_{yT} - a_{yC}}{2T}$$

$$\alpha_y = \frac{a_{xT} - a_{xC}}{2T} - \frac{a_{zF} - a_{zC}}{2F}$$

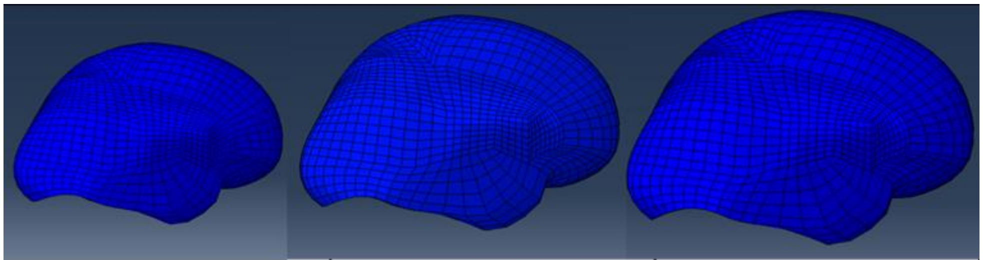
$$\alpha_z = \frac{a_{yF} - a_{yC}}{2F} - \frac{a_{xS} - a_{xC}}{2S}$$

Where ‘α’ denotes angular accelerations calculated from linear accelerations, ‘a’, for accelerometers that are positioned along the orthogonal represented by ‘S’, ‘T’, and

‘F’. For the child and adolescent sized headforms, distances for S, T and F are 0.04, 0.03, and 0.05 m respectively. For the adult headform, the distances were 0.05. 0.08, and 0.05 m respectively. Accelerometers captured at 20 kHz and were processed according to SAE J211 standards for head impact data.

4.1.3 Finite Element Analysis

To examine the influence of external impact parameters on peak strain, the headform linear and angular acceleration histories were input into the University College Dublin Brain Trauma Model (UCDBTM) (Horgan & Gilchrist, 2003, 2004). The UCDBTM is a finite element (FE) model of the human brain representing the adult brain and its constituents (Table 2-5). The material properties were derived from animal and cadaveric tissue tests and are consistent with other FE models (Willinger et al., 1995; Zhou et al., 1995). The dynamic model response has been compared with intracranial pressure, skull-brain displacement, and head and brain acceleration data from cadaver studies (Hardy et al., 2001; Nahum et al., 1977; Trosseille et al., 1992), which includes a parametric study of material definitions to determine the most suitable properties (Horgan & Gilchrist, 2003). The UCDBTM has also been used to estimate intracranial tissue response to traumatic brain injuries from accident reconstructions resulting in brain lesions and concussive injuries resulting in stresses and strains consistent with the literature (Doorly et al., 2005; Hoshizaki et al., 2016; Oeur et al., 2015; Post et al., 2014; Post et al., 2015). In total, the brain model was comprised of 26,000 elements. Element count was verified to examine the integrity of the solution based on aspect ratio. These analyses were run to examine the distortion of the elements throughout the simulation time, which resulted in the exclusion of 217 elements. The UCDBTM was used to analyse head kinematics collected from the adult Hybrid III headform. Head accelerations collected using the child and adolescent headforms were input to scaled versions of the original UCDBTM, consistent with child and adolescent brain dimensions reported in the literature (Figure 4-2) (Kleiven & von Holst, 2002; Lee et al., 2005; Lenroot & Giedd, 2006). Maximum principal strain (MPS) in the cerebrum was obtained from all simulations, as this has been proposed as an important predictor of concussive injury (King et al., 2003; Kleiven, 2007; Margulies & Thibault, 1992; Ommaya & Gennarelli, 1974b; Patton et al., 2013; Willinger & Baumgartner, 2003).



	Child	Adolescent	Adult
X	118.7 mm	122.4 mm	131.6 mm
Y	155.1 mm	160.9 mm	173.0 mm
Z	117.8 mm	121.7 mm	130.9 mm

Figure 4-2. Dimensions of the scaled UCDBTM for the child and adolescent brains, and based model used for the adult brain.

4.1.4 <u>Statistical Analysis</u>

Step-wise multiple linear regression analyses were conducted for each impact site to determine the most influential impact parameters (velocity, mass, compliance) on dependent variable (peak resultant linear and angular acceleration, and strain). Step-wise linear regression adds independent variables to the model based on the partial correlation matrix. Highly correlated variables will be added to the model first, followed by the next highly correlated variable. It has been suggested that a minimum of 10 subjects to be used per predictor variable (Kruskal & Majors, 1989). Due to the nature of this study, no subjects were used. If unique impact conditions can be considered as a single subject, then this study includes 24 unique impact conditions for low compliance, and 48 conditions each for medium and high compliance. For statistically significant predictive models that included the highest number of parameters, standard beta coefficients were reported for those parameters in order to evaluate the relative effects of each parameter on the dependent variable. Beta coefficients indicate the degree of predictability of the parameter on increases to the dependent variable. Furthermore, one-way ANOVAs and Tukey tests were conducted to identify impact conditions that significantly produced concussive level responses. Step-wise multiple linear regression analyses were also run on peak linear and angular accelerations and their respective durations (ms) to determine which aspect most influenced strain for each level of compliance. Statistical tests were conducted using IBM SPSS Statistics V 22.0 (Armonk, New York, USA) with significance accepted at $p < 0.05$.

4.3 Results

4.1.5 Effect of Impact Parameters

Surface compliance had the highest coefficients for peak linear and angular accelerations (0.80 to 0.90), followed by impact velocity (0.22 to 0.72). These trends were consistent across impact site (Table 4-1). Peak maximum principal strain was most influenced by velocity (0.86 to 0.99) and compliance (0.50 to 0.75) was relatively less influential. Headform mass (size) had the least influence on peak response values and was only a significant contributor for half the conditions (-0.07 to -0.19), however the negative sign indicates smaller head sizes produce higher responses.

Table 4-1. Multiple Linear Regression Results: Standard Beta Coefficients (SBC) for significant regression models at $p<0.05$.

	Front	**Front Boss**	**Side**	**Rear Boss (NA)**
Linear Acceleration (g)	$\beta_{Compliance}$ (0.85) $\beta_{Velocity}$ (0.67) $R^2 = 0.762$ $F = 139.114$	$\beta_{Compliance}$ (0.80) $\beta_{Velocity}$ (0.72) $R^2 = 0.742$ $F = 125.097$	$\beta_{Compliance}$ (0.90) $\beta_{Velocity}$ (0.41) β_{Mass} (-0.12) $R^2 = 0.735$ $F = 79.699$	$\beta_{Compliance}$ (0.86) $\beta_{Velocity}$ (0.58) β_{Mass} (-0.11) $R^2 = 0.733$ $F = 78.868$
Angular Acceleration (krad/s^2)	$\beta_{Compliance}$ (0.84) $\beta_{Velocity}$ (0.61) $R^2 = 0.721$ $F = 112.242$	$\beta_{Compliance}$ (0.85) $\beta_{Velocity}$ (0.69) β_{Mass} (-0.14) $R^2 = 0.798$ $F = 113.372$	$\beta_{Compliance}$ (0.85) $\beta_{Velocity}$ (0.22) β_{Mass} (-0.17) $R^2 = 0.672$ $F = 58.678$	$\beta_{Compliance}$ (0.88) $\beta_{Velocity}$ (0.38) β_{Mass} (-0.19) $R^2 = 0.714$ $F = 71.504$
Max. Principle Strain	$\beta_{Velocity}$ (0.86) $\beta_{Compliance}$ (0.74) $R^2 = 0.837$ $F = 223.311$	$\beta_{Velocity}$ (0.99) $\beta_{Compliance}$ (0.50) $R^2 = 0.877$ $F = 309.317$	$\beta_{Velocity}$ (0.91) $\beta_{Compliance}$ (0.75) β_{Mass} (-0.07) $R^2 = 0.905$ $F = 273.040$	$\beta_{Velocity}$ (0.98) $\beta_{Compliance}$ (0.63) $R^2 = 0.912$ $F = 450.192$

4.1.6 Mechanical Contributors to Risk

The data are presented according to headform size to best represent the age and size-related risks and were evaluated against compliance-specific levels of peak linear acceleration (low: 275 g; medium: 120 g; and high: 30 g) and angular acceleration (low: 20.0 krad/s^2; medium: 7.5 krad/s^2; and high: 3.5 krad/s^2) associated with sports-related concussion established by Hoshizaki et al. (2016). Since compliance was less influential for strain, a 0.2 strain level was used for low, medium, and high compliant conditions (Figure 4-3). These values were only used to guide the discussion on risk prevention and not meant as an injury tolerance measure, and are also consistent with the range of injury reconstruction research for low compliant conditions: 103 – 283 g, 7.1 - 19.6 krad/s^2; medium: 82 – 119 g, 5.9-7.6 krad/s^2; high: 27.5 g, 3.3 krad/s^2, and strain: 0.19 - 0.432 (Frechede & McIntosh, 2009; Hoshizaki et al., 2016; Kleiven, 2007; McIntosh et al., 2014; Patton et al., 2013; Post et al., 2015; Zhang et al., 2004). Impact conditions intersecting injury lines guided *post hoc* one-way ANOVAs and Tukey tests for the effect of impact location at specific levels of compliance, velocity, and mass ($p<0.05$).

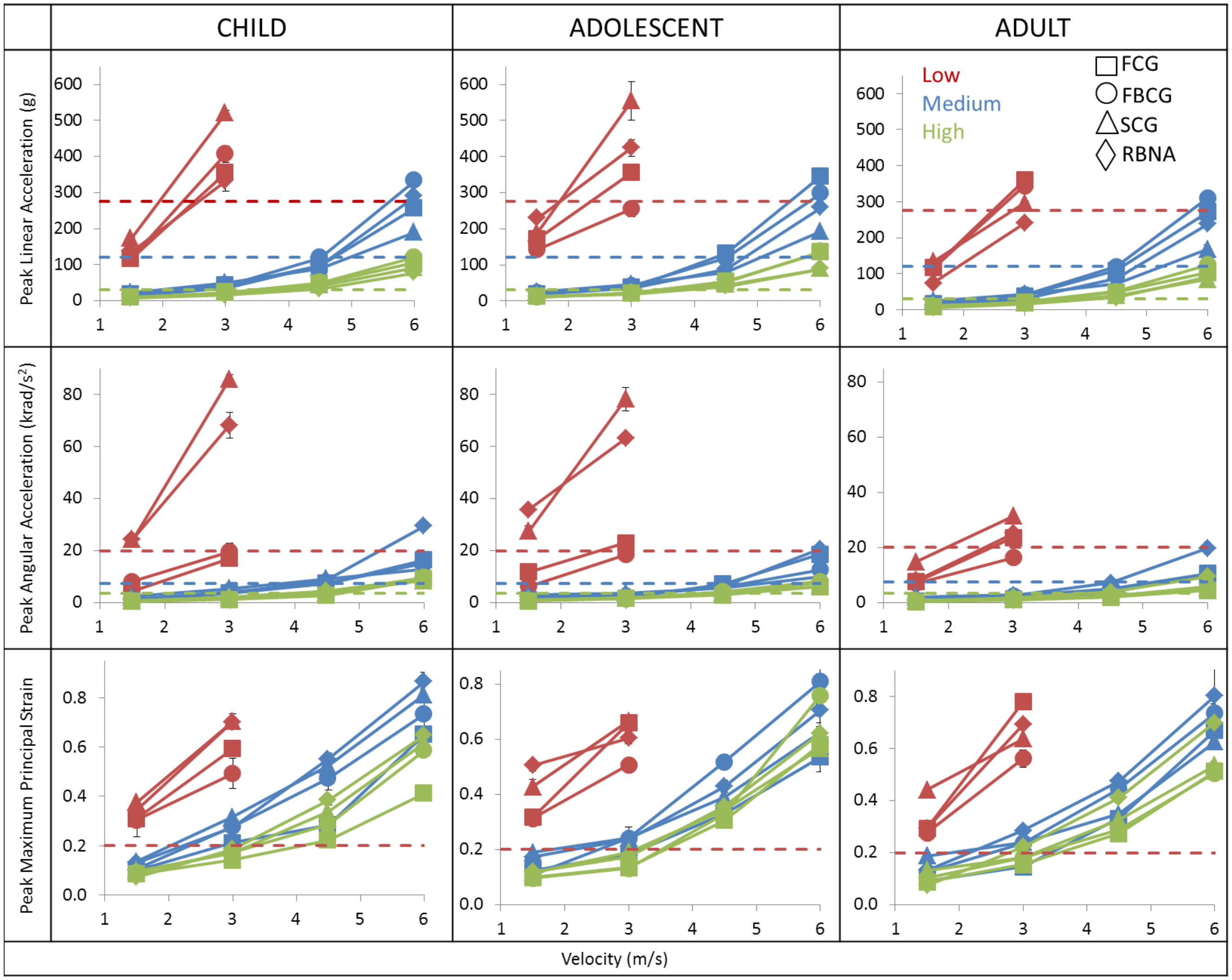


Figure 4-3. Peak resultant linear acceleration, angular acceleration, and FE strain are plotted against velocity for the child (left), adolescent (middle), and adult headforms (right). Dashed lines indicate reference values from Table 4 and low (red), medium (blue), and high (green) compliance conditions. Square: front, circle: front boss CG, triangle: side CG, and diamond: rear boss negative azimuth.

A summary of the specific levels of impact velocity and locations that created concussive levels responses for each headform size are summarized in Table 4-2. Child headform impacts onto low compliant surfaces presented with concussive level strains at 1.5 m/s with impact locations producing similar responses ($p=0.238$). Under medium compliance, strain indicated risk at 3.0 m/s with the side producing the highest values. For high compliance, all dependent variables indicated risk for concussion at 4.5 m/s with the front boss and rear boss showing the highest values. Low compliant impacts to the adolescent headform presented risk under angular acceleration and strain at 1.5 m/s for rear boss. Under medium compliance, strain presented risk at 3.0 m/s for all impact locations, however under high compliance, all dependent variables presented risk at 4.5 m/s. Adult headform impacts onto low compliance presented risk at 1.5 m/s with the side. Medium and high compliant impacts required 3.0 m/s to produce injury level strains, where the front boss and rear boss contributed to higher values.

Table 4-2. Summary of impact velocities and locations associated with concussive level responses for each headform size and compliance level.

	Low Compliance	**Medium Compliance**	**High Compliance**
Child Headform	1.5 m/s All locations	3.0 m/s Side	4.5 m/s Front Boss; Rear Boss
Adolescent Headform	1.5 m/s Rear Boss	3.0 m/s All locations	4.5 m/s All locations except Side
Adult Headform	1.5 m/s Side	3.0 m/s Front Boss	3.0 m/s Rear Boss

4.1.7 Kinematic-Strain Relationships

All kinematic-strain regressions were significant ($F = 85.085\text{-}975.364$, $p<0.001$), where a single peak head acceleration variable was reported in the first predictive model accounted for the largest amount variance (highest R^2). The addition of subsequent kinematic parameters contributed little to predictive equation with small R^2 changes (0.003-0.037). Peak resultant linear acceleration had the strongest relationships with strain for low and medium compliance, and angular acceleration for high compliance (Figure 4-4).

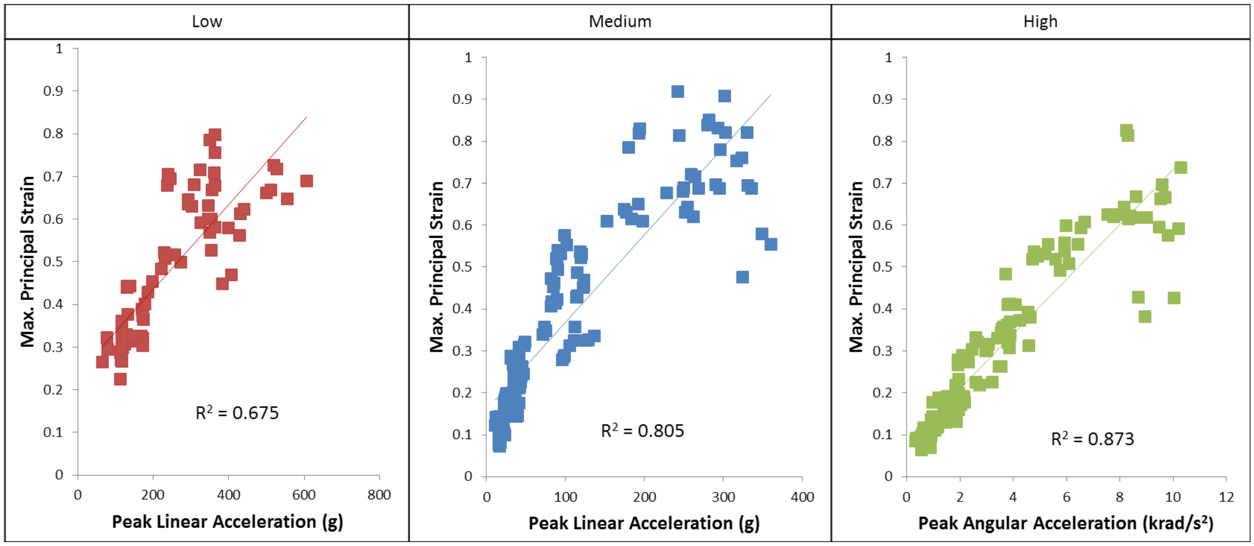


Figure 4-4. Peak head acceleration - strain relationships for low (left), medium (center), and high (right) compliance conditions

4.4 Discussion

This study identifies mechanically relevant risk factors for fall events in sport and offers two avenues for informing injury prevention. The first involves an analysis of the environmental condition to determine the specific levels of impact parameters that cause injury risk. Overall, a reduction in impact energy, by decreasing velocity and increasing compliance, is the most effective strategy for lowering head acceleration and strain from sport-related falls. Peak linear and angular acceleration were most influenced by surface compliance, with stiffer surfaces producing higher values. Gurdjian and colleagues (1966), who conducted cadaver drop tests with and without a helmet across varying levels of impact velocity. The authors demonstrated that tests with a helmet consistently lowered levels of peak linear acceleration in comparison to the bare head condition (Gurdjian et al., 1966). Our study further illustrates this phenomenon for angular acceleration and shows that adding compliance will similarly contribute to overall decreases in peak magnitudes. Risk management strategies for lowering risk of concussion from head acceleration entails identifying low compliant conditions in sport and children's play environments and adding protective layers (e.g., adding carpets and mats or wearing helmets in play areas) (Mohan et al., 1979).

Unlike head acceleration, increases to peak strain in the FE model was most sensitive to impact velocity on the headform. Strain values derived from FE take into account aspects of surface compliance as calculations are determined using the acceleration loading histories, which reflect compliance effects on loading curve shape, peak, slope, and pulse duration. Risk management strategies for minimizing impact velocity in an environment can be achieved by breaking the fall using pre-impact movements (e.g., by sticking out the hands and bending the knees or hips). Fall techniques are commonly taught as injury prevention strategies in judo and gymnastics, however the majority of accidental injuries are from unexpected falls in sport (Bak et al., 1994; Takeshi et al., 2013). Therefore, further management of impact velocity can be achieved through identifying the most frequent causes of fall related concussion in sport and managing the game rules and player conduct to decrease risk.

Head size contributed little to overall increases in head acceleration and strain based on the low beta coefficients in the regression results. The negative coefficients for mass (Table 4-1) demonstrate that smaller head sizes are associated with higher magnitudes of acceleration and strain than larger headforms. These findings are consistent with Ommaya et al. (2002) whom proposed that children may have a higher tolerance to acceleration injuries than adults based on the notion that the same level of input force will cause a smaller head mass to accelerate at higher rates than larger masses. Therefore, it would be inappropriate to use adult levels of injury tolerance and criteria to govern and dictate the protective needs for children. This data provides further support for age and size appropriate definitions of risk in sport. Due to the paucity of concussion injury data for children, risk levels from adult sport were used to evaluate children and adolescent impact responses. The specific levels of impact velocities and locations contributing to concussive level responses are summarized in Table 4-2 and highlight the incongruities between headform sizes. Falls onto low compliant surfaces presented concussion risk at 1.5 m/s, or a drop height of 0.15 m for all head sizes, suggesting rule and design limitations for sport related falls for rigid surfaces be below these parameters. Protective layers or helmets added in response to these conditions offsets the presentation of risk to 3.0 m/s or a 0.52 m fall height, similarly suggesting helmeted falls be below these parameters. This is consistent with reconstructions of adolescent ice hockey falls resulting in medically diagnosed concussions reported fall velocities occurring between 3.5 – 6 m/s (Oeur et al., 2015). For high compliance conditions, children and adolescents required 4.5 m/s to approach concussive level responses, but only 3.0 m/s was required for adults at the rear boss (non-centric) location. Throughout impact testing, headform impact locations were aligned with the centre of the impact surface and it is likely that the larger size and mass distributions of the adult headform resulted in a greater moment arm for the non-centric location contributing to the high strains observed at a lower velocity. This is a reflection of headform mass and velocity interactions that create conditions associated with increased risk and support the need to evaluate the effective performance range of the materials for different sized heads. There are limits as to the operating ranges of these energy absorbing materials at which a high impact velocity will cause compliant layers to bottom out rendering them no longer effective at energy management (Gimbel & Hoshizaki, 2008). Ideally, materials selected for these protective qualities should match the energy needs of the impact event, balancing out its ability to manage the impact taking into account mass-velocity interactions, the amount of energy transferred to the head, and resulting injury (Hoshizaki & Brien, 2004; Hoshizaki et al., 2014; Newman, 2002).

The second avenue informs on the relationship between head acceleration and calculated strain. Previous investigations have sought to identify correlations between head kinematic variables and brain response with the intent of identifying a single kinematic injury parameter that can be used as a target variable to expedite head protection and helmet developments that rely on physical impact tests (Forero Rueda et al., 2011; Ueno & Melvin, 1995). Overall, the first predictive equation consisted of a single kinematic parameter able to predict strain with R^2 values from 0.68 to 0.87. The inclusion of more kinematic parameters in subsequent equations resulted in minimal improvements in predictability (small R^2 change values). These findings suggest that there is robustness in the association between a single kinematic parameter and strain for compliance specific conditions in sport related falls. Previous investigations have reported that angular acceleration better associated with FE strain for America football head impacts (R^2Linear = 0.65; R^2Angular = 0.84) (Kleiven, 2007) and equestrian helmet impacts (R^2Linear = 0.47; R^2Angular = 0.7) than linear acceleration (Forero Rueda et al., 2011). Helmeted impacts are represented by the medium compliance level in this study, and while linear acceleration was shown to be the better predictor of strain (R^2 = 0.80), angular acceleration was only slightly lower (R^2 = 0.78). The study further presents linear acceleration as a better predictor of strain under low compliance (unprotected rigid impact conditions) and angular acceleration for well-padded, high compliant conditions.

This research was limited to the rigid humanoid headforms and the finite element models that are idealized representations of the complex human head and brain system. The head and neckforms were composed of steel, vinyl, and rubber necessary for enduring repeated, multiple impacts to elicit the impact phenomena studied. Standard Hybrid III neckforms are available, however these neckforms have a directional stiffness built in that was designed for cervical flexion and extension in car crash environments and may not be suitable for direct head impact testing (Kang et al., 2005; Mertz et al., 1989; Mertz & Patrick, 1971). Compliance was represented using flat surfaces neglecting the effects of local surface geometries that influence foam compression and head dynamic response (Spyrou et al., 2000). The finite element model was derived from CT scans and the material characteristics were taken from animal and cadaver specimens that do not reflect age-related geometries or material properties. In reality, the head and brain tissues exhibit rate dependent loading characteristics that are regionally and structurally specific and respond differently depending on the direction of that loading (Chatelin et al., 2010; Gennarelli et al., 1998; Prange & Margulies, 2002).

The data was interpreted according to concussion risk levels representing adult sport (Hoshizaki et al., 2016; Kleiven, 2007; McIntosh et al., 2014; Patton et al., 2013; Zhang et al., 2004) and share many similar inherent limitations common to head injury reconstruction techniques. There are a number of subject-specific characteristics and risk factors such as injury history, gender, age, and genetics that have been known to influence an individual's risk to concussion (Kutcher & Eckner, 2010). The specific values of head acceleration and strain used for risk assessment will change depending on the cohorts studied and techniques used however, the definition of increased injury risk was defined as higher values of response (linear and angular acceleration and strain). The originality of this research is in establishing relationships between mechanical impact parameters and the patterns of acceleration and strain response. Identification of these relationships provides a valuable road map for cause-and-effect parameters for guiding risk management strategies for falls in sport.

4.5 Conclusions

During falls, the most influential impact parameters on head acceleration and strain were surface stiffness and impact velocity, suggesting that the most effective strategies for risk reduction based on these dependent variables is to increase compliance and lower impact velocity. While mass is less important mechanically to overall increases in head acceleration and strain as indicated by smaller beta coefficients, smaller headform sizes result in higher responses and provides support for age and size appropriate risk definitions to govern the protective needs for children. Furthermore, interactions between headform mass and velocity created conditions where the larger adult Hybrid III headform resulted in increased angular accelerations and strain from non-centric impacts during the high compliance condition. It is hypothesized that the greater circumference of the headform resulted in an increased moment arm for an impact at the rear boss negative azimuth location in comparison to the smaller child and female headforms. This study demonstrates that the effective performance range of the materials needs to be evaluated for different sized heads to ensure that currently available protection does not increase risk. The data is evaluated according to concussive level responses from the literature, and design limits in sport environments as well as relevant acceleration variables used to characterize risk are proposed.

5 The Role of Impact Parameters on Concussion Risk from Collisions in Sports

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5.0 Abstract

Background/Aim: Collisions with the head are a common cause of concussions in sport. The nature of the collision plays a primary role in the ensuing injury risk. Therefore, the purpose of this study was to describe the effects of impact velocity, striking mass, surface compliance, and head impact location on head acceleration and strain to identify the most influential parameters and their interactions on prediction of concussion risk.

Methods: A variable-mass pendulum (3-21 kg) with three levels of compliance representing unprotected impacts, a protected (helmet-to-helmet) and a well-padded (i.e. shoulder pad) condition impacted a Hybrid III headform at four locations (front, front boss, side, rear boss) and four velocities (1.5, 3.0, 4.5, and 6.0 m/s). Hybrid III head accelerations were input into a finite element model to determine the associated levels of brain tissue strain.

Results: Multiple linear regression analysis revealed that surface compliance was the most influential for increases to head acceleration and impact velocity was the most influential for strain. While striking mass was the least important on response, mass-compliance interactions created conditions in which 9 kg, representing the effective mass of a punch, produced the highest risk.

Conclusion: The most effective strategies for reducing concussion risk is to increase compliance, such as adding protective layers and to minimize striking velocities between athletes. Mass-compliance interactions created conditions where the 9 kg mass, characteristic of a boxing punch produced increased responses that were greater than helmet to helmet collisions (15 kg), which emphasizes the importance of understanding the 'effective performance range' of protective devices.

5.1 Introduction

Contact and combat sports present a high risk of concussion where collisions with the head are the most common cause of injury (Daneshvar et al., 2011; Delaney et al., 2006; Hutchison et al., 2014; Tommasone & Valovich McLeod, 2006). In American football, ice hockey, football, and rugby, player-to-player collisions account for 70-90% of all concussions (Barnes et al., 1998; Cusimano et al., 2013; Hutchison et al., 2013; Pellman et al., 2004). In combat sports, such as boxing and mixed martial arts (MMA), nearly all contact involves delivering blows to the head (Hutchison et al., 2014; Karpman et al., 2016). A collision impact imparts a rapid impact force (<50ms) setting the head and brain in motion (Hodgson & Thomas, 1972). Linear acceleration of the head has been correlated with the onset of pressure gradients within the cranium (Gross, 1958). These pressure gradients have been proposed to result in dynamic stresses that affect the cerebrum and brainstem that are responsible for concussive signs and symptoms (Gurdjian et al., 1953; Gurdjian et al., 1963; Gurdjian et al., 1954). Angular acceleration has been proposed as an important kinematic measure, due to the brain's characteristically low shear modulus and its susceptibility to shear-strain injury modes on the tissue (Holbourn, 1945; Ommaya, 1968). Angular kinematics have been found to play important role in concussive outcomes in animals has been associated with the spectrum of diffuse brain injuries, from mild concussions to diffuse axonal injuries, where higher angular acceleration resulted in increased severity of injuries showing increased strains and poorer outcomes (Gennarelli et al., 1998; Gennarelli et al., 1971; Holbourn, 1943; Margulies & Thibault, 1992; Ommaya & Gennarelli, 1974a, 1974b).

Mechanical impact parameters that modulate the nature of the impact force include striking velocity, mass, compliance (stiffness) of the object, and head impact location (Denny-Brown & Russell, 1941; Gurdjian et al., 1966; Hodgson et al., 1983). In sport, mass and compliance characteristics are inherent properties of the striking object. Impact parameters associated with common collision events are summarized in Figure 3-6 along with impact velocities that have been reported to be associated with concussive events. These parameters are categorized into three levels of compliance, low, medium, and high to best represent unique event stiffness. Low compliant events have acceleration pulses lasting around 5 milliseconds (ms) and are generally from player collisions that involve no padding or protection, helmets or otherwise. Injuries in rugby and football typically fall into this category (McIntosh et al., 2000; Withnall et al., 2005). Missed headers in football can result in head-to-head or head-to-elbow collisions with opponents, while goalie dives can result in knee-to-head collisions with advancing players. Punches and kicks to the head from mixed martial arts (MMA) and Taekwondo are an exception, as they involve a layer of padding (head or hand), but due to the high energy nature of these events, the acceleration pulses are between 5-7 ms (Fife et al., 2012; Hoshizaki et al., 2016). Medium compliance encompasses 15 ms acceleration pulses and involves collisions between protected body parts and player helmets (Pellman et al., 2003; Zhang et al., 2004). Punches with boxing gloves, elbow impacts in ice hockey, and helmet-to-helmet collisions in American football constitute this category (Rousseau & Hoshizaki, 2015; Walilko et al., 2005). High compliance conditions involve well-padded surfaces onto player helmets where the combination of protective layers results in accelerations lasting 25 ms (Rousseau & Hoshizaki, 2015). Events unique to this compliance are extended elbow impacts in ice hockey and shoulder pad to helmet collisions common to hockey and American football (Hoshizaki et al., 2016). Despite the multitude of impact conditions that can cause concussions in sport, connecting characteristics of the impact event with the ensuing injury risk is necessary to determine which parameters contribute to creating concussion risk and when. Therefore, the purpose of this study was to examine the effects of impact velocity, surface compliance, striking mass, and head impact location on head acceleration and strain to identify the most influential parameter and their interactions on head and brain response. This information can be used to guide impact protection strategies specific to collision sports.

5.2 Methods

5.2.1 Impact Variables

Four impact masses, 3, 9, 15, and 21 kg were used to represent the range of striking masses common to sport collisions (Figure 3-6). Impact velocities associated with concussive events can range from 4-15 m/s for punches, elbow strikes, and helmet collisions (Pellman et al., 2003; Rousseau & Hoshizaki, 2015; Withnall et al., 2005). Four test velocities were selected, 1.5, 3.0, 4.5, and 6.0 m/s to examine the graded effects of momentum that shifts responses from no-injury to concussion. Three levels of compliance were used: steel (low), 0.025 m vinyl nitrile 602 foam (medium), and 0.067 m R338 Rubatek Rubber foam (high; Rubatek LLC, Virginia, USA), to establish 5 ms, 15 ms and 25 ms acceleration pulse durations, characteristic of unprotected impacts to bare head, protected (helmet), and well-padded shoulder collisions, respectively

(Hoshizaki et al., 2016; Hoshizaki et al., 2014). The ASTM Shore A stiffness values for each of these surfaces was 99.3, 17.8 and 3.0. Steel impacts were limited to 4.5 m/s since this velocity produced 400 g responses that were well beyond those for sport concussions (Hoshizaki et al., 2016; McIntosh et al., 2014). Impact locations included the front and side impacts to cover motion about the frontal and sagittal planes, front boss for combined planes, and a non-centric rear boss location to elicit a rotationally-dominant response (Figure 5-1) (Oeur & Hoshizaki, 2016; Walsh et al., 2011).

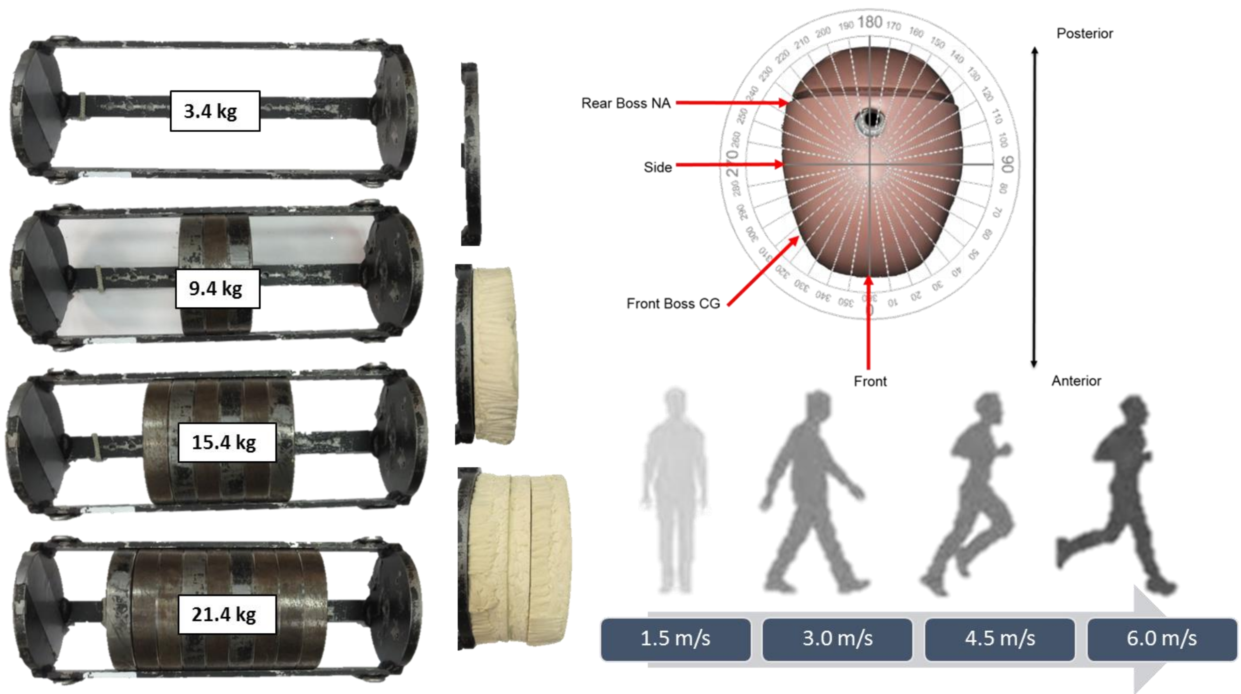

Figure 5-1. Levels of impact mass, impact compliance, impact location, and velocities demonstrated using the multi-mass pendulum and Hybrid III headform.

5.2.2 Equipment

A variable-mass pendulum as described by Karton et al. (2014) was used to deliver impacts to a head- and neckform assembly. The pendulum striking body is comprised of a steel frame (3.36 kg) fitted with circular steel weights to parameterize striking mass. An average male Hybrid III headform (Humanetics Innovative Solutions, Michigan, USA) in conjunction with a non-directional neckform (University of Ottawa, Canada) was instrumented with 9 linear accelerometers (Endevco 7264C-2KTZ-2-300, Meggitt Inc., San Juan Capistrano, CA), was used to capture three-dimensional linear and angular accelerations (Padgaonkar et al., 1975). Accelerations were recorded at 20 kHz and processed according to SAE J211 protocols for head impact data using a TDAS ProLab Module (Diversified Technical Systems, Inc., California, USA). Linear and angular accelerations histories were input into the University College Dublin Brain Trauma Model (UCDBTM) to obtain peak maximum principal strain. The finite element model was developed by Horgan and Gilchrist (2003, 2004) and represents an average male geometry, consisting of 26,000 elements with material properties consistent with other FE models in the literature (Table 2-5) (Willinger et al., 1995; Zhang et al., 2001; Zhou et al., 1995). While the current finite element model contains fewer elements (26, 000) than other models currently available (up to 300 000) (Zhang et al., 2001), this helped with faster computational times as a total of 528 impact trials were simulated in this experiment.

5.2.3 <u>Statistical Analysis</u>

Step-wise multiple linear regression analyses were conducted at each location to determine the most influential impact parameter on response by comparing standardized beta coefficients. A step-wise regression was conducted to ensure that the independent variables added to the model included those that were best correlated with the dependent variable first (based on the partial correlation matrix), followed by the next highly correlated variable. Multiple linear regression requires at least 10 subjects per predictor variable. While no subjects were used in this study, each unique impact condition can be considered a subject (Kruskal & Majors, 1989). Therefore, this study included 48 unique conditions for the low compliance, and 64 conditions each for the medium and high compliance. The relationship between head acceleration and strain were also evaluated using step-wise multiple linear regression analysis with peak resultant linear and angular acceleration and the respective durations (ms) as predictor variables and strain as the response variable. Statistical analyses were conducted using IBM SPSS Statistics V 22.0 (Armonk, New York, USA) and significance was accepted at an alpha level of 0.05.

5.3 Results

5.3.1 Evaluation of Impact Parameters

Impact compliance was the most influential for linear and angular acceleration across all locations with beta coefficients ranging from 0.77-0.86 (Table 5-1). Impact velocity was most influential on strain (0.70-0.86), except for side where compliance had a higher coefficient (0.79). The spread of strain values from compliance resulted in a wide range of responses at the side, in which low compliance had very large values and medium and high compliance had much lower values. When the data points were regressed, this resulted in higher beta coefficients for compliance at this location. Striking mass was the least influential (0.08-0.12) and was only a significant contributor half the time.

Table 5-1. Standard beta coefficients from step-wise multiple linear regression results for significant models that included the highest number of predictors (at p<0.05).

	Front	**Front Boss**	**Side**	**Rear Boss (NA)**
Linear Acceleration (g)	$\beta_{Compliance}$ (0.80) $\beta_{Velocity}$ (0.54) β_{Mass} (0.12) $R^2 = 0.758$ $F = 133.579$	$\beta_{Compliance}$ (0.78) $\beta_{Velocity}$ (0.52) β_{Mass} (0.12) $R^2 = 0.727$ $F = 113.374$	$\beta_{Compliance}$ (0.84) $\beta_{Velocity}$ (0.45) $R^2 = 0.747$ $F = 189.939$	$\beta_{Compliance}$ (0.78) $\beta_{Velocity}$ (0.55) β_{Mass} (0.12) $R^2 = 0.742$ $F = 122.456$
Angular Acceleration (krad/s^2)	$\beta_{Compliance}$ (0.77) $\beta_{Velocity}$ (0.55) β_{Mass} (0.16) $R^2 = 0.748$ $F = 126.386$	$\beta_{Compliance}$ (0.80) $\beta_{Velocity}$ (0.42) $R^2 = 0.676$ $F = 134.331$	$\beta_{Compliance}$ (0.86) $\beta_{Velocity}$ (0.33) $R^2 = 0.731$ $F = 174.992$	$\beta_{Compliance}$ (0.79) $\beta_{Velocity}$ (0.53) $R^2 = 0.736$ $F = 180.164$
Max. Principle Strain	$\beta_{Velocity}$ (0.70) $\beta_{Compliance}$ (0.67) $R^2 = 0.744$ $F = 186.965$	$\beta_{Velocity}$ (0.77) $\beta_{Compliance}$ (0.67) β_{Mass} (0.10) $R^2 = 0.831$ $F = 209.900$	$\beta_{Compliance}$ (0.79) $\beta_{Velocity}$ (0.66) β_{Mass} (0.08) $R^2 = 0.855$ $F = 250.844$	$\beta_{Velocity}$ (0.86) $\beta_{Compliance}$ (0.56) β_{Mass} (0.12) $R^2 = 0.872$ $F = 289.856$

5.3.2 Implications for Concussion Risk

Impact conditions contributing to concussive level peak linear acceleration, angular acceleration and strains were assessed according to each level of compliance; low: 275 g, 20 krad/s^2; medium: 120 g, 7.5 krad/s^2; high: 30 g, 3.5 krad/s^2 (Figure 5-2). A strain level of 0.2 was used to evaluate low, medium, and high compliance conditions for all FE results since compliance was less influential on strain. These values were selected to guide the discussion and since a large range of concussive response values have been reported in the literature: 27.5 – 283 g; 3.3 - 19.6 krad/s^2; 0.19-0.43 strain (Bain & Meaney, 2000; Frechede & McIntosh, 2009; Galbraith et al., 1993; Hoshizaki et al., 2016; Kleiven, 2007; McIntosh et al., 2000; Oeur et al., 2015; Patton et al., 2013; Zhang et al., 2004). For each cell in Figure 5-2, two-way ANOVAs (2 levels: striking mass and location) and Tukey *post hoc* tests were run at each velocity to identify specific conditions that contributed to increased risk (responses intersecting concussion lines).

This was done to identify at which levels of velocity, striking masses and impact locations caused higher values.

For low compliance, strain indicated risk at 1.5 m/s in which the side consistently produced the highest values. At the medium compliance, strain presented at 3.0 m/s with the rear boss location, and high compliance indicated injury at 4.5 m/s with linear acceleration and strain for all locations except the front. Interactions between mass and compliance show that 9 kg consistently produced the highest values from 3.0 - 4.5 m/s at all levels of compliance ($p<0.001$). While this effect disappeared at 6.0 m/s under high compliance, it remained at medium compliance for angular acceleration and strain with rear boss (non-centric) impacts.

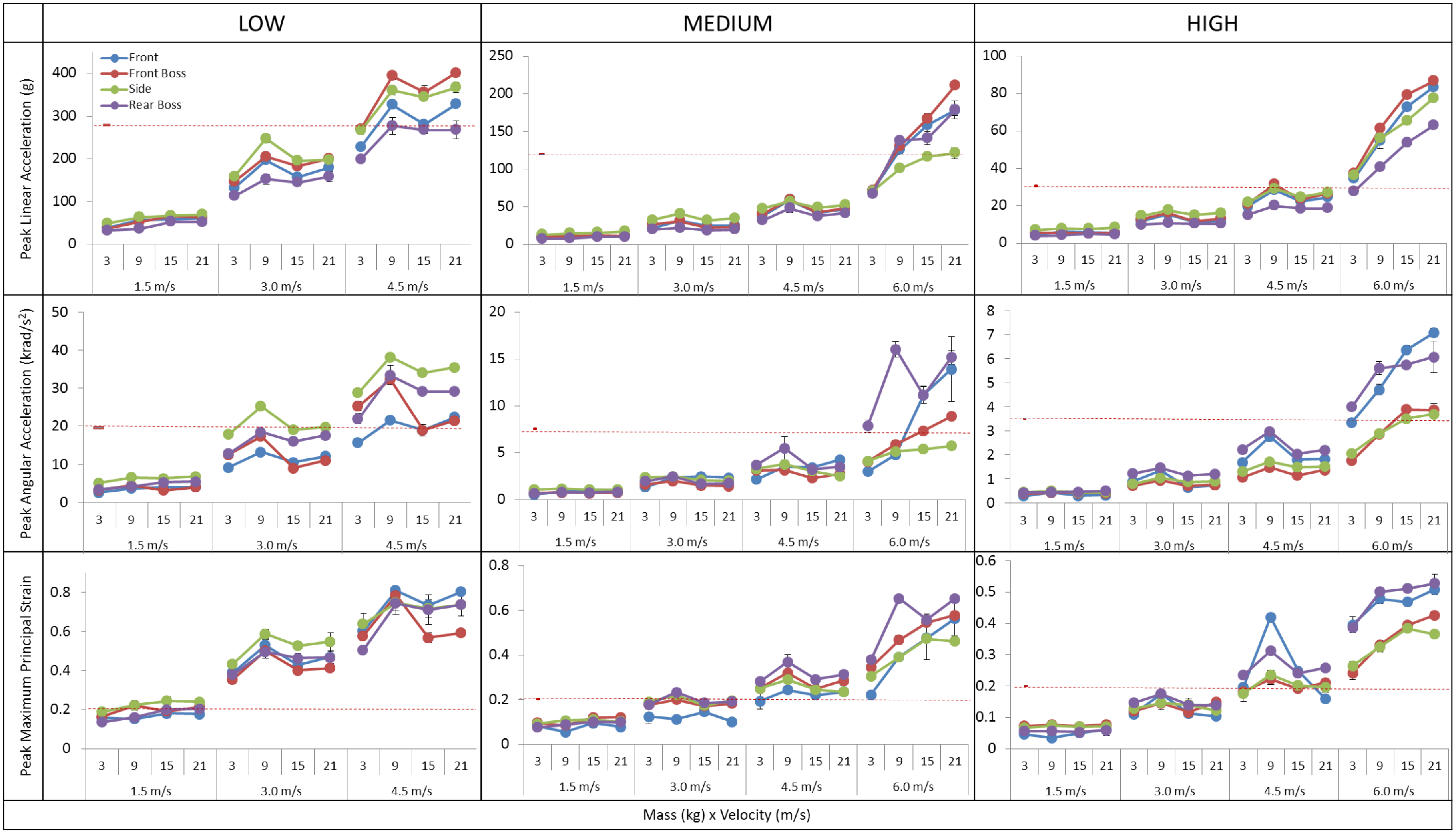

Figure 5-2. Low compliance (left), medium compliance (centre) and high compliance (right) plotted against mass (kg) and velocity (m/s) for peak resultant linear acceleration, angular acceleration, and strain, for each of the locations: front (blue), front boss (red), side (green), rear boss (purple). Dotted lines indicated concussive level responses established by Hoshizaki et al., 2016.

5.3.3 Head Acceleration-Strain Relationships

All head acceleration-strain relationships were found to be significant (F=431.401-1184.451, p<0.001) however the first predictive equation consisting of a single variable accounted for the most variance and subsequent variables had small increases in R^2 change (0.002-0.037). Linear acceleration best predicted strain for low and medium compliance conditions and angular acceleration for high compliant conditions (Figure 5-3).

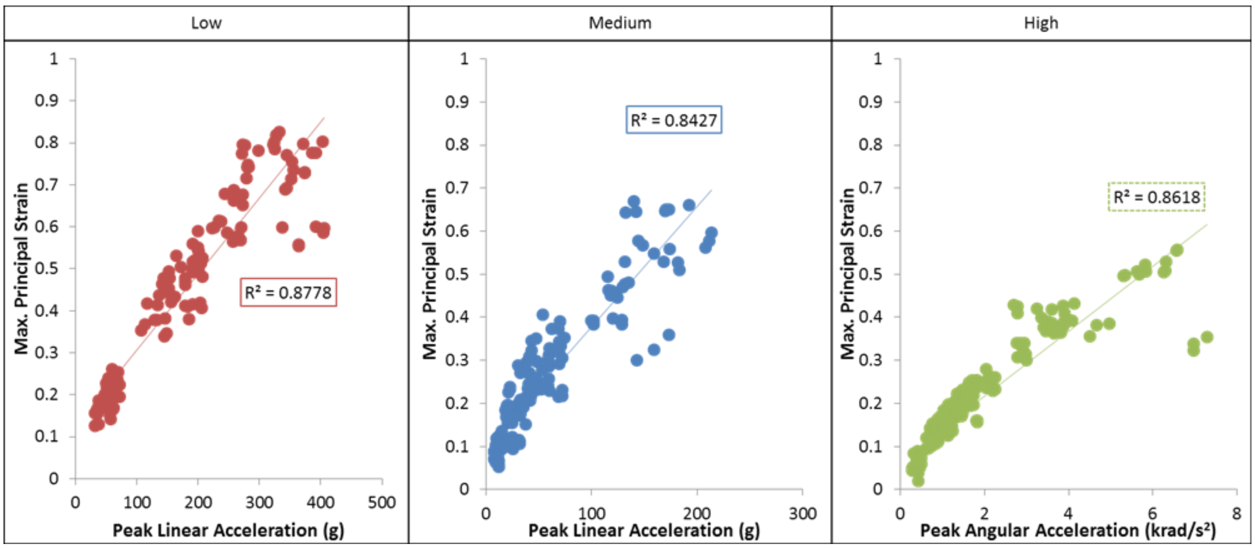

Figure 5-3. Head acceleration-strain relationships for low (left), medium (centre), and high (right) compliant collision conditions.

5.4 Discussion

Compliance was the most influential parameter for increases to peak head acceleration and impact velocity was for FE strain. Therefore, the most effective strategies for reducing concussion risk in collision sports is to increase compliance by making protective equipment thicker and softer and to minimize striking velocities between players by enforcing rule changes. These findings are similar to those found for sport related falls where the head collides with the infinite mass of the earth (Oeur & Hoshizaki, 2016). Adding compliance attenuates the impact by compression of the foam liner, which causes a decrease in the peak and an elongation of the acceleration pulse (Gurdjian et al., 1966; Newman, 1998). These findings agree with reports by Gurdjian et al. (1966), where a helmet slows down the rate of increase of peak linear acceleration of the cadaver head in comparison to an unhelmeted cadaver. This study further demonstrates this phenomenon for peak angular acceleration, whereas strain was most influenced by impact velocity as FE calculations take into account peak, slope, duration, and shape characteristics that are a reflection of impact compliance.

5.4.1 Strategies for Risk Reduction

The tolerance to concussion in humans is a function of the relationship between the magnitude and duration of the acceleration pulse (Hoshizaki et al., 2016; Lissner et al., 1960). Based on this paradigm, this study connects the loading condition (compliance, velocity, mass, location) with concussion responses, providing a road map to identifying the specific conditions within sport environments that contribute to a high risk of injury. Risk assessment identified low compliance impacts presenting risk at 1.5 m/s with the side location producing the highest responses. It would be beneficial to wear head protection in sports requiring none and adding special innovation to the side to minimize head accelerations at this location. However, increasing compliance by wearing a helmet changes the conditions of risk, which are now subject to the limits identified at the medium level. Impact velocities of 3.0 and 4.5 m/s were required for medium and high compliance respectively, with the rear boss non-centric location contributing to increased angular accelerations and strains. These findings confirm the dangers of glancing blows in American football and ice hockey, and hook punches in boxing.

Previous reports examining the relationship between head acceleration and strain have identified that angular acceleration is a stronger predictor under helmeted conditions (R^2 = 0.7-0.84) in comparison to linear acceleration (R^2 = 0.47-0.65) (Forero Rueda et al., 2011; Kleiven, 2007). Contrary to this study, linear acceleration had a stronger relationship with strain under medium compliance (R^2 = 0.84), however angular acceleration was only slightly lower (R^2 = 0.80). This study further adds that linear acceleration is best predictive of strain under low compliance, and angular acceleration for high compliance conditions. Strong relationships between head acceleration variables and strain can allow researchers to infer about intracranial strain levels from direct head acceleration measures during physical impact tests. This can be considered an advantage for head protection design and innovation as finite element analysis can be a time consuming process.

5.4.2 Effective Performance Ranges

The goal of designing head protection is to incorporate padding strengths (thickness, surface area, density) that attenuate impact energies characteristic of the injury loading conditions in sport (Newman, 2002). Mass-compliance interactions causing increased responses for the 9 kg mass illustrate the concept of 'effective performance ranges' that demonstrate the intricate balance between the attenuating properties of protective layers, when they are no longer effective and can act to increase harm. Particularly with medium compliance, these findings suggest that a single punch from an elite boxer is more dangerous than a helmet- helmet collision in American football (Figure 5-2). Unlike American football, punches are delivered multiple times in a single match and taking into account the frequency of impact trauma sustained throughout an athlete's career, the risk for long-term neurological complications (e.g., chronic traumatic encephalopathy) is unparalleled to non-contact sport (Gavett et al., 2011; McKee et al., 2009).

In general, striking masses larger than the head-neck (> 6.0 kg) would cause the head to accelerate at impact and continue on its path post impact. Adding compliance resulted in interactions where the 15 and 21 kg masses were sufficiently large enough to compress the compliant layers involved (vinyl skin on the headform or foam layers for medium and high compliance). Compression of these materials resulted in slightly longer contact times (increased durations) and lowered head acceleration and strain responses. On the contrary, the 9 kg mass was not large enough to engage the compliant layers, resulting in greater magnitude accelerations over shortened durations. A comparison of linear and angular acceleration time histories for 9 and 15 kg demonstrates phenomenon (Figure 5-4

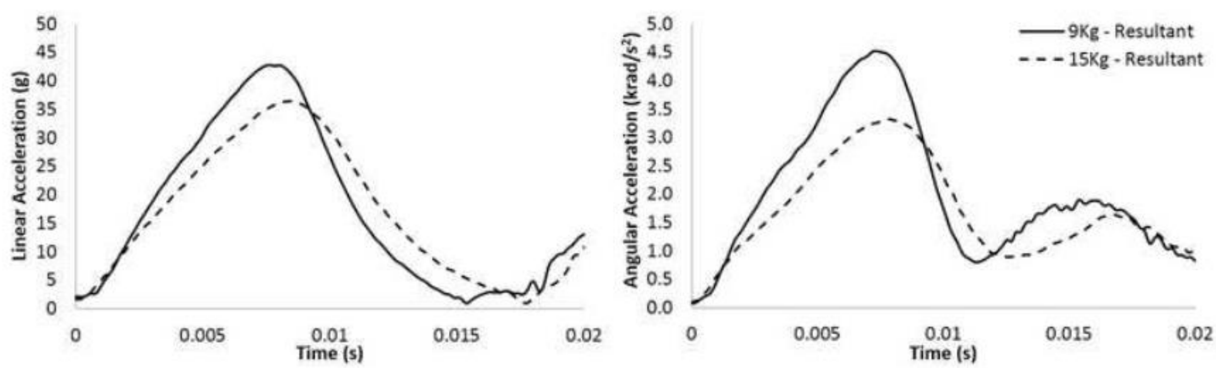

Figure 5-4. Resultant linear (left) and angular (right) acceleration time histories for 9 and 15 kg VN impacts at 4.5 m/s for the rear boss location.

5.4.3 Limitations

Simplified rigid headforms, impact surfaces, and an idealized finite element model were used to represent the collision mechanics involved in sport and the complex dynamic head and brain response to loading. The head and neckforms represented the mass and geometrical dimensions of the average male are composed of steel, rubber, and vinyl necessary for the repeated impact tests conducted (Mertz et al., 1989). A non-directional neckform was used in place of the standard Hybrid III neckform due to sagittal plane biases designed to capture whiplash injury responses in car crash investigations, not necessarily meant for studying direct head impact (Kang et al., 2005; Walsh et al., 2012). Additionally, the UCDBTM has been compared with cadaveric intracranial pressures and brain motion data from impact, as well as in clinical investigations of brain lesions, bleeds, and concussions producing responses consistent with the literature (Doorly & Gilchrist, 2006; Hardy et al., 2001; Nahum et al., 1977; Oeur et al., 2015; Post et al., 2015; Trosseille et al., 1992).

The specific levels of head acceleration and strain used to assess the data for concussion risk served discussion purposes only. There exists a large range of concussive responses in the literature, which is a reflection of human biological variance, the wide-ranging clinical signs and symptoms associated concussion, and study design and methodological differences. Additionally, experimental work has demonstrated directional and regional vulnerabilities of the brain to loading, and that neural injury processes can be initiated at much lower energy levels than those that require an outward expressions of symptoms (Bailes et al., 2013; Hodgson et al., 1983; Prange & Margulies, 2002). The value in this research is in elucidating the interactions and influences of mechanical impact parameters causing increased head and brain response that contribute to creating concussion risk in sport.

5.5 Conclusions

The most effective strategies for reducing head acceleration and strain from head impacts in collision sports are to increase compliance and decrease impact velocity by making protective equipment thicker and softer, and to minimize striking velocities between players by enforcing rule changes. Designing larger protective equipment is a challenge for manufacturers as a balance between the size, weight, and practicality for use in sport needs to be considered without sacrificing protective capacity. The novel aspect of this study is in describing interaction effects of impact parameters that create increased risk within sport conditions. Interactions between mass and compliance created conditions where the 9 kg mass produced increased head accelerations and strains, which emphasize the importance of understanding the ‘effective performance range’ of protective devices. The minutiae of these types of interactions are often hidden in impact studies as incremental effects of impact parameters are seldom tested due to the large number of parametric tests required. These findings can be used to guide impact management strategies, inform of the design limits for sport equipment, and help to manage game rules to reduce risk of concussion in sport.

6 Mechanical Impact Parameters Affecting the Volume and Distribution of Brain Trauma from Sports Collisions

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6.0 Abstract

Diffuse brain injuries depend on the severity and the total number of damaged axons, and the anatomical location of the disrupted networks. The nature of the impact and effects on the brain are influenced by impact velocity, mass, surface compliance, and impact location. The purpose of this study was to determine the most influential parameter on peak and volume of tissue strain for collision impacts. A pendulum delivered four masses (3, 9, 15, 21 kg) at four velocities (1.5, 3.0, 4.5, 6.0 m/s) to impact a Hybrid III headform through centre gravity (CG) and non-CG. Three materials simulated low, medium, and high compliance conditions in sport to capture response ranges from sub-concussion to concussion. Headform accelerations were input into a brain finite element model to obtain peak strain in the frontal, temporal, parietal, and occipital lobes and the volume of the brain experiencing 10, 15, 20, and 25% strains are reported. Strain was most sensitive to velocity with the temporal lobe displaying the highest values. Mass-compliance interactions resulted in the 9kg striking mass able to increase trauma to the brain. Under low compliance, the CG location had the highest responses however medium and high, representing helmeted and padded conditions, non-CG resulted in higher values, confirming the dangers of glancing blows and hooks in boxing.

6.1 Introduction

Impacts to the head are the main cause of sports related brain trauma from repetitive sub-concussive hits, concussive injuries, and risk of long-term neurological diseases (Bailes et al., 2013; Gavett et al., 2011). Contact and combat sports, such as American football, ice hockey, boxing and mixed martial arts (MMA) present a substantial risk for serious brain trauma as these sports are associated with the highest incidences of concussion, a high rate of repetitive impact trauma, and athletes in these sports have demonstrated clinical and neuropathological evidence of chronic traumatic encephalopathy (CTE) (Hutchison et al., 2014; McKee et al., 2009; Tommasone & Valovich McLeod, 2006).

A direct impact to the head results in linear and angular acceleration responses (Newman, 1998; Ommaya et al., 2002). Previous physical and animal models of concussion found that linear acceleration was highly correlated with pressure changes in the brain, affecting the cerebrum and brainstem, playing a primary role in concussive effects (Gross, 1958; Gurdjian et al., 1963; Gurdjian et al., 1958). Angular accelerations, as a consequence of rotationally dominant head motions, have been proposed as a primary cause of brain injuries due to the relative weakness of the brain tissue to shear strains (Holbourn, 1943). Angular accelerations and strains on the white matter tissues have been linked to diffuse brain injuries, ranging from mild concussion to severe diffuse axonal injuries (DAI), and death (Margulies & Thibault, 1992; Ommaya & Gennarelli, 1974). Strain elongation of an isolated giant squid axon at 10, 15, 20, 25% strains displayed graded effects on the electrical conduction and integrity of the tissue from delayed recovery, residual effects, irreversible injury, and structural failure (Thibault et al., 1990). Ommaya and Gennarelli (1974) proposed the 'Centripetal Theory of Concussion' stating that less severe injury occurs when only the cortical-subcortical areas of the brain are affected (low magnitude and volume), and more severe injury will occur as the level of trauma extends to the core of the brain (higher magnitudes and volumes), namely the brainstem based on experimental research on primates. Gennarelli et al. (1982) observed a link between severities of DAI outcomes in animals with more severe axonal injury combined with increased volume of damaged axons surrounding and including the brainstem resulted in prolonged coma. Jenkins et al. (1986) provided evidence in support of the 'Centripetal theory of Concussion' based on observations of severely brain injured patients displaying lesions on magnetic resonance imaging (MRI) scans that correlated with the degree of consciousness (full, impaired, coma) with the centripetal progression of damage from the cortical, subcortical, and deep white matter tissues. Therefore, the severity of diffuse brain injuries is related to the magnitude of neural trauma, anatomical region affected, and volume of tissue trauma, with higher severities displaying greater strains in higher amounts (Gennarelli et al., 1998; Ommaya & Gennarelli, 1974). While these effects have only been confirmed for the severe end of the spectrum of diffuse brain injuries, it has been proposed that concussion severity follow the same principles at the mild end of the spectrum (Ommaya et al., 1994).

The relationship between the nature and amount of brain trauma is related to the mechanical effects of loading on the brain (Denny-Brown & Russell, 1941). In sport, impact loading on the head is a common occurrence in contact sports such as American football, ice hockey, and soccer (Cusimano et al., 2013; Daneshvar et al., 2011). Characteristics that govern the nature of the impact load include impact velocity, striking mass, and compliance (stiffness) of the colliding object (Holbourn, 1943). These energy related parameters govern the amount and rate of force transfer from an impact (Gurdjian et al., 1966). The location of an impact is another important injury characteristic as this governs the direction of head and brain motion in relation to anatomical and structural constraints, such as the falx cerebri and tentorium cerebelli that have been proposed to limit coronal and sagittal motions (Gennarelli et al., 1987; Ommaya et al., 1994). In animal models, the brain has demonstrated increased susceptibility to injury from coronal loading, with subjects displaying prolonged loss of consciousness and increased brain tissue damage (Gennarelli et al., 1982; Hodgson et al., 1983). Impacts to the side of the head have been demonstrated to present with a high risk of concussion in contact sport and has been identified as the primary location for increased knock-outs in MMA (Hutchison et al., 2014; McIntosh et al., 2000). The relationship between mechanical impact parameters increasing trauma to the brain by peak and volume of strain damage has not been described for collision sports as a potential indicator of concussive severity (Ommaya et al., 1994). Identification of dangerous sport conditions associated with increased risk and brain injury severity can better inform athletes of the potential dangers of participating in contact sport and can help to guide risk reduction strategies through improved head protection or game rule changes. Therefore, the purpose of this study was to determine the influence of mechanical impact parameters on strain in four brain lobes (frontal, occipital, temporal, and parietal) and the volume of brain experiencing 0.10, 0.15, 0.20, 0.25 strain.

6.2 Methods

6.2.1 Equipment

Impacts were delivered using a multi-mass pendulum that consisted of a steel hollow frame (3.36 kg) with adjustable masses of secured to the body as described by Karton et al. (2014). Four impact masses were tested and compared, to represent a characteristic range of striking masses in sports: 3, 9, 15, and 21 kg (Pellman et al., 2006; Rousseau & Hoshizaki, 2015; Walilko et al., 2005); and moreover, four impact velocities were selected, to capture a range of responses associated with sub-concussive concussive injuries: 1.5, 3.0, 4.5 and 6.0 m/s (Walilko et al., 2005; Withnall et al., 2005). Three levels of impact compliance were selected to create acceleration pulse durations characteristic of unprotected (5ms), protected (15ms) and well-padded (25ms) surfaces in collisions comprising of steel, 0.025 m vinyl nitrile (VN 602) foam, and a 0.067 m R338 Rubatek Rubber foam (Oeur & Hoshizaki, 2016). A centre of gravity (CG) impact location and a non-centric (NC) location were selected to compare linear and rotationally dominant impacts in which both are applied from the side, forcing the head to move laterally (Figure 6-1). The rear boss NC location measured 0.08 m from the side CG (0.00 m) and used as quantitative predictor variables for the regression analysis.

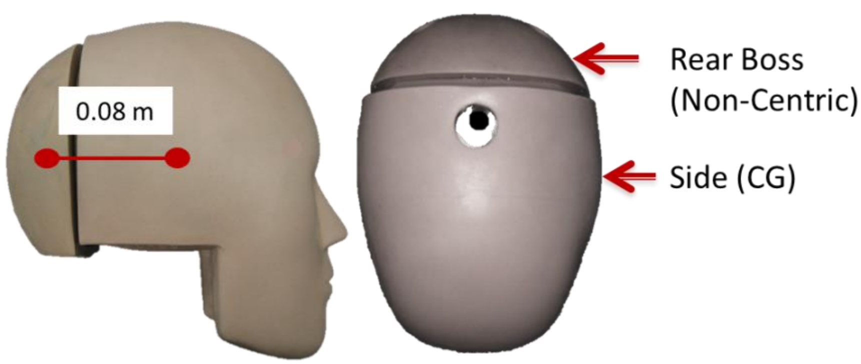

Figure 6-1. Centric and non-centric impact locations.

Head accelerations were captured using an instrumented average male Hybrid III headform (Humanetics Innovative Solutions, Michigan, USA) and non-directional neckform (University of Ottawa, Ottawa, Canada). The Hybrid III headform is composed of a single cast aluminium part covered with vinyl skin with a hollow compartment fitted with nine-linear accelerometers (Endevco 7264C-2KTZ-2-300, Meggitt Inc., San Juan Capistrano, CA) arranged to capture centre of gravity linear accelerations and calculate angular accelerations (Padgaonkar et al., 1975). A non-directional neckform was used in lieu of the standard Hybrid III neckform to remove directional restrictions built into the original neckform designed for evaluating neck loads in car crash scenarios (Kang et al., 2005; Walsh et al., 2012).

6.2.2 Finite element Analysis

Linear and angular headform accelerations were input into the University College Brain Trauma Model (UCDBTM) to determine intracranial strains associated with the loading conditions. The UCDBTM is a finite element model of the human brain constructed using a computed tomography scan of a cadaver and with material definitions consistent with the literature (Table 2-5) (Horgan & Gilchrist, 2003, 2004). To evaluate the spatial distribution of strain, the brain was divided into four regions, representing the frontal, temporal, parietal, and occipital lobes where peak maximum principal strains were recorded in each lobe (Figure 6-2). The temporal lobe receives and processes auditory inputs for sound and language, whereas the frontal lobe is involved in planning, memory, reasoning, memory, controlling emotions and behaviours, speech production and motor control. The parietal lobe receives input from the senses (touch, pain, temperature, and proprioception) and plays a role in spatial orientation and the occipital lobe is responsible for receiving visual information and processes colour and depth perception (Marieb & Hoehn, 2007). Cumulative strain damage measures (CSDM) has been proposed as a finite element analysis injury criteria extracting the volume of brain tissue above a critical level of strain (Bandak, 1995). CSDM levels at 10, 15, 20, and 25% strain were evaluated as an indication of severity of diffuse axonal injury (Takhounts et al., 2003).

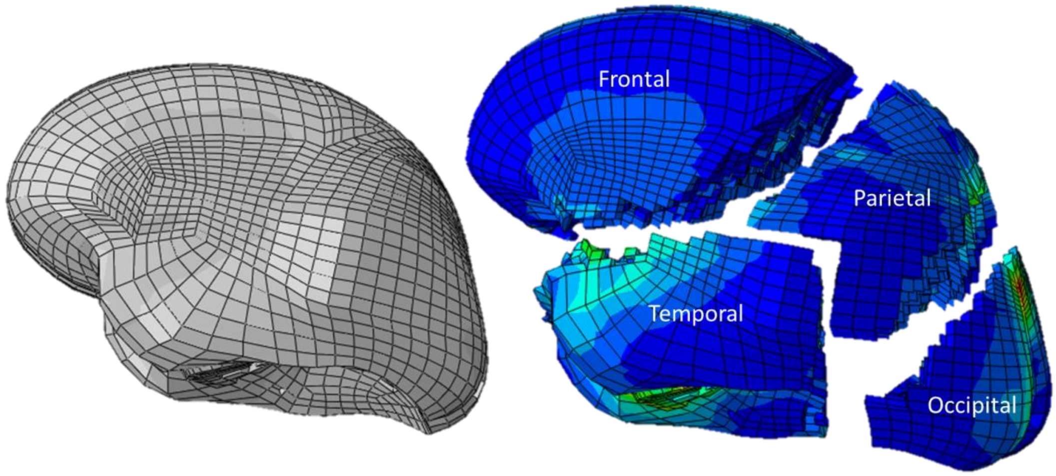

Figure 6-2. UCDBTM divided to approximate the frontal, temporal, parietal, and occipital brain lobes.

6.2.3 Statistical Analysis

Step-wise multiple linear regressions assessed the influence of each of the four impact variables (velocity, compliance, mass, location) on CSDM (10, 15, 20, 25%) and on peak strain in each brain lobe (frontal, temporal, parietal, and occipital). The data was plotted according to compliance levels to best illustrate the effects impact parameter in light of common sport loading conditions. These plots guided further data analyses that include one-way ANOVAs and Tukey tests for the effects of striking mass and brain lobes on peak strains. Additionally, independent t-tests were conducted to determine the effect of head impact location ($p<0.05$). These analyses were conducted to answer specific questions as to when mass contributed to higher values, whether non-centric impacts are more dangerous in sport collisions, and if certain lobes present with higher values. Statistical analysis was performed using IBM SPSS Statistics V 22.0 (Armonk, New York, USA) and significance was accepted at an alpha level of 0.05.

6.3 Results

Significant regression models that included the greatest number of impact variables were identified and corresponding standardized beta coefficients (β) for each variable, the model R^2 and F values are reported in Table 6-1. The order of effects of impact parameters on peak strain and CSDM displayed the same trends. Striking velocity was the most influential for CSDM and peak strains in each brain lobe (0.45-0.77), followed by compliance (0.32-0.57), head impact location (0.11-0.24), and striking mass (0.08-0.24). An exception was the occipital lobe, where velocity (0.62) and compliance (0.64) had similar effects and is likely because strain values were equally spread by these two factors.

Table 6-1. Significant step-wise multiple linear regression models with standardized beta coefficients, p<0.05.

	Strain Volume		
CSDM10	**CSDM15**	**CSDM20**	**CSDM25**
$\beta_{Velocity} = 0.72$	$\beta_{Velocity} = 0.62$	$\beta_{Velocity} = 0.53$	$\beta_{Velocity} = 0.45$
$\beta_{Compliance} = 0.55$	$\beta_{Compliance} = 0.47$	$\beta_{Compliance} = 0.37$	$\beta_{Compliance} = 0.32$
$\beta_{Location} = 0.20$	$\beta_{Location} = 0.18$	$\beta_{Location} = 0.16$	$\beta_{Location} = 0.13$
$\beta_{Mass} = 0.08$	$\beta_{Mass} = 0.10$	$\beta_{Mass} = 0.12$	$\beta_{Mass} = 0.12$
$R^2 = 0.695$	$R^2 = 0.521$	$R^2 = 0.373$	$R^2 = 0.280$
$F = 197.692$	$F = 70.381$	$F = 38.537$	$F = 25.148$
	Brain Lobes		
Frontal	**Temporal**	**Parietal**	**Occipital**
$\beta_{Velocity} = 0.74$	$\beta_{Velocity} = 0.77$	$\beta_{Velocity} = 0.70$	$\beta_{Compliance} = 0.64$
$\beta_{Compliance} = 0.62$	$\beta_{Compliance} = 0.67$	$\beta_{Compliance} = 0.62$	$\beta_{Velocity} = 0.62$
$\beta_{Location} = 0.15$	$\beta_{Mass} = 0.11$	$\beta_{Location} = 0.20$	$\beta_{Location} = 0.24$
$\beta_{Mass} = 0.09$		$\beta_{Mass} = 0.07$	$\beta_{Mass} = 0.08$
$R^2 = 0.773$	$R^2 = 0.841$	$R^2 = 0.738$	$R^2 = 0.691$

6.3.1 Risk Assessment

Data plots are presented in Figure 6-3 and risk assessment was conducted by comparing peak strain data against a 0.2 value (red dotted line), indicative of levels associated with a 50% risk of sports related concussions (0.19-0.27) from sport injury reconstruction (Kleiven, 2007; Patton et al., 2013; Zhang et al., 2004) and is an indicator of significant morphological injury to the axon (Bain & Meaney, 2000; Galbraith et al., 1993; Maxwell et al., 1997; Thibault et al., 1990). CSDM15 results were compared to a 0.55 value indicative of a 50% risk of DAI at CSDM 15 levels (Takhounts et al., 2003).

Low compliant impacts to the side (CG) produced the higher strain responses (CSDM and peak) than the rear boss (NC) with the temporal lobe presenting with the largest values. Mass-compliance interactions from 3.0-4.5 m/s, demonstrated that the 9 kg mass had the highest strain responses. For medium compliant impacts, the rear boss (NC) impacts contributed to higher strain responses with the temporal lobe similarly exhibiting the highest values. Mass-compliance interactions also showed that the 9 kg mass gave the highest responses, however at

6.0 m/s, the 9kg response was not significantly different from 21 kg. Under high compliance, rear boss (NC) contributed to higher responses (CSDM and peak) with the temporal lobe showing the highest peaks ($p<0.05$). Mass and velocity interactions similarly showed the 9 kg mass having the highest values from 3-4.5 m/s; however, at 6.0 m/s the effect of mass was diminished, where 3kg produced the lowest values and 9-21 kg had the same effect.

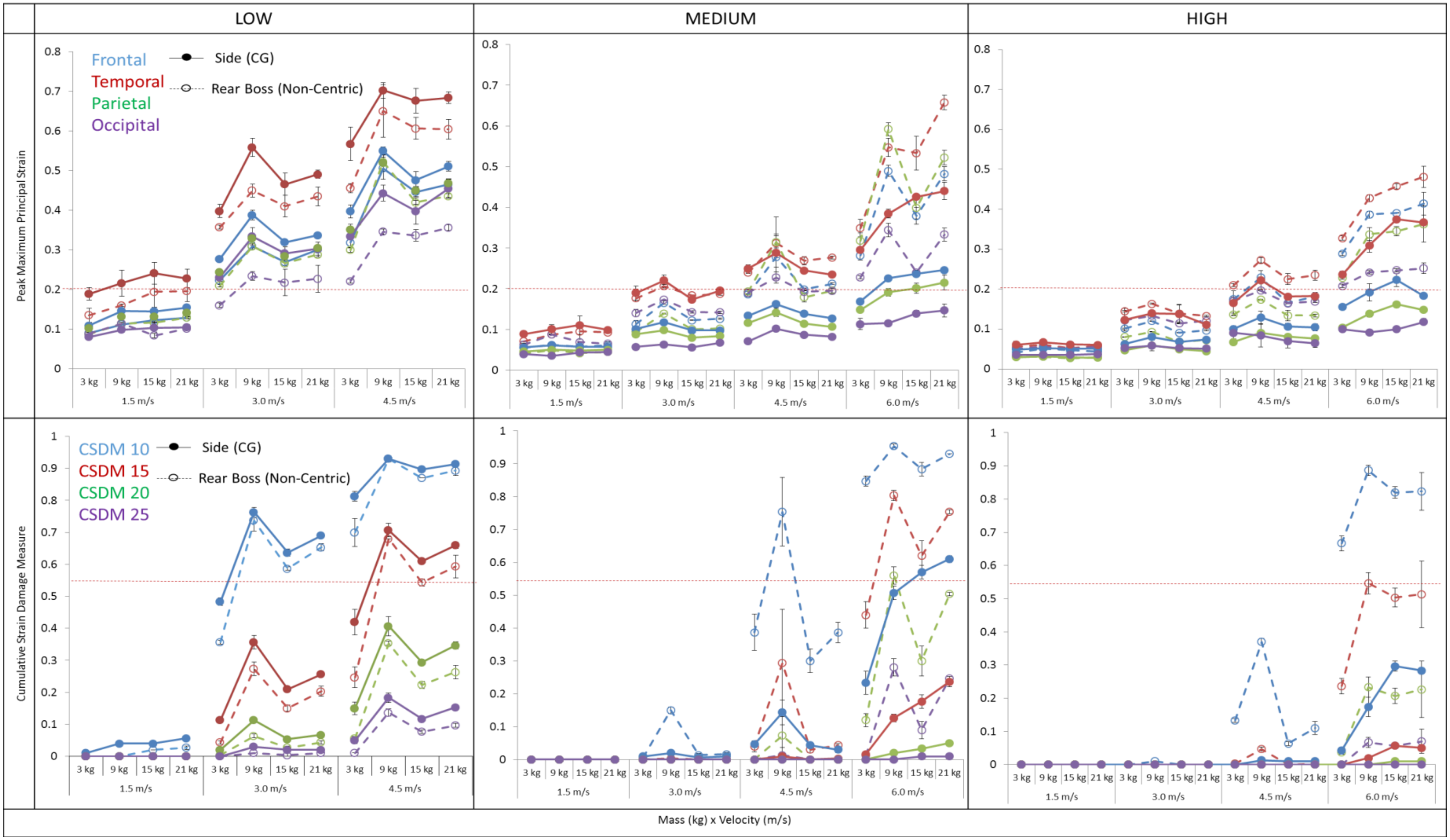

Figure 6-3. Plots of peak maximum principal strain and CSDM results for low (left), medium (centre) and high (right) compliance condition. Dotted lines for peak strain plots indicate a 0.2 strain level for concussion and lines for CSDM plots indicate 50% risk of diffuse axonal injury for CSDM15 = 0.55 as reported by Takhounts et al. (2003).

6.4 Discussion

The greatest increases in strain magnitudes and volumes (CSDM) were from striking velocity. Compliance of the striking object was less influential on strain responses. The stronger influence of striking velocity can be explained by the finite element model taking into account compliance effects (curve shape and duration) in the calculation of strain, as the full linear and angular acceleration time histories were considered. The temporal lobe consistently exhibited the highest strains for all conditions, regardless of mass, velocity, compliance, or impact location (CG versus NC). This may be due to the proximity of the locations (side and rear boss) to the temporal lobe as the impacts were directed to the side of the head causing primarily coronal plane motion. The temporal lobe region has been demonstrated to be sensitive to strain deformations associated with concussive symptoms (Viano et al., 2005), and is a hallmark region for neuropathological changes associated with chronic traumatic encephalopathy (CTE) from repetitive sub-concussive and severe concussive traumas (McKee et al., 2009).

Low compliance collisions represent MMA punches and elbow to head impacts in football when players compete for jump balls, in addition to Taekwondo kicks, knee and full body to head impacts from colliding players chasing loose balls (Fife et al., 2012; Hoshizaki et al., 2016; McIntosh et al., 2000; Withnall et al., 2005). While all masses produced concussive strains (0.2) at 1.5 m/s, 50% risk of DAI proposed by Takhounts et al. (2003) was present for 9-21 kg at 4.5 m/s (CSDM 15 results in Figure 6-3). The high-energy nature of low compliant impacts indicate that centric impacts (side CG) produce higher values than non-centric locations (rear boss), since centric conditions allow for a complete connection between the pendulum and the head, allowing for higher amounts of energy to be transferred as compared with the nature of non-centric, glancing blows. In contrast to medium and high compliance conditions, glancing blows (NC) contributed to higher volumes and peak strains and represent boxing punches, helmet, elbow, and shoulder collisions (Pellman et al., 2004; Rousseau & Hoshizaki, 2015; Tommasone & Valovich McLeod, 2006; Walilko et al., 2005). These findings further support empirical observations of the dangers of hook punches from elite boxers and glancing blows in ice hockey and American football. These results also provide insight into the logic behind developing larger and thicker helmets meant to attenuate more impact energy. The medium compliance level can be seen as a typical helmet (0.025 m thick) and the high compliance level to represent theoretically larger helmets (0.067 m thick). The higher compliance offsets the presentation of concussion risk from 3.0 m/s and 4.5 m/s, however both present a 50% risk of DAI at 6.0 m/s for 9-21 kg striking masses. While this may seem like a plausible strategy for increasing protection, the challenge with the design of head protection is to balance the size and weight of the helmet for practical use in sport and its protective capabilities (Hoshizaki & Brien, 2004).

Peak strain magnitudes are an important indicator of concussion risk, however in a combination with strain volume, both may give a more complete picture of the potential risk of brain trauma severity as proposed by Ommaya and Gennarelli (1974) as part of the ‘Centripetal Theory of Concussion’. A medium compliant, 9 kg mass traveling at 4.5 m/s striking the head in a NC manner engages nearly 75 % of the brain in 0.1 strain, which has been associated with functional trauma at the level of the tissue (Bain & Meaney, 2000; Galbraith et al., 1993; Maxwell et al., 1997). In addition, interactions between mass and compliance for 9 kg provides further support the importance of defining the effective performance range of protective equipment as blows delivered outside of these windows can cause increased magnitudes and volumes of strain. Mass-compliance interactions were similarly reported for peak linear and angular acceleration measured from a Hybrid III headform (Figure 5-2). An understanding of the relationships between mechanical impact parameters on the volume and distribution of strain helps to identify risky environmental situations in sport, which can guide rule changes and strategies that better protect athletes.

6.4.1 Limitations

Data interpretation and extrapolation to injury risk is limited to the tools and modelling methods used to represent the complex human head and brain system response to impact loading. The skull and brain exhibit complex non-linear viscoelastic, anisotropic, regional and structural material properties that remain a challenge to accurately model (Bandak, 1995; Chatelin et al., 2013; Prange & Margulies, 2002). While the mass and geometrical dimensions of the head and neckform represented average male characteristics, they were composed of robust materials (steel, vinyl, and rubber) necessary for repeatability of head impact testing. Additionally, the flat surfaces and levels of compliance used in this study were selected to represent overall loading conditions in sport, neglecting localized material and surface geometries that are important during impact energy transfer (Spyrou et al., 2000). Despite the UCDBTM used in this study being more rudimentary and smaller in size than other sophisticated models, it was compared with cadaveric intracranial responses and used in clinical investigations of brain lesions and concussive injuries (Doorly & Gilchrist, 2006; Hardy et al., 2001; Oeur et al., 2015; Post et al., 2015; Trosseille et al., 1992). Specific levels of strain and CSDM responses were presented to infer about the potential risk of concussion and DAI from the data however neglect a number of biological and subject-specific factors (e.g., subject history, age, sex, genetics) that affect injury risk in the population (Kutcher & Eckner, 2010). This study highlights the relationships between mechanical impact parameters, defining the loading condition and contributions to increased strain responses.

6.5 Conclusions

Head collisions are the leading cause of sports-related brain trauma including concussions, particularly in contact and combat sports. Striking velocity and was the most influential parameter on peak and volume of strain, however the temporal lobe consistently displayed the highest values, which may indicate that this region may be especially vulnerable to impact. Due to the high-energy nature of low compliant impacts, side centric impacts resulted in higher strain responses, however for medium and high compliant conditions, non-centric, rotationally dominant impacts created higher strain responses, confirming the dangers of glancing blows in contact sport. This study provides some insight into the notion of creating larger helmets to provide increased protection, and while going from medium to high compliance affords some protective effects, the trade-off is developing helmets with increased thickness, size, and weight, which is not ideal in sport environments. Furthermore, mass-compliance interactions within sport collision environments have the potential to increase trauma to the brain.

III Closing

7 Discussion

Summary | Implications | Limitations | Conclusion

7.0 Summary

Sport and recreation activities are important to maintain a healthy and active life style (Warburton et al., 2006), however participation in some sports present a significant risk of brain injury with approximately 300,000 people incurring concussions from sport each year in Canada (Billette & Janz, 2011; Langlois et al., 2006; Thurman et al., 1998). Most concussions resolve within a few days, however, reports of long-term problems and permanent damage has been associated with severe and repeated concussive brain injuries (Marshall et al., 2012; McKee et al., 2009; Omalu et al., 2005; Rimel et al., 1982; Rimel et al., 1981). The economic burden of concussion has been estimated to be $16.7 billion in the US alone and has been considered an underestimation of the real cost as this figure does not take into account lost productivity, decreased quality of life, costs taken up by family members and friends (Thurman, 2001).

The majority of these types of injuries are preventable when principles of injury biomechanics are developed and appropriately applied (Ommaya et al., 1994). Contact with the head is an important characteristic of sport concussion as they are a necessary cause for injury (Cusimano et al., 2013; Daneshvar et al., 2011). An impact imparts a mechanical load on the head and brain that initiates a complex injury process at the level of the cells and tissues that disrupt overall brain function responsible for injury (Ommaya et al., 1994). A paradigm for head impact research establishing pathways between mechanical loading, head motion (linear and angular acceleration), brain deformation (strain) and injury proposed by Ommaya and Gennarelli (1974); Ommaya et al. (1994) has guided this dissertation in delineating the role of mechanical impact parameters contributing to concussion risk for head impacts in sport (Figure 1-2). Each event sets a unique platform by which energy is transferred to the head. Impact parameters that govern the mechanical effect of an impacting body on the brain include inbound velocity, object mass, head impact location, and surface compliance (stiffness) (Denny-Brown & Russell, 1941; Gurdjian et al., 1966; Hodgson et al., 1983). It was the purpose of this dissertation to describe the role of each mechanical impact parameter on creating injurious levels of loading of head acceleration and brain tissue strain for sports related concussions.

7.1 Implications

7.1.1 Fall vs Collision Impacts

Plots of peak resultant linear and angular acceleration as a function of the duration of the loading pulse (ms) comparing falls (hollow squares) and collisions (filled circles) with a differentiation for low (red), medium (blue), and high (green) compliance conditions representing unprotected, helmeted, and well-padded surfaces are demonstrated in Figure 7-1. Peak maximum principal strain results comparing falls and collisions for each level of compliance are demonstrated in Figure 7-2. Independent t-tests (IBM SPSS Statistics V 22.0, Armonk, New York, USA) compared falls and collision impacts at each level of compliance for peak linear and angular acceleration and strains ($p<0.05$).

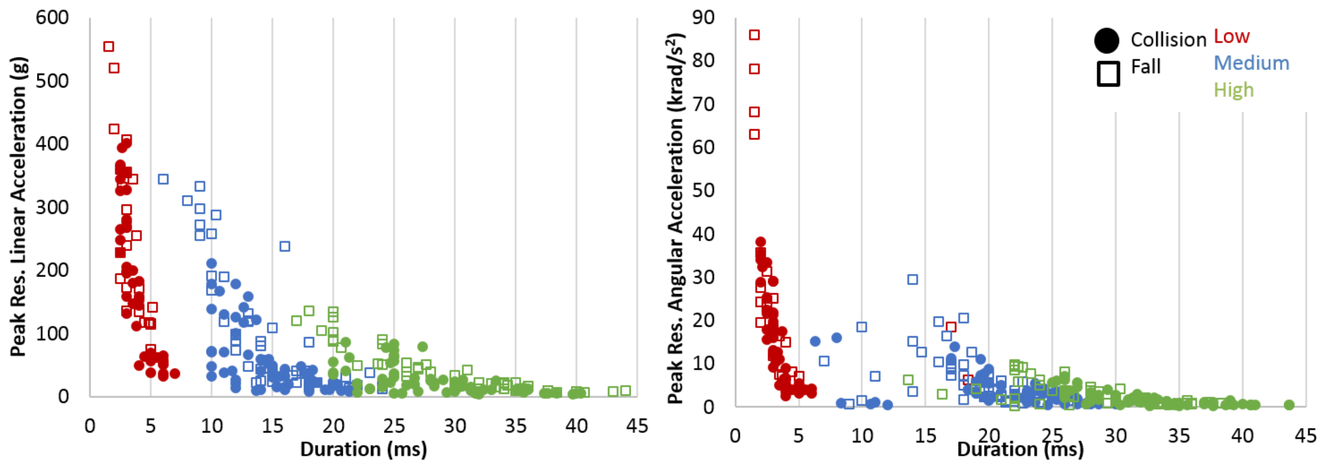

Figure 7-1. Peak resultant linear (left) and angular (right) accelerations as a function of loading pulse duration comparing falls (hollow squares) and collisions (filled circles) for low (red), medium (blue), and high (green) compliance.

In all comparisons, Levene's test demonstrated unequal variances due to the different sample sizes as low compliance had fewer conditions due to the high energy nature of this event and falls had 3 levels of impact mass (head size) and collisions had 4 levels. Falls produced significantly higher magnitudes of peak linear and angular accelerations and strains than collisions under all conditions ($p<0.001$), except strain under low compliance ($p=0.094$). This is likely due to the low compliance conditions transferring a large amount of energy to the head that the method of impact no longer had an effect on peak strain values. Overall, falls in sport present an increased risk of brain injury due to a more complete transfer of energy during the impact that contribute to increased head accelerations and strains.

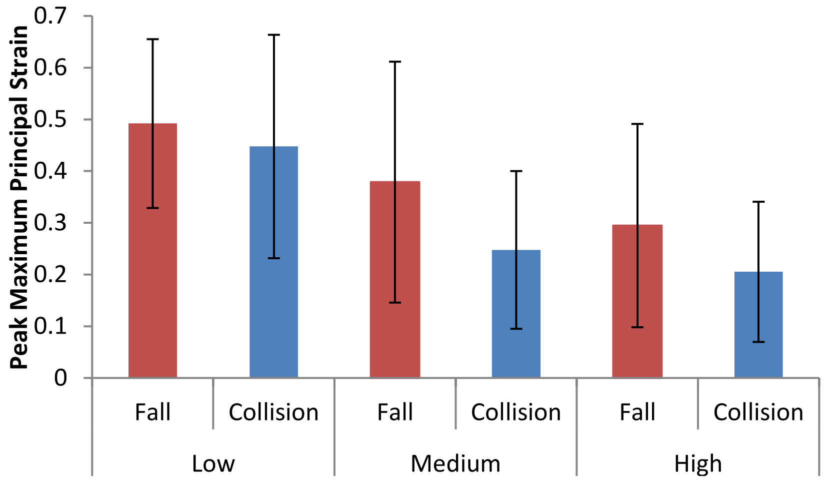

Figure 7-2. Peak strain and standard deviation bars illustrating the responses for falls and collisions according to each level of compliance.

7.1.1.1 Most Influential Variable

For both impact events, surface compliance was consistently the most influential parameter for increases to peak head acceleration where lower compliance (stiffer surfaces) caused higher values. Furthermore, increases to FE strain were most influenced by impact velocity, suggesting that the most effective strategies for lowering head acceleration and strain and therefore concussion risk in sport is to increase compliance by increasing padding thickness and to minimize impact velocity through game and rule changes. These considerations should be made within limits of reason as increasing the thickness of helmets and body protection by making them bigger can work to increase the moment arm of a glancing blow, which would introduce an increased angular component at impact. Furthermore, managing impact velocities should be done with the overall goal in decreasing injury risk without drastically changing the nature of the sport and how the game is played. Impact mass had the least effect on overall increases in response for both falls and collisions but resulted in interaction effects that created increased responses for specific masses.

7.1.2 Injury Risk Assessment

While a large range of injury related responses have been associated with concussion in the literature (Figure 2-2 and Table 2-2), identification of impact parameters contributing to injury risk was assessed according to values in Table 2 to simplify the discussion (Hoshizaki et al., 2016). Additionally, a 0.2 strain level was selected as an indicator of concussion because it reasonably represents conservative values determined from injury reconstructions and is an indication of axonal injury in anatomical tissue tests (Table 2-1) (Bain & Meaney, 2000; Galbraith et al., 1993). This strain level was used as an overall indicator for low, medium, and high compliance conditions, as compliance was less influential on peak strain. These values serve as a reference point to evaluate the data and to determine the specific levels of impact parameters that contribute to injury level responses. These levels can then be used to guide rule and game limits in sport environments and inform head protection manufacturers on the necessary requirements to remain protective under conditions of use.

Table 7-1. Injury response values associated with concussion taken from Hoshizaki et al. (2016) and Table 2-1.

Compliance Condition	Linear Acceleration g	Angular Acceleration krad/sec^2	Strain %
Low	275	20.0	0.2
Medium	120	7.5	0.2
High	30	3.5	0.2

7.2 Limitations

The limitations of the equipment, tools, and test set-ups employed in this research have been presented in earlier chapters and are summarized below:

- Simplified test impact scenarios were used to represent fall and collision events that isolated the contacting surface and the head. Real fall and collision mechanics involve a number of degree of freedoms that affect the ensuring injury risk. With an understanding that the body, neck, and extremities all playing a part in reducing impact velocity and managing the movement of the head, the velocities selected in this study (1.5, 3.0, 4.5, and 6.0 m/s) attempted to capture the best and worst-case scenarios for concussive level injuries.
- The impact surfaces (compliances) were simplified to limit geometrical interactions between the test surface and the localized geometry of the headform. Test compliances were circular, flat surfaces with a diameter of 0.13 m.
- Artificial Hybrid III headforms and neckforms were used to simulate the complex head-neck impact dynamics of the human system. Test headforms are considered valid up to skull fracture levels, which are beyond levels for sports related concussions. The Hybrid

III test headform family are one of the most sophisticated test devices currently available and remains widely used in car crash and experimental head impact testing. A set of non-directional neckforms were used in place of standard directionally biased Hybrid III neckforms designed for whiplash experiments. A comparison of non-directional and standard neckforms on the peak linear and angular acceleration have been conducted for the diverse head impact conditions studied in this research and are presented in the Appendix B for reference.

- The University College Dublin Brain Trauma Model (UCDBTM) is a finite element model of the human brain representing the gross approximation of the complex multi-phasic soft materials of the brain, CSF, and blood vessels. The UCDBTM is comparatively smaller (26 000 elements) than other sophisticated models (300,000 elements) available (Zhang et al., 2004) but this allowed for faster computation times than the much larger models (30 minutes versus 5 hour run times). This was an important consideration given nearly 900 impacts were conducted in this thesis and processed using FE. A subset of the collision tests were run on the Wayne State Brain Injury Model (WSUBIM) developed by Zhang et al. (2004) consisting of approximately 300,000 elements and are presented in the Appendix C. The UCDBTM had a tendency to report lower peak strains than the WSUBIM, and therefore it can be concluded that the UCDBTM reports more conservative values.

7.3 Conclusion

The impact condition plays an important role in determining the risk of injury as it defines how mechanical impact energy can be transferred to the head to create levels of head acceleration and strain necessary for concussion. While main effects of velocity, mass, location, and compliance have been previously studied in the literature, the novelty of the research is the presentation of the interaction effects between these mechanical impact parameters and their implications for understanding concussions from head impacts within the context of sport environments.

Interaction effects are unique to fall and collision events. While both falls and collisions are most influenced by compliance and velocity as the two most contributing factors for increases in head acceleration and strain, headform size (mass) is an important characteristic for concussion with smaller heads resulting in higher values, and larger heads with more surface area have the potential to increase risk of injury from rotation effects. Therefore, these findings support the need for age and size appropriate risk definitions that govern protection criteria for different head sizes. Mass-compliance interactions for collision events resulted in increased responses for the 9 kg mass, emphasizing the need to identify effective performance ranges of protective devices to ensure they are protective under typical impact energies characteristic of the sport and do not increase harm. Glancing blows under medium and high compliance conditions (boxing, helmeted, and shoulder collisions) present a high risk for concussion as they presented higher levels of rotation contributing to increased peak and volumes of strain.

An understanding of the patterns of increased response as a factor of loading parameters offers opportunity to inform injury prevention and mitigation strategies in sports. These findings highlight the importance of impact parameters and their influence on eliciting the performance range of head protection. These findings identify the most effective strategies for reducing concussion risk from head acceleration and strain would be to increase compliance and minimize impact velocity for falls and collisions. These changes should be approached with caution as creating larger helmets can work to increase harm by creating a larger moment arm for higher rotations and that by minimizing striking velocities within the game without changing the nature of the sport. This research emphasizes the need for a better understanding of effective performance ranges for impact protection to be able to identify the limitations of these devices, as impacts beyond these ranges can result in increased trauma to the head and brain. Changes to the game rules, innovations to protective equipment, coaching falling and colliding techniques offer avenues by which these changes can be made and provide opportunities for location specific design requirements and mass-compliance interaction considerations.

7.4 Recommendation for Future Research and Head Injury Prevention

Based on the findings in this thesis, future efforts in head injury prevention should incorporate size and mass appropriate test conditions when designing head protection and developing test protocols for helmet evaluation. In addition, the effective performance ranges should be examined for all head impact protection for levels of impact velocities and masses that are typical during play. This research used flat test surfaces and an unhelmeted Hybrid III headform to simplify the number of tests conducted. Future work should include sport specific tests that use helmets, impact surfaces, and equipment with the correct geometries and material stiffness typical of the sport. Currently there is a paucity of age appropriate concussion risk thresholds for children and adolescents. Future research efforts should aim to identify age-appropriate concussion risk data through accident reconstruction. In addition, age appropriate brain tissue tests and finite element modelling efforts should focus on children and adolescent brains to better model the response to impact for these age groups. Once this data becomes available, the impact conditions necessary to cause injury level responses identified in this research are likely to change for these age groups.

IV Appendix

A List of Contributions

| Published | Submitted | Presented

Studies related to head impact parameters during the course of the doctoral research program

Journals

Published

1. Hoshizaki TB, Post A, Oeur RA, Brien, SE. 2014. Current and Future Concepts in Helmets and Sports Injury Prevention. *Neurosurgery 132*(2): 415-422.

Submitted (in order of thesis chapters)

2. Oeur RA, Gilchrist MD, Hoshizaki TB. The Role of Impact Parameters on Concussion Risk from Collisions in Sport. Submitted to *Sports Engineering*. 2017.
3. Oeur RA, Gilchrist MD, Hoshizaki TB. Mechanical Impact Parameters Affecting the Volume and Distribution of Brain Trauma from Sport Collisins. Submitted to *Materials & Design*. 2017.

Book Chapter

4. Hoshizaki TB, Oeur RA, Post A, Koncan D, Kendall, M, Karton C. *How Do Concussions Happen?* In: Keightly M, Gagnon I, Ptito A (Eds). Sports Concussions Continuum: A Complete Guide to Recovery and Management. CRC Press: Forthcoming 2017; 10-15.

Conferences

Podiums

5. Oeur RA, Hoshizaki TB. 2016. The Effect of Impact Compliance, Velocity, and Location in Predicting Brain Trauma for Falls in Sport In Proceedings of the International Research Council on the Biomechanics of Injury.
6. Hoshizaki TB, Oeur RA. 2016. The Relationship between Impact Characteristics and Brain Tissue Strain Distributions. Keystone Symposia on Molecular and Cellular Biology – Traumatic Brain Injury: Clinical, Pathological, and Translational Mechanisms (J3). Sante FE, NM, USA.

Posters

7. Oeur RA, Gilchrist MD, Hoshizaki TB. July 2017. The Importance of Interacting Mechanical Impact Parameters on Increasing Trauma to the Brain. In Proceedings of the Frontiers in Traumatic Brain Injury. London, UK.

Non-Refereed

8. Oeur RA, Hoshizaki TB. 2017 Connecting Impact Events to Head Acceleration and Brain Tissue Strains for Concussions in Ice Hockey. In Mayo Clinic Ice Hockey Summit III: Action on Concussion. Rochester, MN, USA. Invited Faculty Member.

9. Oeur RA, Hoshizaki TB. 2016. Relationship between Impact Location, Compliance, ad Injury Risk: Implications for Helmet Protection. Scientific Meeting with Reebok-CCM, Montreal, QC, Canada.

B Neckform Comparison

Non-directional Neckform | Hybrid III Neckform

B.1. Purpose

A comparison between the standard adult male Hybrid III neckform and the University of Ottawa adult sized non-directional neckform were tested in a combination with a standard Hybrid III headform using a monorail drop tower under the range of impact conditions investigated in this thesis (Figure B-1). The head and neckform were dropped using a monorail drop tower onto low, medium, and high compliant surfaces at 1.5, 3.0, 4.5, and 6.0 m/s at the front, front boss, side and rear boss impact locations. Impacts were conducted for three consecutive trials measuring peak linear and angular acceleration response. The effect of neck-type on peak linear and angular acceleration (Figure B-2) was compared using independent samples t-tests ($p<0.05$).

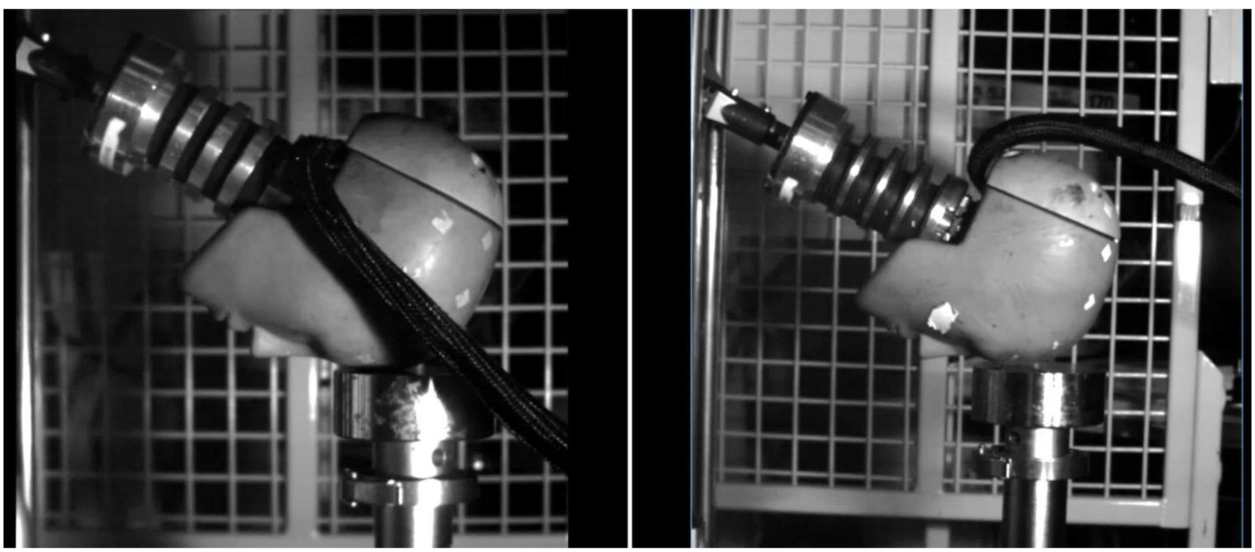

Figure B-1. Monorail drop test set-up using the non-directional neckform (left) and standard neckform in conjunction with the Hybrid III headform.

B.2. Results

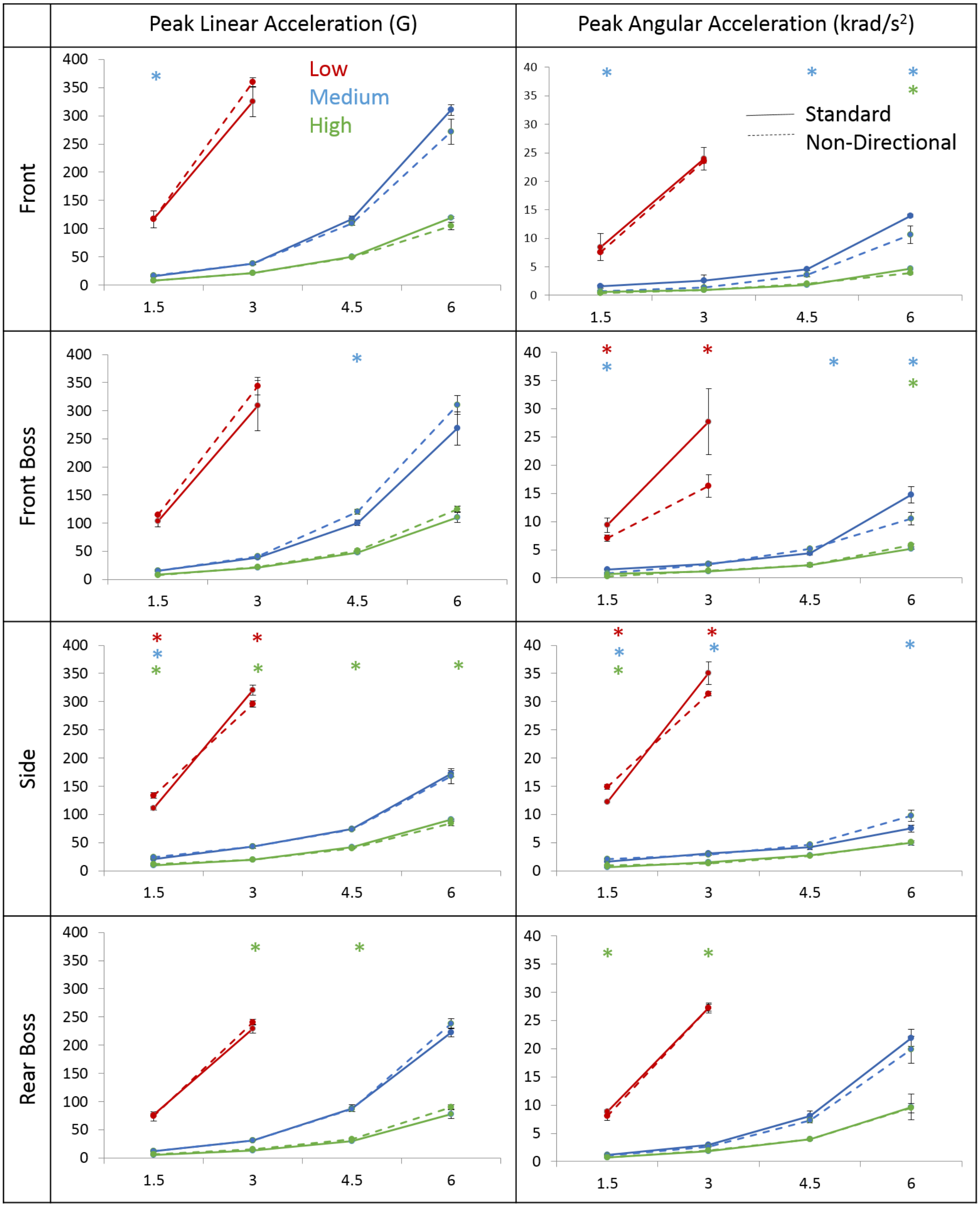

Figure B-2. Peak linear and angular acceleration results for the standard Hybrid III neckform and the non-directional neckforms for 4 impact locations, 4 velocities, and 3 compliances (low, medium, high). Statistically significant comparison are indicated with the appropriately coloured asterisk, representing the compliance level and velocity for which they were determined different ($p<0.05$).

B.3. Conclusions

While neck-type contributed to significantly different peak linear and angular acceleration responses, many of these differences were small and the neckforms produced comparable responses. Peak angular accelerations at the front boss location under low compliance produced the greatest differences between the two neck types (Hybrid III neck: 30 krad/s^2, non-directional neck: 15 krad/s^2) and is likely due to the slits in the standard neckfrom contributing greater extension during an oblique type of impact (mid-point between the front and side locations). These data provide a full comparison between the standard and non-directional neckforms demonstrating comparable results for neck types. This dissertation used the non-directional neckforms in conjunction with the standard headforms to remove any potential effects of the standard necks.

C Finite Element Model Comparison

University College Dublin Brain Trauma Model | Wayne State University Brain Injury Model

C.1. Purpose

To compare peak strain responses from the University College Dublin Brain Trauma Model (UCDBTM) to the Wayne State University Brain Injury Model (WSUBIM) using a subset of collision impact data collected in this thesis. A single trial of medium compliant impacts the rear boss location covering 4 levels of velocity and 4 levels of impact mass were processed using the WSUBIM for a total of 16 simulations. This data was compared with the UCDBTM results.

C.2. University College Dublin Brain Trauma Model (UCDBTM)

The University College Dublin Brain Trauma Model (UCDBTM) was developed by Horgan and Gilchrist (2003). The model includes the scalp, three layered skull, dura, falx cerebri, tentorium, pia, cerebrospinal fluid, brain stem and brain tissue distinguished into grey and white matter. The model consists of 26, 000 elements. The model was validated using cadaver intracranial pressure data (Nahum et al., 1977; Trosseille et al., 1992), brain motion data (Hardy et al., 2001). The material properties are reported in Table-C 1.

Table-C 1. UCDBTM material properties.

Material	Shear Modulus (kPa)		Decay Constant	Bulk Modulus	Density
	G_0	**G_∞**	**(s^{-1})**	**GPa**	**(kg/m^3)**
Grey Matter	10	2.0	80	2.19	1040
White Matter	12.5	2.5	80	2.19	1040
Brain Stem	22.5	4.5	80	2.19	1040
Cerebellum	10	2.0	80	2.19	1040

C.3. Wayne State University Brain Injury Model

The Wayne State University Brain Injury Model (WSUBIM) was developed by (Zhang et al., 2001). The model was created to reflect a 50th percentile male head, including the scalp, three layered skull, dura, falx cerebri, tentorium, pia, sagittal sinus, transverse sinus, cerebrospinal fluid, lateral and third ventricles, brain tissue distinguished into grey and white matter, the cerebellum, brainstem, as well as bridging veins. The model consists of 314,500 elements. The model was validated using cadaver intracranial pressure data (Nahum et al., 1977; Trosseille et al., 1992), brain motion data (Hardy et al., 2001) and facial bone impacts measuring impactor penetration and acceleration from Nyquist et al. (1986) and force and displacement data from Allsop et al. (1988). The material properties are summarized in Table-C 2.

Table-C 2. WSUBIM material properties.

Material	Shear Modulus (kPa)		Decay Constant	Bulk Modulus	Density
	G_0	G_∞	(s^{-1})	GPa	(kg/m^3)
Grey Matter	10	2.0	80	2.19	1060
White Matter	12.5	2.5	80	2.19	1060
Brain Stem	22.5	4.5	80	2.19	1060
Cerebellum	10	2.0	80	2.19	1060

C.4. Results

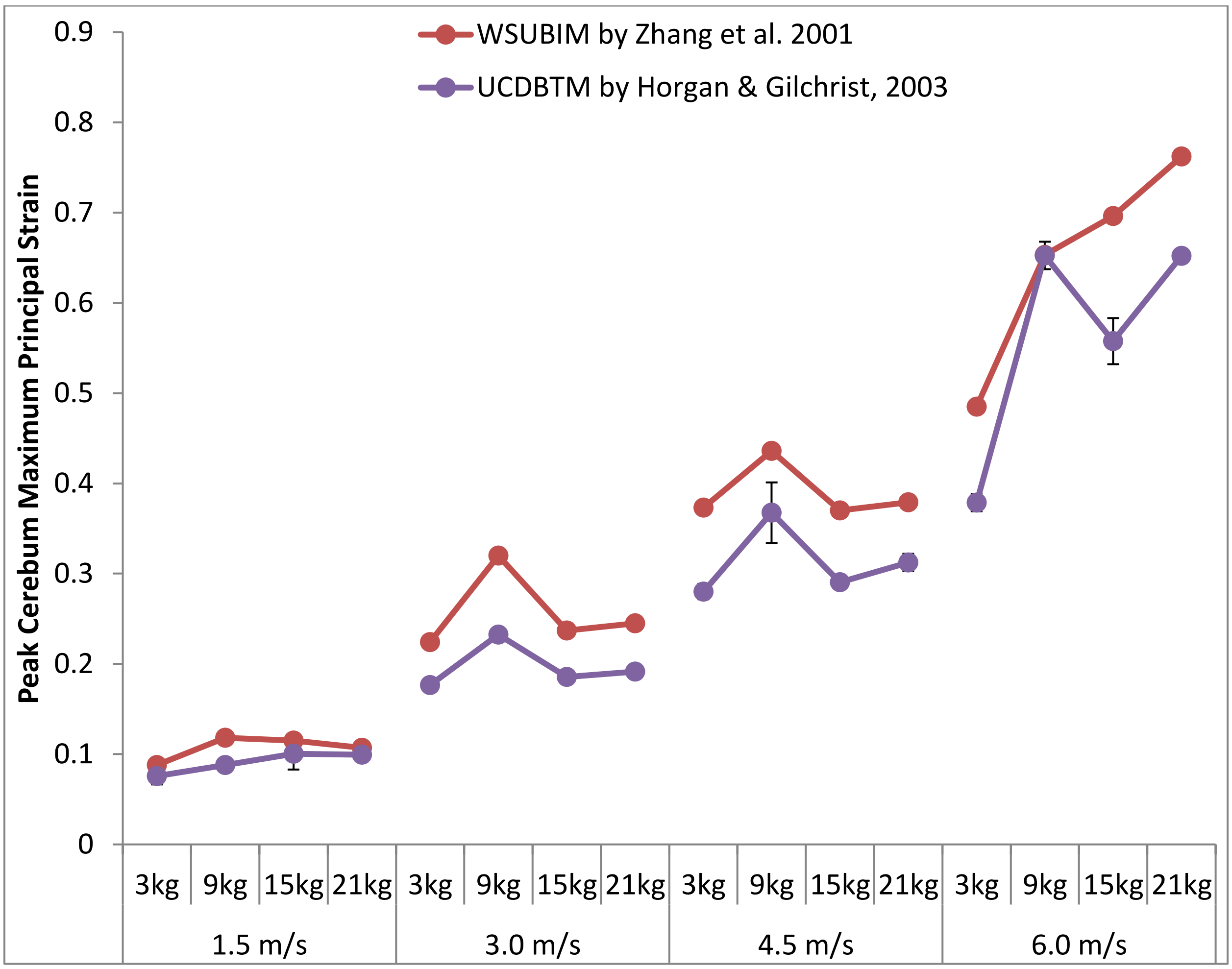

Figure-C 1. Comparison of peak maximum principal strains in the cerebrum of the WSUBIM (red) and UCDBTM (purple).

C.5. Conclusions

The UCDBTM is comparatively smaller (26 000 elements) than the WSUBIM (300 000 elements) but this allowed for faster computation times than the much larger models (30 minutes versus 5 hour run times). This was an important consideration given nearly 900 impacts were conducted in this thesis and processed using FE. The UCDBTM had a tendency to report lower peak strains than the WSUBIM, and therefore it can be concluded that the UCDBTM reports more conservative values (Figure-C 1).

D Definitions

Bulk Modulus	is a measure of the compressibility of a material in Pascal.
Decay Constant	Rate constant that defines the diminishing effect of a constant load (s^{-1})
Density	Physical property of material and is mass of material per area (kg/m^3)
Poisson's Ratio	Describes shape/geometry change. Calculated by ratio of the change in length to change in thickness.
Shear Modulus	Describes the shear stiffness of a material to shear loading. Shear modulus is represented by two constants and measured in Pascal. G_0 characterizes the stored energy/elastic portion of the material. G_∞ characterizes the amount of energy lost/dissipated as heat and is also known as the viscous portion.
Young's Modulus	Describes the relative strength of a material and is measure of material stiffness. It is characterized by the ratio of stress to strain (slope of the line in stress-strain curve).

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