Impact Characteristics Describing Concussive Injury in Youth

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ABSTRACT:

The incidence of concussive injury has continued to arise annually with up to 3.8 million concussions reported per year (Thurman 1999) and 15% of these injuries occurring with persistent symptoms (Kraus and Chu, 2005). Few studies have examined the differences between youth and adult concussion (Yeates et al, 2012; Gosselin et al, 2010) therefore it is unknown whether youth and adults pose a similar risk of sustaining a concussion following impact. For this reason, the purpose of this study is to determine if differences exist in the dynamic response of the head and brain tissue deformation characteristics between children and adolescents for falls in comparison to adult data which have resulted in concussive injuries.

Patient data was collected from emergency room hospitals across Canada. After exclusion criterion was applied, 11 child and 10 adolescent falls were reconstructed using mathematical (MADYMO) model, physical model (Hybrid III Headforms) and finite element modelling. Both groups were compared to each other as well as an adult group collected by Post et al (2014b) using a one-way ANOVA and Welsh test. The results of this study show that the children produced the lowest values for all variables when compared to the adolescents and adults whereas the adolescents produced the largest (with the exception of MPS where the adolescent and adult MPS was the same). Although all results were above the suggested thresholds for risk of concussive injury, the youth produced the lowest brain tissue strain yet still suffered a concussion. This is important to note as it may suggest that children are at an increased risk of injury at a lower brain tissue strain level. Understanding the differences in parameters influencing concussive injury may aid researchers in comprehending the unique risk for youth at difference ages. This information would be useful in terms of protective equipment design, promoting safe play in games and management of patients following injury.

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LIST OF EQUATIONS:

Equation 1: Shear characteristic of the viscoelastic behavior of the brain

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

Equation 2 : Hyperelastic Law

$$C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15} \text{ (Pa)}$$

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CHAPTER 1: INTRODUCTION

1.1 OVERVIEW

Reported incidence of mild traumatic brain injury (mTBI) commonly known as 'concussion' has been increasing in both the youth and adult populations. It has been estimated in the United States alone that sport or recreational play related concussions have reached 3.8 million people per year with up to 15% of these injuries resulting in persistent symptoms (Langlois et al, 2006; Kraus and Chu, 2005). For youth specifically, up to half a million mTBI cases occur each year with falls being the leading mechanism of injury for mTBI injuries (Kuppermann et al, 2009). Furthermore, studies focusing on high school athletes and have found that the adolescent athletes may be at the highest risk of prolonged concussion when compared to collegiate and professional aged athletes [Covassin et al, 2012; Field et al, 2003]

Concussion is characterized as a subset of traumatic brain injury caused by biomechanical forces translated to the brain and diagnosed by symptomology (McCrory et al, 2009; McCrory, 2013; Kutcher et al, 2010). The risk of concussion at different stages of youth is not fully understood as concussion is currently diagnosed by symptomology. This creates an additional challenge for diagnosing young concussion patients, as they may be unable to recognize or communicate their symptoms. For these reasons, concussive injury in youth is a complex injury to prevent, diagnose and treat within sport and primary care settings (Rivera et al, 2015).

In the majority of concussive injuries, the symptoms following impact are transient and resolve on their own within 7-10 days; however, in up to 15% of cases, persistent symptoms may last longer than 4 weeks (Kraus and Chu, 2005). If these periods of lasting deficits occur four-six weeks post injury, the injury is defined as post concussive syndrome (PCS) (National Institute of Health, 1999; McCrory et al 2013, Zemek et al, 2013). In these chronic cases, lasting deficits may occur, which can affect the individuals working life or in the case of youth affect their education, social interactions, and play. (Wrightson and Gronwall, 1998; Bigler, 2008; Kraus and Chu, 2005; Post et al, 2014b).

Youth, and specifically high school aged athletes have been found to have a more challenging recovery than adults in terms of recovery time and symptom severity (Makdissi, et al, 2013; McCrory et al, 2013; Covassin et al, 2012; Field et al, 2003; Sim et al, 2008). Youth tend to have a less positive recovery time from a cognitive perspective, which may stem from the fact that they are in a period of development and constantly learning (Field et al, 2003; Pellman et al, 2006; Gioia et al, 2010; Bruce et al, 1981). This is different from adults who have often mastered the skills which are needed to function in everyday life. The demands of constant learning may encourage persistent symptoms in the brains of youth, as their brains may not be able to rest to the same degree as an adult brain. This inability to give the brain sufficient rest may explain the prolonged and/or more challenging recovery in youth.

In order to minimize the disruption in youth's emotional and academic development that could impact their future education, it is important to understand how youth are injured during different stages of development. (Choe et al, 2012; McCrory P, 2011; Ponsford et al, 2011). Multiple studies have begun to analyze injurious impacts by examining the parameters of the impact (Karton et al, 2013; Rousseau and Hoshizaki, 2015; Post et al, 2014b). Defining youth concussions based on parameters such as location of impact, impact velocity and impact surfaces can be essential in the determining risk factors for differing ages of youth. This knowledge may assist in a better understanding of which factors create high-risk scenarios of concussions from impact. Being able to recognize children, who may be at a high risk for injury, could provide health care workers with the ability to properly assess youth and ensure the management of symptoms to facilitate their recovery and minimize the effects on a child's education and social well-being.

Concussions have been primarily studied in adult populations with few studies focused solely on youth populations therefore the biomechanical risk factors for the youth populations remain unknown. Halstead and co-workers postulated children might be at an increased risk of concussive injury and severity of injury because of distinct anatomical and physiological differences during brain development (Halstead and Walter, 2010). Differences such as head mass may influence the characteristics of an impact resulting in a unique risk of injury. Youth have a smaller skull size and mass when compared to adult skulls leading to a decrease in the force/mass ratio (Procter & Cantu, 2000; Webbe, 2003). A decreased skull mass can result in higher accelerations if force is held constant which would increase risk of stress and strain of the brain tissue (Gimbel and Hoshizaki, 2008). Higher accelerations may place children at a unique risk of injury as this increase in acceleration can lead to increased stresses and strains of the brain tissue which is correlated with concussive symptoms (Martin et al, 1994; Post et al, 2015). Additionally, in regards to adolescents, the mass and heights are much closer to that of an adult however research suggests they still may be at a greater risk of sustaining a concussion. This may be an effect of adolescents having anatomical similarities to that of an adult but are not yet as skilled as adults for example in techniques of play (ex. blocking) or falling (Dayan et al, 2010; Pharo et al, 2011). These factors may influence the location that the adolescents are struck along with inbound mass and velocity, which have all been shown to create unique risk of injury (Karton et al, 2013; Rousseau and Hoshizaki, 2015). For this reason it is important to examine the effect of head mass, location of impact, inbound velocity and impact compliance on head accelerations resulting from head impacts contributing to concussive symptoms in youth.

Furthermore, youth as a whole may be at a unique risk for sustaining a concussion compared to adults as they differ from adults in their physiology such as synaptic density, grey and white matter ratios, cerebral blood flow and glucose metabolism which may create differences in the way their brain tissue responds to impact (Bourgeois et al 1994; Huttenlocher and Dabnolkar, 1997; Giedd et al., 2012; Tontisirin et al., 2007; Choe et al 2012, Shrey et al, 2011). Evaluating if differences in tissue strain are present between children, adolescent and adult concussion would add valuable information about the understanding of youth concussion. Additionally, understanding the mechanics, which contribute to the risk of concussive injury in youth such as impact location, velocity, mass and compliance, can be used in the development of new technology and protective sporting equipment to mitigate concussive injury at all ages of youth. An understanding of how these factors affect the biomechanical characteristics of child, adolescent and adult concussive impacts resulting in concussions would guide researchers to design equipment to mitigate the risk of concussive injury for youth.

Examining the impact parameters contributing to the risk of concussive injury in children and adolescents compared to adults may provide important information to decrease the incidence and severity of the injury. Understanding the differences in conditions associated with concussions may aid researchers in understanding the unique risk for youth at difference ages. The purpose of this study is therefore to determine if differences exist between the dynamic response of the head and brain tissue deformation characteristics between children and adolescents for falls compared to adults resulting in concussions.

1.2 RESEARCH QUESTION

Do head dynamic response and brain tissue deformation values differ between head impacts from falls resulting in concussive injuries in children and adolescents when compared to similar events in adults?

1.3 OBJECTIVES

1. Measure the dynamic response values (peak resultant linear and rotational acceleration) in youth and adolescent real world concussion cases.
2. Measure brain tissue deformation in both youth and adolescent concussion cases.
3. Measure rotational Velocity for the youth, adolescent and adult concussion cases
4. Compare youth PCS cases and adolescent concussion cases to adult concussion-PCS cases for fall events (compliance, mechanism).

1.4 VARIABLES

1.4.1 INDEPENDENT VARIABLES

- Child case studies 10 concussion
- Adolescent case studies 10 concussion
- Adult case studies 10 concussion

1.4.2 DEPENDENT VARIABLES

- Dynamic Response
  - Peak Resultant Linear Acceleration (g)
  - Peak Resultant Rotational Acceleration (rad/s2)
  - Rotational Velocity (rad/s)
- Brain Tissue Deformation
  - Peak Maximum Principal Strain

1.5 EXPERIMENTAL HYPOTHESIS

It is hypothesized that dynamic response and brain tissue deformation values will be lowest for youth concussion reconstructions, followed by the adolescent age group and highest for adult concussion reconstruction cases.

1.6 NULL HYPOTHESIS

No difference will be found between adult, adolescent and youth concussion cases for all dependent variables

1.7 SIGNIFICANCE

The proposed study will undertake injury reconstructions of real life events resulting in concussions to provide a comparison of the dynamic response of the head and brain tissue strains for children, adolescents and adults. This is significant as concussive injuries in youth often result in missed school, which can create delays in learning and development. It is important to understand the factors contributing to concussive symptoms in different stages of youth, as this information will provide guidance for developing protocols for management and prevention of concussive injuries in youth.

1.8 LIMITATIONS

1. The hybrid III headforms to be used in this study are commonly used in head impact testing; yet, they are not designed to be biofidelic. This may cause the head acceleration values to be lower as there is a 1-inch skinform, which may act as a buffer. Additionally it is a rigid system compared to the human body therefore the results do not allow for variance which is commonly seen between humans.

2. Since a bare headform will be dropped on concrete, the upper velocities of some cases may not be reconstructed to protect the equipment.
3. The maximum principle strain will only represent the greatest strain value of all three axis therefore it is not a global representation of the strain on the entirety of the brain.

1.9 DELIMITATIONS

1. Cases selected for this study will be selected from across Canada and may not be representative of the world population.
2. The material properties of the regions of the head for the finite element model determine the response of the brain however there is variation of material properties in the literature causing no definitive properties to be established.
3. There is currently no child specific finite element model to use therefore the model used in this study will be a scaled adult finite element model

CHAPTER 2: LITERATURE REVIEW

Increased awareness around concussion has encouraged an influx of research surrounding the causes and parameters influencing risk of concussive injury. However, few studies have been done surrounding youth concussion (Yeates et al, 2012; Gosselin et al, 2010) especially in the field of biomechanics. Understanding the relationships between biomechanical parameters influencing risk of concussions would be useful in terms of protective equipment design, promoting safe play in games and management of patients following injury.

This review summarizes current knowledge of the developing brain in terms of anatomy, physiology as well as the biomechanics of concussion. Lastly, the use of dependent variables (peak resultant linear and rotational acceleration, and brain tissue deformation) and finite element modeling will be discussed in terms of their relevance in identifying the risk of injury.

2.1 EPIDEMIOLOGY

An estimated incidence of concussions per year is up to 3.8 million in the US alone (Thurman 1999). However, many studies suggest that concussions often go underreported for several reasons including the person and/or parent/coach does not recognize the signs of a concussion. Additionally, in the case of athletes, they may deliberately under report so they are able to continue to participate in sport (McCrea et al, 2004; Gerberich et al, 1983; Valovich McLeod et al, 2007).

Studies have shown that a large percentage of youth and young adults recover fully from concussions especially in sport related settings or uncomplicated injuries such as a first concussive impact (Carroll et al 2004; Satz 2001; Yeates et Taylor, 2005) however, recovery in up to 15% of concussions have reported persistent symptoms (Kraus et Chu, 2005). If this period of persistence lasts longer than the 7-10 day transient period, the individuals are diagnosed with persistent concussive syndrome. Although studies have shown up to 15% of adult cases are classified as PCS, the risk of persistence in youth has yet to be examined therefore it is unknown whether youth and adults pose a similar risk of sustaining persistent symptoms following impact. Zemek and coworkers have reported the adolescent population at potentially the highest risk of persistence compared to other age categories (Zemek et al, 2015).

It may be postulated that because youth differ in anatomy and physiology compared to adults, youth may be at a unique risk of concussive injury (Halstead and Walter, 2010). Additionally, youth may be at a higher risk of a challenging recovery as their brain and skull is still immature therefore an impact during this time may cause a significant impact on their ability to learn (Graham, 2014, pg. 218). At this moment, no known biomechanical comparison has been done to identify the unique risk of concussive injury in youth (children and adolescent) compared to adults therefore research is needed better to understand the parameters that influence concussion specific to all age categories.

2.2 BIOLOGICAL DIFFERENCES BETWEEN YOUTH AND ADULTS

Millions of youth are engaged in sports and recreational play every day putting them at risk of injury. What is unknown is if youth of differing ages pose a unique risk of concussive injuries compared to adults due to biomechanical differences such as head mass and height or differences related to their developmental stage (e.g. style of play). It has been postulated that the risk for youth may be different to that of adults as youth differ in terms of: anatomy, physiology, and developmental maturity (McCrory et al, 2004). It is currently unknown if these factors influence the outcome of an impact and if they create vulnerability to concussion in youth. For this reason it is important to examine the scenarios associated in creating concussive injuries within each category age category (child, adolescent, and adult).

2.2.1 Biomechanical/ Anatomical Differences

Concussions occur from impacts that can be characterized by velocity, mass, compliance and location of impact on the head. The effects of these parameters have shown to influence the outcomes of an impact in terms of dynamic response and brain tissue deformation (Karton et al, 2013; Rousseau and Hoshizaki, 2015; Post et al, 2014b). Injury occurs when brain tissue exceeds its tolerance threshold therefore characterizing the parameters of an impact may define its risk (Zhang et al, 2004). Previously, Post et al (2014b) have reconstructed PCS cases for adults examining the ranges of these parameters, which created risk for that specific age range however it has yet to be examined for different age categories. It is therefore unknown if children and adolescents exhibit the same risk parameters as adults as anatomical differences may put youth at higher risk for obtaining a concussion.

One observation is that adults possess an increase in mass and height fallen (inbound velocity) in comparison to youth and possibly adolescents therefore adults may be at an increased risk of concussive injury due to higher acceleration and forces from increased mass and velocity creating an increase in brain tissue deformation (Gimbel and Hoshizaki, 2008). However, other research has shown that youth continue to get injured with less force to obtain the same injury as adults yet often take longer to recover when compared to adults (Kirkwood, 2006; McCrory et al, 2004). Additionally, it has been shown that adolescents specifically take longer to recover than an adult even though they may have a similar mass to that of adults depending on age stage of brain development (Kirkwood, 2006). For this reason, it would be important to examine other impact variables that cause concussive symptoms in adolescents such as compliance or location of impact as these variables can influence the severity of concussive injury (Gennarelli et al., 1982; 1987; Kleiven, 2003; Post et al, 2012a). Evaluating the influence of biomechanical and anatomical differences between children, adolescents and adults would aid in distinguishing if a unique risk of injury exists for children and adolescents when compared to adults.

2.2.2 Physiological Differences

Individuals may experience physiological changes in their metabolic cascades in response to concussive injury that in extreme cases could lead to cell death (Giza and Hovda, 2014). The differences in the disturbance of the metabolic cascade of youth versus adults is still not well understood however it can be postulated that youth are at a unique risk because of a difference in physiologic factors from being in a period of brain development (Graham, 2014, pg. 83). These factors may lead to an influence on level of tissue loading following impact which can be associated to impact parameters such as impact mass, velocity, location and compliance. A physiologic factor that may create risk is that youth have a predisposition to cerebral edema and shear injury due to a decrease in myelination in youth, which could increase the effects of concussive injury specific to that population (Cook et al, 2006). This vulnerability to shear strain may result in forces extending further into the brain. This interruption to the pathophysiological cascade has been shown to be age-dependent in TBI cases in the developing brain therefore it could be considered that this relationship would also be seen at the concussion level (Kirkwood, 2006). Additional evidence has been reported in clinical research which supports the notion that physiologic responses of concussions are age-dependent as catastrophic injuries such as second impact syndrome cases are more commonly associated with adolescents specifically when compared to collegiate players (Kelly et al, 1991; Saunders and Harbaugh, 1984). This increase of second impact syndrome following mTBI as well as additional physiological differences may suggest that physiology could play a role in the outcome of concussive injury at different ages compared to adults (Kirkwood, 2006).

2.2.3 Influences of Developmental Stage

A concussion in youth could have additional implications as their brains are still developing, which may impact the child's ability to acquire new knowledge (McCrory et al 2004). Studies involving TBI have focused on how the immature or developing brain is more vulnerable to injury causing a slower recovery rate (Anderson and Moore, 1995; Brookshire et al, 2000; Kirkwood 2006). Specifically, adolescent athletes have been found to obtain persistent neurocognitive dysfunction more often than adults or collegiate athletes which may further suggest that age could influence the risk for prolonged concussions (Covassin et al, 2012; Field et al, 2003). Furthermore the adolescent age range has been regarded as having an increase in risky behaviors as the prefrontal cortex responsible for decision-making is still developing (Pharo et al, 2011), which can lead to an increased risk of injury (Dayan et al, 2010; Pharo et al, 2011). Although there is still discussion on the vulnerability of the immature brain, evidence now points to the fact that youth may be more susceptible to diffuse injury (Kirkwood et al, 2006; Satz P, 1993). This is important as diffuse axonal injury can affect a child's skill acquisition and learning ability (Ewing-Cobbs et al, 2003; Taylor and Alden, 1997). This decrease in skill acquisition and learning ability can lead to disturbances in the child's education, which can have a major implication on the education of the child at all ages.

2.3 MECHANISM OF INJURY

2.3.1 Falls

Specific events have been shown to produce varying risks of injury within real-world environments (Kendall et al, 2012a; Post et al, 2015). The event of a fall is the most commonly seen event in both youth and adult data leading to brain injury they are commonly associated with high magnitude and short durations impacts (Post et al, 2013a; 2014a; Doorly and Gilchrist 2006; Styrke et al, 2007; Kuppermann et al, 2009). Falls have been examined through injury reconstruction research, which uses simulations from a mathematical model (MADYMO) to obtain inbound velocities (Post et al, 2014a) as well as dynamic response (Doorly and Gilchrist, 2006). A study completed by Kendall (2012a) examined falls onto ice and discovered that these impacts generated values which surpassed the thresholds for linear acceleration and MPS which are associated to a high risk of concussive injury (Kendall et al, 2012a; Zhang et al, 2004; Kleiven et al, 2007). These results show that the event of a fall can produce magnitudes that are great enough to create a high risk for concussive injury however it is unknown if all age categories obtain the same magnitudes. Reasons why magnitudes may be influenced could be associated to the inbound velocity or location of impact. For this reason it is important to look at the impact parameters as well as event for each age to determine risk or injury that is age and event specific.

2.3.2 Location of Impact

As mentioned above, location of impact can have an effect on the magnitude of dynamic response and brain tissue deformation responsible for representing head injury (Gurdjian et al 1953; Hodgson et al 1983; Zhang et al 2001a). The side location has been shown to result in a greater risk of concussive injuries than other locations (Kleiven, 2003; Zhang et al, 2004, Delaney et al, 2006). For this reason, impacts should be examined by location for each age category as this may influence the degree of risk.

2.4 PREDICTORS OF CONCUSSION

Difference in risk for traumatic brain injuries at different ages have shown that age may contribute to that risk (Kirkwood, 2006; McCrory et al, 2004). However, it remains unclear if higher head accelerations are needed to produce a mild traumatic brain injury in children and adolescents compared to adults therefore, examining the differences in head accelerations causing concussions in child, adolescent and adult concussion may aid in understanding the risk for injury in youth.

2.4.1 Linear Acceleration

Linear acceleration is used to measure the change in pressure gradient responsible for injury (Gurdjian, 1975). This change in pressure gradients aids in measuring risk of focal or traumatic injury such as skull fractures (Genarelli et al, 1972; Ommaya et al, 1974; Holbourn 1943; King et al, 2003). Additionally, linear acceleration was among one of the first methods for analyzing head impacts along with rotational acceleration later described in Holbourn's study in 1943 (Holbourn, 1943). This method of measuring only linear acceleration was acceptable in animal studies as most of these studies focused on the evaluation of traumatic/focal injuries, which are typically linear dominant injuries (Gurdjian, 1975). This change in pressure gradients aids in measuring risk of focal or traumatic injury such as skull fractures however its relevance in measuring risk of mTBI on its own is debatable (Genarelli et al, 1972; Ommaya et al, 1974; Holbourn 1943; King et al, 2003).

The first studies to examine rotational acceleration came from Gurdjian in 1963, which used mongrel dogs to determine if rotational acceleration alone was enough to produce a concussion. This research deemed rotational acceleration alone was not enough to cause concussive injuries, which was supported by Ommaya in 1966. Furthermore, Ono and colleagues (1980) studied the effect of linear acceleration in monkeys and found a high correlation between linear acceleration and concussion. For this reason, it was stated that measuring linear acceleration alone was enough to capture the risk of concussive injury. Therefore it was unnecessary to capture rotational acceleration to determine the risk of concussive injury from impact despite previous research conducted by Holbourn (1943).

It is important to consider linear acceleration as this measurement is still used in standards which evaluate the safety of equipment to determine their pass or fail criteria as well thresholds for injury have been suggested for concussions based on linear acceleration values of real world data. How this variable correlates with brain tissue deformation has yet to be shown therefore it is important to also look at the rotational characteristics of the impact to fully describe the impact responsible for injury.

2.4.2 Rotational Acceleration

Impacts to the head contain both a linear and rotational components however the rotational component of the impact has been more strongly associated to the mechanism of concussion (Holbourn, 1943; Post et al 2013a,b; Kleiven, 2007). Holbourn first proposed this in 1943 by using a flask full of water to demonstrate the interaction found between the skull and brain. What this showed was the ability of the water to stay in place as the glass rotates around it mimicking the brain and skull interface (Holbourn, 1943). Water particles dissociated from the particles attached to side of the flask allowing the conclusion to be drawn that large shear strains would be caused by rotational forces that would not be seen from translational forces (linear acceleration). Although there is often an interaction of both linear and rotational acceleration, it has been shown that rotation causes the shear stress and strain of the brain tissue responsible for concussive injury within the brain (Holbourn, 1943; Hodgson and Thomas, 1979; Ommaya et al 1974; Genarelli et al, 1972). A study involving brain tissue strain from rotational acceleration was conducted in 1979 by Hodgson and Thomas on monkey brains. This study examined the effect of linear, rotational and a combination of the two on stresses and strains of the brain. This study concluded that it was in fact the rotational acceleration that causes the greatest degree of shear strain and displacement from impact (Hodgson and Thomas, 1979). Rotational acceleration and brain tissue deformation metrics should therefore be captured to fully describe the impact.

Linear and rotational acceleration have been previously used in an attempt to identify risk of concussive injury however they have not been overly successful. Finite element modeling is therefore used to determine strain of the brain tissue, which attempts to account for the limitations of the dynamic response metrics. For this reason, both linear and rotational acceleration will be measured and used as inputs for a finite element brain model.

2.4.3 Rotational Velocity

Alongside rotational acceleration, rotational velocity has been proposed as a mechanism of concussive injury (Holburn, 1943; Takhounts et al, 2008, 2013; Hardy et al, 2001, 2007; Ouckama and Pearsall, 2014). Holburn explained this mechanism is especially important for short duration impacts, as the injury is proportional to the change of rotational velocity. Additionally, this variable has also been suggested as the most convenient approach in measuring brain and skull motion (Hardy et al, 2001). This is important to measure as researchers such as Takounts et al. (2013) found that the only way to strain the brain is to rotate the skull. This study also showed that angular velocity is correlated with maximum principle strain (Takhounts et al, 2013; Ouckama & Pearsall, 2014) therefore, rotational velocity is an important measure to consider when attempting to examine the risk of concussive injury.

2.4.4 Brain Tissue Deformation

The largest benefit to finite element modeling is its ability to calculate an approximate stress of the brain tissue from impact that combined with physical models has the highest correlation in predicting injury (Zhang et al, 2004, Kleiven 2007, Willinger and Baumgartner, 2003). Brain tissue deformation metrics such as maximum principle strain have been utilized in real world reconstruction cases in order to assist in the prediction and analysis of concussive injuries to define risk of injury from impact. Finite Element modeling uses variables such as maximum principle strain (MPS) in an attempt to describe deformation of the tissue from impact. (Bain and Meaney, 2000; Kleiven, 2007). Physical models such as the hybrid III headform were developed to capture head motions using linear and rotational acceleration that were more similar to human response. A limited capacity was noted in regards to the prediction of injury using dynamic response captured from the physical model therefore finite element modelling was then used to estimate the brain strains from impact in order to determine a more appropriate measurement of injury risk. It was found that the measurement of strain of the tissue was more highly correlated to the outcome of concussive injury (Morrison et al, 2003; Elkin and Morrison, 2007).

The correlation between tissue strain and concussive injury corresponds to anatomical and animal model data injury due to a mechanical fault resulting in a disruption in the transmission of signals or an increase in cell death (Bain and Meaney, 2000; Morrison et al 2003). Brain tissue is viscoelastic in nature with a low shear modulus, creating an increase in vulnerability to strain leading to injury (Smith et al, 2000). It is these shear forces, which have been attributed as the likely mechanism for tissue damage resulting in concussion (Meaney et al, 2011). Many tissue studies involving strain result in cell death and shown that the amount of strain applied along with the rate of onset is directly related to the brain tissue's ability to function (Galbraith et al, 1993, Mao et al, 2006).

Although this link had been established, it was unknown how the strains of the brain tissue occur following an impact that result in head trauma. Animal models were then used to bridge the link between the strains of the tissue resulting in injury and the head motion associated with those impacts (King et al, 2003). Peak linear and rotational acceleration were measured using animal models as a way to quantify head motion (Ommaya et al, 1974, Gurdjian 1963). FE models of the human brain were then developed to analyze the effects of linear and rotational acceleration on brain tissue deformation. It was found that a higher correlation was seen between concussion and rotational acceleration (Forero Rueda et al, 2010) therefore it was suspected that rotational acceleration was the main contributor for the severity of stress and strain of the brain tissue (Zhang et al 2006; Kleiven 2006).

Brain tissue strain has been found to be associated with concussive symptoms from anatomical tissue models by correlating rotational head motion causing strain to injury using FE modeling (Bain et al, 2000; Morrison et al 2003; King et al, 2003; Forero Rueda et al, 2010; Zhang et al 2006). For this reason, MPS is considered an important variable to represent the risk of injury from impact. Examining the differences in strain value of the brain tissue between youth, adolescent and adult injury is important in understanding differences in the injuries seen between the populations at the level of the tissue.

2.5.1 Dynamic Response Thresholds

Dynamic response values have been used to represent the amount of force received by the head during impact. Studies have suggested thresholds for injury based on these values by using reconstructions from real world injury events. A study from Zhang et al 2004 created thresholds for injury based upon concussive events from the NFL where physical models were used to reconstruct the events and determine the linear and rotational acceleration of the impact (Zhang et al, 2004). Zhang proposed injury thresholds for linear acceleration in adults for a probability of 25, 50, and 80% chance of concussive injury to be 66, 82, and 106g respectively.

Additional research was then done examining rotational acceleration by reconstructing concussive impacts using adult cadavers as well as reconstructing real-world football concussions in an attempt to identify thresholds for risk of concussive injury in adults (Hardy et al, 2001; Zhang et al, 2004). Adult thresholds of injury are therefore used as an estimation of classifying severity of risk of injury by measuring both linear and rotational acceleration. One set of thresholds for risk of injury for rotational acceleration was established by Zhang (2004) with probabilities of 25, 50, and 80% chance of concussive injury. These values for sustaining an mTBI were estimated for rotational acceleration to be 4600, 5600, and 7900 rad/s2 respectively.

Apart from research by Zhang surrounding thresholds for linear and rotational acceleration, Rowson et al (2012) conducted a study involving collegiate football players using Head Impact telemetry (HIT). They found a 50% risk of concussion when the threshold for rotational velocity reached 28.3 rad/s. McIntosh et al (2014) however conducted a study involving Australian football players using MADYMO in which they determined that a rotational velocity of 22.2 rad/s could be attributed to a 50% risk of concussive injury. Both of these values are important to consider when examining rotational velocity results to predict risk of injury.

These values are reported in concussion research however; studies have begun to use brain tissue deformation measures to describe risk. Brain tissue deformation values are calculated using linear and rotational acceleration as well as the material properties of the tissue, which then gives a representative value of strain, which can be used to provide a more appropriate prediction of risk.

2.5.2 Brain Tissue Tolerance Thresholds

Thresholds corresponding to risk of concussive injury based on tissue level responses have been created by several authors based mainly on professional football impacts (Zhang et al, 2004; Kleiven et al, 2007). Using finite element models, these studies have created risk curves for metrics such as MPS and dynamic response variables. These thresholds for brain tissue strain may be a more accurate method in predicting risk of concussive injury as they represent the strain of the tissue which has been the associated mechanism of concussion (King et al, 2003; Post and Hoshizaki, 2012b). In 2007, Kleiven determined threshold values representing a 50% chance of injury to be 0.21 to 0.26 for strain (MPS) depending on the location in which the injury occurred for adults in the brain tissue. Zhang (2004) proposed 0.14, 0.19 and 0.24 values of strain rate to correlate with a 25, 50 and 80% risk of injury for adults. These values aid researchers in interpreting their results from real world reconstructions to determine a proposed risk of concussive injury for adults however thresholds for youth have yet to be described.

2.6 SUMMARY

Differences between children, adolescents, and adults were described in terms of their anatomical/biomechanical differences, physiological differences, and behavioral differences. Furthermore, this chapter reviewed the possible implications these differences may have on impact parameters such as mass and velocity. Other parameters, which are related to increasing risk of injury, were also discussed in regards to injury event (ex. Falls) as well as impact location. To examine the differences in terms of a biomechanical evaluation, predictors of injury such as peak linear and rotational acceleration were discussed and how these variables are used to calculate brain tissue deformation, which aids in a further understanding of the injury outcome. As noted above, there are few studies regarding the risk of youth concussion and the differences in parameters which contribute to injury between different age groups. Therefore, this study is novel in that it aims to examine the biomechanical differences associated with concussive injury between three age categories using dynamic response and brain tissue metrics. This knowledge will be important to guide improvements in protective equipment, game regulation and patient management that are age-specific. This information may aid future work in the reduction of concussion across varying ages.

CHAPTER 3: METHODOLOGY

3.1 SUBJECT DATA

All patient data was collected from large urban hospital emergency care centers in Canada. Post et al (2014a) characterized concussive impacts presenting with persistent symptoms for adults. This data was used as a comparative group to the child and adolescent concussion reconstruction cases which included both transient and persistent concussive cases. The cases selected from Post et al (2014a) consisted of a fall to a non-compliant impact surface such as concrete or ice to match the mechanism and compliance of the youth cases. This was done to maintain consistency of the injury events between groups to minimize variance from different injury events and impact surfaces when comparing age groups.

3.1.1 Defining Age Category By Head Circumference

Separating the youth cases into child and adolescent data was done using the headform circumference of the 6yr old and 5th hybrid III headform. The 6 yr old headform was to represent the children whereas the 5th percentile headform represented the adolescents. Head circumference of the headforms were compared to anatomical literature to determine the approximate ages and gender that corresponded to each headform (Juliusson PB et al, 2013; Nellhaus G, 1968; Zong X and Li H, 2013). By using head circumference within the literature as a guide, the child category using the 6-year-old headform is representative of females 6-10 years old and 6-7 year olds males. Therefore there is a range of 6-10 year olds for the child category depending on gender (circumference: 0-52.9cm). Secondly, the 5th percentile Hybrid III headform was used for the adolescents and represents males aged 9-16 and age 10-18. For this reason the age range for the adolescent age category is 9-18 depending on gender of the person (circumference: 53.0-56.4cm).

3.1.2 Child/ Adolescent Subject Data

All youth concussion data was collected from 9 of 12 Canadian paediatric hospital's emergency departments. Patient or Parental consent was given to a research assistant before the child was included into the study if the patient was under 7 years old or unable to provide informed consent. Cases were accepted into this study if they had a concussion from a fall mechanism within the last 48 hours as defined by the Zurich consensus statement when first admitted. The Zurich consensus statements is as following "A direct blow to the head, face, neck or elsewhere on the body with an impulsive force transmitted to the head, resulting in one or more of the symptoms in one or more of the following clinical domains (which may or may not have involved loss of consciousness): Somatic symptoms, Cognitive symptoms, Emotional/ behavioural symptoms, and/or Sleep disturbances" (Zemek et al, 2013). Additionally, patients were diagnosed with PCS four weeks post injury as per ICS-10 criteria, which state multiple symptoms (3+) to be present as compared to baseline four weeks post injury. Patients were excluded from the study if they presented with: a Glasgow scale of less than 13, any abnormality on neuroimaging or positive CT findings, neurosurgical intervention, intubation or if intensive care was used, multisystem injuries, severe neurological development delay, no clear history or an incomplete description of the event, or if they were already enrolled in the study. The information needed for injury reconstruction included impact surface, location, estimated velocity, direction and orientation of the impact. Each patient was assessed by an acute concussion evaluation. Follow up sessions via web based or telephone survey were then conducted for patients at 1,2,4,8, and 12 weeks post injury. Patients only returned to the clinic for appointments if they were enrolled in the neuropsych or balance substudies. Furthermore, youth were diagnosed with PCS if three or more symptoms remained at four weeks post injury (Zemek et al, 2013).

3.1.3 Adult Subject Data

All adult concussion cases were diagnosed with persistent symptoms by a medical doctor which presented after a fall mechanism. These cases collected for a study done by Post et al (2014b) were used as a comparative data set to the youth age groups. The subjects in these cases also originated from Canadian hospital emergency rooms. Furthermore, patients and/or witness had to have a full recall of the event, which included: head impact velocity, location, surface, and surface geometry in order to be used for injury reconstruction in a laboratory. In addition to a full recall of the event, all patients had to have presented three or more symptoms lasting longer than 4 weeks, which classified them as having PCS. If a full recall of the event could not be completed, patients were excluded from the study along with any presence of neurological legions found from CT/MRI analysis. One thousand cases were collected with 21 cases selected as acceptable for reconstruction. For this study, only cases involving non-compliant impact surface were included to maintain a common impact surface condition to decrease variance from impact conditions.

3.2 EQUIPMENT

3.2.1 MADYMO

A Mathematical Dynamic Model was used in this study to determine the inbound velocity of the fall impact. This model is commonly used in car accident reconstructions and falls to predict the kinematics of the human before impact (Adamec et al, 2010; Doorly and Gilchrist, 2006; Forero Rueda and Gilchrist, 2009; McIntosh et al, 2014). This mathematical model (MADYMO) has been used in multiple reconstruction protocols to estimate inbound velocity when it was not possible to determine from video analysis (Fredreche and McIntosh, 2007, Post et al, 2014a). This model was used to estimate inbound velocity from the information provided on the reconstruction impact forms from the paediatric hospitals. This method has also been used in a study by Post et al (2013a) in which reconstructions of adult concussion cases were modeled to determine inbound velocity for traumatic brain injuries. The Neurotrauma Impact Science Laboratory form (Post A, 2013a) was used as a guide to determine if the patient fell from its own height or a structure an approximate to then reconstruct the drop height. From this position, the MADYMO model was then run in multiple scenarios of limb positions and possible situations in which the impact may have occurred to obtain a range of plausible velocities. From these simulations, the lowest and highest velocities found were used to drop the hybrid III headforms using the monorail drop rig (Post, 2013).

3.2.2 Monorail Drop Rig

The monorail drop rig system consists of a 4.7m rail in which a drop carriage attaches to carry a Hybrid III headform. This drop carriage was attached by ball bushings to the monorail to reduce friction while carrying the headform to impact. The headforms were released by a pneumatic piston, which then allowed the headform to impact the anvil. The headform was dropped using a guided rail and impact velocity was captured using a photoelectric time gate 0.02m above the impact site.

3.2.3 Hybrid Headforms

A 6yr old or 5th percentile hybrid III headform was used to reconstruct each youth case depending on which age category they fell into defined by gender and age. Both headforms are biofidelic and designed to be used for multiple impacts. The neckform used to attach the headform to the monorail drop rig was an unbiased neck form developed by Neurotrauma impact science laboratory (Walsh and Hoshizaki, 2012). Both headforms were equipped with a 3-2-2-2 array, which aids in capturing both the linear and rotational acceleration of the impact. The accelerometers in this array were developed by Endevco (code: Endevco 7264C-2KTZ-2-300) and sampled at 20Hz. This data collected from the accelerometers was then processed using a DTS TDAS system combined with a CFC class 180 filter. The headforms were instrumented to capture data acceleration in six degrees of freedom. The reference system was: x-axis forward direction, y-axis to the left direction, and z-axis direction upward.

3.2.4 Finite Element Model

For each subject, the linear and rotational acceleration time histories from the monorail drop rig were used as inputs to a finite element model to determine the brain tissue deformation associated with each impact. The model that was used in this study was developed by the University of College Dublin and named the University College Dublin Brain Trauma Model (UCDBTM). This model used CT and MRI imaging to create the geometry of the model, which enclosed 26,000 hexahedral elements. The UCDBTM consists of 10 components: the scalp, skull, pia falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brain stem. The components were taken from anatomical and cadaveric research with further defined its properties shown in table 1 and 2.

Table 1. Finite element model material properties

Material	Young's modulus (Mpa)	Poisson's Ration	Density (kg/m^3)
Scalp	16.7	0.42	1000
Cortical bone	15000	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	1140
Tentorium	31.5	0.45	1140
CSF	-	0.5	1000
Grey Matter	Hyperelastic	0.49	1060
White Matter	Hyperelastic	0.49	1060

To represent the behavior of the brain tissue, large deformation theory was applied and these behaviors were then represented by a linear viscoelastic model. The tissue was characterized as viscoelastic in shear with a deviatoric stress rate dependent on the shear relaxation modulus. The compressive nature of the brain tissue was defined as elastic. Shear characteristic of the viscoelastic brain was represented by the equation:

$$(1) \quad G(t) = G_{\infty} + (G_0 - G_{\infty})e^{-\beta t}$$

where G_{∞} is the long term shear modulus, G_0 is the short term shear modulus and β is the decay factor (Horgan and Gilchrist, 2003). A Mooney-Rivlin hyperelastic material model was used for the brain to maintain these properties in conjunction with a viscoelastic material property in ABAQUS, giving the material a decay factor of $\beta = 145\ s^{-1}$ (Horgan and Gilchrist, 2003). The hyperelastic law was given by:

$$(2) \quad C_{10}(t) = 0.9C_{01}(t) = 620.5 + 1930e^{-t/0.008} + 1103e^{-t/0.15}\ (\text{Pa})$$

where C_{10} is the mechanical energy absorbed by the material when the first strain invariant changes by a unit step input and C_{01} is the energy absorbed when the second strain invariant changes by a unit step (Mendis et al., 1995; Miller and Chinzei, 1998) and t is the time in seconds. The cerebral spinal fluid was represented in the model by solid elements to define the brain/skull interaction. These elements allowed for no separation as the bulk modulus of water and a low shear moduli was used in combination with a sliding boundary. The coefficient of friction between these surfaces was 0.2 (Miller et al, 1997).

Table 2. Finite element model characteristics for different regions of brain tissue

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

The model was validated against intracranial pressure (Nahum et al, 1977), neutral density target data detailing relative brain-skull motion (Hardy et al 2001), as well as further validation against real world TBI cases (Doorly & Gilchrist, 2006).

3.3 RECONSTRUCTION PROTOCOL

Youth cases were selected based on a complete impact reconstruction form containing the surface compliance, mass, height, age and location of impact. All cases chosen were a fall mechanism and to a concrete impact surface to exclude the influence of compliance between cases.

A mathematical dynamic model (MADYMO) simulation to determine an appropriate range of inbound velocity was used as no video was collected for velocity analysis (Doorly and Gilchrist, 2009). The drop height input variable used in the MADYMO model was taken from the impact reconstruction form (Post et al, 2014b). From this position, the MADYMO model was then run in multiple situations in which the impact may have occurred to obtain a range of plausible velocities (Doorly and Gilchrist, 2009). From these simulations, the lowest and highest found velocities were used to drop the hybrid III headforms using the monorail drop rig (Post et al, 2014b).

A 6yr old hybrid III head and neckform was used to represent a children's head mass and a 5th percentile hybrid III headform was used for the adolescents. These headforms were designed for multiple impacts and were equipped with Padaonkar's 3-2-2-2 array (1975), to capture both the linear and rotational acceleration of the impact (Oeur et al, 2015; Post et al, 2012a). The headforms were then attached to a monorail drop rig and dropped at both velocities obtained using the MADYMO model onto a concrete anvil (Post et al, 2012a; Zhang et al, 2004; Oeur et al, 2015; Doorly and Gilchrist 2009). Resultant peak linear and resultant peak rotational acceleration values were captured for each impact. The acceleration curves were then used as input into a finite element model of the human brain to determine the stress and strain of the brain tissue from impact.

The finite element model used for the children was a 90% scaled model of the original adult University of Dublin brain trauma model (UCDBTM) and a 95% scaled model for the adolescents. This model was developed using an adult CT and MRI imaging and scaled to represent the size and mass of youth. The model is used to predict the risk of concussion through brain tissue strain which is thought to be more informative as it has been shown to have a higher correlation with concussion than dynamic response itself (Zhang et al, 2004; Kleiven, 2007; Post et al, 2012a). The adult model was validated using cadaveric and real world TBI reconstruction cases and scaled to be appropriate in evaluating youth concussion injury reconstructions as a model representing the physiological differences of youth has yet to be developed (Horgan and Gilchrist, 2003; Nahum et al, 1997; Hardy et al 2001; Doorly & Gilchrist, 2006). The outputs from the model along with dynamic response variables were then compared to adult fall data (Post et al, 2014b).

3.4 STATISTICAL ANALYSIS

For the purposes of this study, only the results from the low velocity were used for statistical analysis. The low velocity results were chosen, as this would most likely be the lowest value plausible to create this injury (Post 2013a; Post et al, 2014b). Four one-way ANOVA's were conducted to determine if significant different exist between all three age categories for peak linear and rotational acceleration, rotational velocity, and maximum principal strain. The test of homogeneity of variance showed a violation for rotational acceleration, rotational velocity, and maximum principle strain. For this reason, the Welch test for unequal variance was performed. Significant main effects were found for linear acceleration using an ANOVA and rotational acceleration, rotational velocity and maximum principle strain using the Welsh test. A Post-hoc tukey test was then performed to determine where the significance lied. The alpha level was set at $p < 0.05$. The analyses will be conducted using the statistical software package SPSS 21 for Windows (IBM Inc., Armonk, NY, USA).

CHAPTER 4: RESULTS

Once the exclusion criteria were taken into consideration, 11 cases were accepted for the youth group, 10 cases for the adolescent age category, and 11 cases for the adult age category. In all cases, the head impacted concrete and represented a range of symptoms, impact velocities, and impact locations. A full description of each event can be found in Appendix B along with the location of impact in Appendix C. These cases then underwent physical reconstruction and finite element analysis to obtain dynamic response and brain tissue deformation metrics. The dynamic response results (peak resultant linear and rotational acceleration and rotational velocity) from physical reconstruction testing, finite element modeling and the statistical analysis are presented in the following section.

Four one-way ANOVA were conducted to determine if differences existed between falls causing concussive injury in three varying age groups for each dependent variable. Participants were classified into three groups: children (n=11), adolescents (n=10) and adults (n=11). For all variables, significant main effects were found at the $p<0.05$ level using an ANOVA for linear acceleration and the Welsh test for rotational acceleration, velocity and MPS as homogeneity was violated. Post hoc comparisons were then completed with Tukey HSD for peak resultant linear and rotational acceleration, rotational velocity and maximum principle strain.

4.1 DYNAMIC RESPONSE

The results for dynamic response for all age categories can be found in Appendix A along with a full description of the event (Appendix B). Table 3 contains the means of all age categories for each dependent variable for comparative purposes. It is important to note that all cases were reconstructions of real world events therefore events varied in terms of impact location and inbound velocity. The Means of each group were calculated from all cases containing three trials each. Standard deviation can be found in brackets (Table 3).

Table 3. Comparison of Average Results between Age Groups

Measurement	Children	Adolescents	Adults
Velocity (m/s)	2.75 (0.69)	3.70 (0.83)	3.53 (0.86)
Linear Acceleration (g)	211.1 (73.7)	319.1 (99.9)	286.5 (75.1)
Rotational Acceleration (Rad/s^2)	15232 (12165)	28814 (10328)	20057 (4409)
Rotational Velocity (rad/s)	24.0 (13.4)	36.3 (12.7)	29.1 (7.6)
MPS	0.465 (0.171)	0.626 (0.195)	0.630 (0.106)

4.1.1 Peak Resultant Linear Acceleration

For peak resultant linear acceleration, significant main effects were found using an ANOVA [$F(2,93)=14.14$, $p<0.0001$]. Tukey post hoc analysis for peak resultant linear acceleration revealed that the children had significantly lower values (M=211.1g, SD=73.7, $p<0.0001$) compared to the adolescents (M=319.1g, SD=99.9) and adult category (M=286.5g, SD=75.1, $p=0.001$) however the adolescent group was not significantly different than the adult age group (Figure 1) ($P=0.270$).

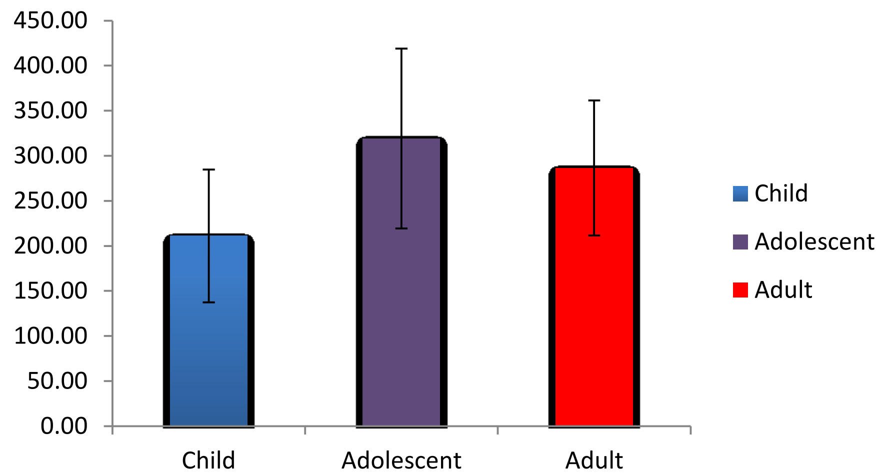

Figure 1. Peak Resultant Linear Acceleration Results (g)

4.1.2 Peak Resultant Rotational Acceleration

For peak resultant rotational acceleration, the test of homogeneity was violated therefore a Welch test was run. Significant main effects were found between all three age categories [$F(2,50)=12.77$, $p<0.0001$]. Tukey post hoc analysis for peak rotational acceleration revealed that the adolescent group produced significantly higher values (M=28814, SD=10328) rad/s^2, which was significantly greater than the child, (M=15232, SD=12165, $p<0.0001$) rad/s^2 and adult category (M=20057, SD=4409, $p=0.001$) however, no significant difference was seen between the children and adult age group (Figure 2) ($p=0.105$).

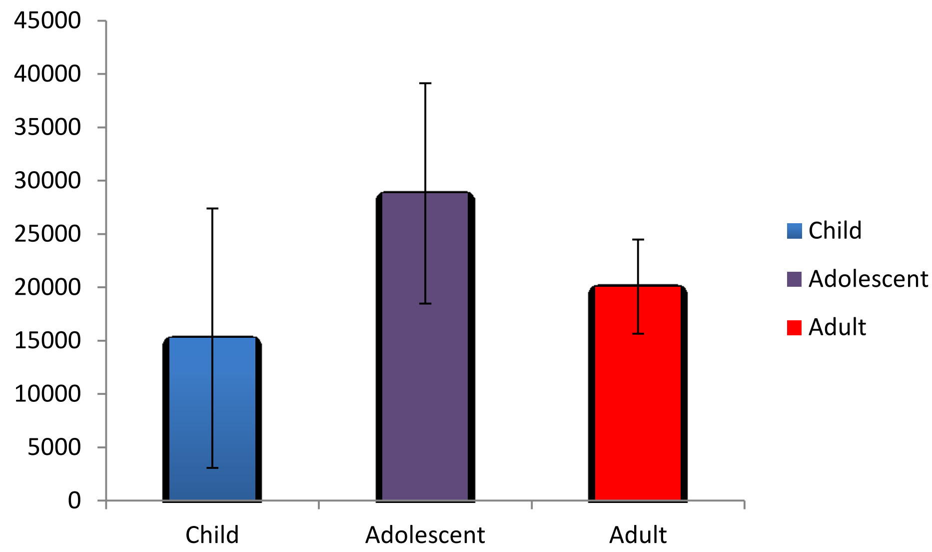

Figure 2. Peak Resultant Rotational Acceleration Results (rad/s^2)

4.1.3 Rotational Velocity

For rotational velocity, significant differences between groups were found to follow the same trends as seen in the mean results of peak rotational acceleration. Significant main effects were found between all age categories after a Welch test was used to correct homogeneity of variance [F(2,56) =6.93, p=0.002]. Tukey post hoc analysis for rotational velocity revealed that the child category produced significantly lower results (M=24.0 rad/s, SD=13.4) compared to the adolescent category (M=36.3, SD=12.7, p<0.0001). Additionally, the adolescent group was significantly higher that the adult category (M=29.1, SD=7.6, p=0.041) however no significant difference was seen between the child and adult age group means for rotational velocity (p=0.177) (Figure 3).

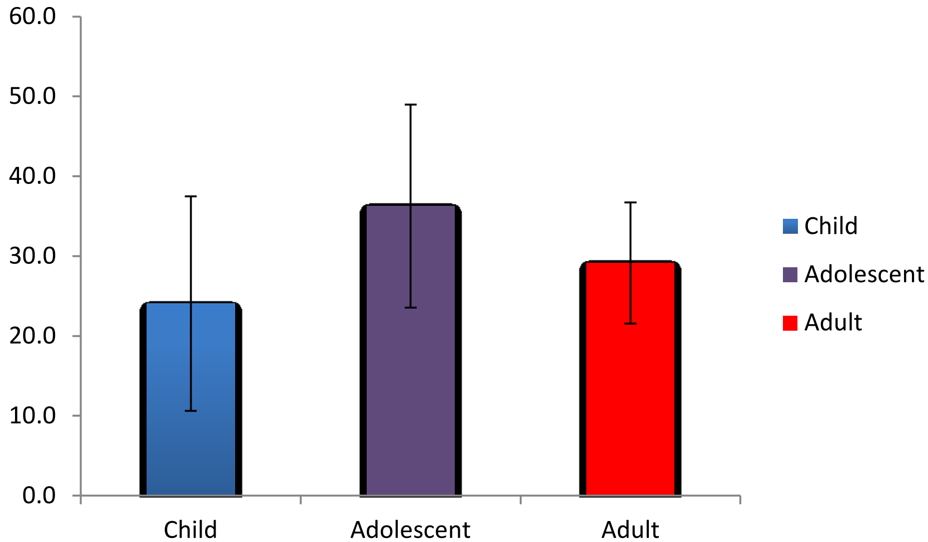

Figure 3. Rotational Velocity Results (rad/s)

4.2 MAXIMUM PRINCIPLE STRAIN

Maximum principal strain was calculated from the finite element analysis using the dynamic response data from impact (Table 4-6) for all age categories, which then were compared using a one-way ANOVA. The results of this ANOVA found significant main effects between age categories however the homogeneity test of variance was violated therefore a Welch test was performed [$F(2,56)=11.55$, $p<0.0001$]. Tukey post hoc analysis of MPS calculated significantly lower strain values for the child group (M= 0.465, SD=0.171) than the adolescent (M=0.626, SD=0.195, $p<0.0001$) and the adult group (M=0.630, SD=0.106, $p<0.0001$). The adolescent group was not significantly different than the adult group for this dependent variable ($p=0.994$). The means of all cases (Table 3) are presented in figure 3.

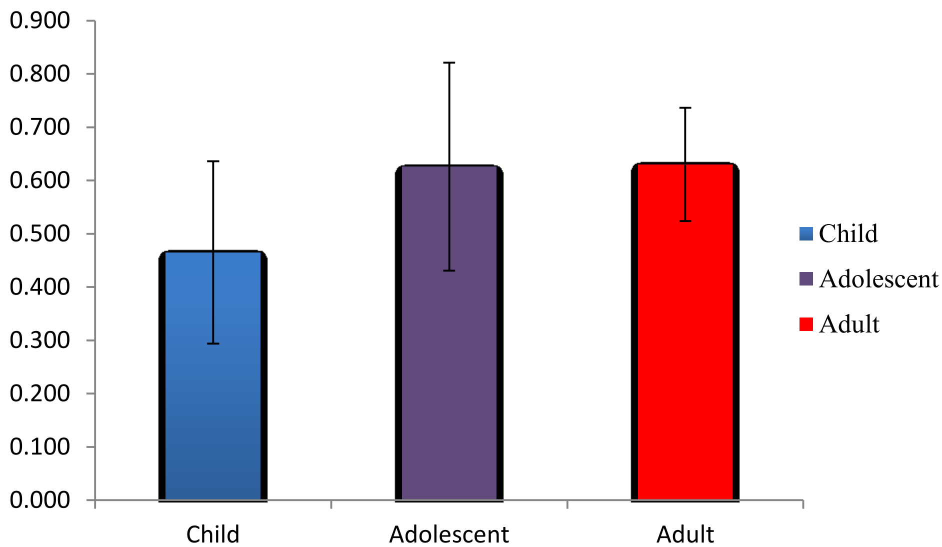

Figure 4. Brain Tissue Deformation Metric (MPS) results

4.3 INBOUND VELOCITY

Inbound velocity was captured for each impact by a time gate on the monorail drop rig. The velocity for between cases varied therefore a range of velocities was seen for each age category (Table 4-6). For the child category, velocity ranged from 1.73m/s to 4.05ms. For the adolescent age category, velocity ranged from 1.85 m/s to 4.90 m/s and the adult ranged from 1.23 m/s to 4.77 m/s. The mean and standard deviation for each category is listed in Table 3.

4.4 DISTRIBUTION OF LOCATION

The location was recorded for each event. The spread of impact locations per age category is reviewed in Appendix C. For the child events, the breakdown of locations consisted of 5 to the front, 4 to the side and 2 at the rear. For the adolescents, 1 impact was seen to the front, 6 to the side and 3 to the rear. Lastly for the adults, 1 was to the front, 2 to the side and 8 to the rear of the head. A full description of the event along with the exact location can be found in Appendix B and C.

CHAPTER 5: DISCUSSION

In order to understand situations which are associated with a risk concussive injury, it is important to evaluate the events that contribute to the outcome of a concussion across multiple age categories. For this reason, the biomechanical parameters of falls to concrete causing concussions in three age categories of children, adolescents and adults were examined in this study. It was discovered throughout all dependent variables in this study that the child age category produced the lowest accelerations, velocities, and brain tissues strains when compared to the adolescent and adult groups yet these values were still were associated with concussive symptoms.

Concussions can occur from moderate to high magnitudes of impact which are created by the characteristics of the impact such as the compliance of the impacting surface, mass of the head, and inbound velocity (Post and Hoshizaki, 2012; Karton et al, 2013). Persistent symptoms are often associated with higher magnitude events (Post et al, 2014a). These magnitudes may lead to a mechanical fault or physiological response of a metabolic cascade that is associated with an increased length and severity of symptoms (Bain and Meaney, 2000; Morrison et al, 2003; Galbraith et al, 1993; Mao et al, 2006). Although all age categories were considered high magnitude events, the children results were lower than the other age groups. This may suggest that the events associated with creating this cascade of response leading to concussive symptoms may be produced at lower trauma magnitudes in children. The following section will outline factors that may contribute to the differences between age groups for each variable followed by the possible implications of these results on risk of injury.

5.1 DYNAMIC RESPONSE

The results of this study showed that the events causing concussive symptoms in children had significantly lower peak linear resultant acceleration magnitudes than the adolescents and adults (who were not significantly different from each other). The results of this variable may illustrate differences regarding the impact energy required to produce concussive injury between groups (energy=$1/2mv^2$). The dissimilarity in energy may be a result of a range of velocities seen between groups along with differences in head masses. As shown in table 3, differences in inbound velocity were seen between ages. The average velocity in which a child was injured was significantly lower (2.75m/s) compared to the adolescents (3.70 m/s) and adult age group (3.42 m/s) ($p<0.05$). This phenomenon of inbound velocity influencing the results of dynamic response has been shown in other studies, which have reconstructed fall events causing injury (Post et al, 2012a; Kendal et al 2012b). The differences in velocities captured may have been a result of the differences in height as most cases fell from their own height. The fact that most children fell from their standing height or lower may indicate that a fall from a play structure or far distance creating a high magnitude response was not required to result in concussive symptoms in this study. Lastly, significant differences were seen between the child and the adolescent/adult category however, no difference was seen between the adolescents and adults. The absence of significant differences between the adolescent and adult group may be a result of their comparable mass and velocities, which created similar environments resulting in no differences in accelerations.

The results for peak rotational acceleration and peak rotational velocity showed a significantly greater response in the adolescent group compared to the child and adult age groups. However, the child and adult groups produced rotational acceleration and rotational velocity results that were not significantly different from one another. Further analysis into these groups discovered that location of the impact might have contributed to the variance seen within these results. For the child and adult category, the majority of impacts were to the front and back of the head (8/11). However, the majority of impacts for the adolescents were to the side of the head (6/10) (Appendix C). Multiple studies within biomechanical research demonstrate that the location of impact can have a large influence on the dynamic response results (Gennarelli et al., 1982; Zhang, Yang & King, 2001b; Kleiven, 2003; Pellman et al, 2003). Furthermore, side impacts in particular have been shown to be associated with higher magnitudes of response (Gennarelli et al, 1982; Kleiven, 2003; Zhang, Yang & King, 2001b; Zwahlen et al, 2007; Delaney et al, 2006). This increase in magnitude of response at the side location has been captured in not only anatomical but also physical testing with impacts to the Hybrid III headform (Post et al, 2013b; Walsh et al, 2011). For this reason, the variance between the age categories could be attributed to the locations in which the headforms were impacted. Therefore, when evaluating differences between groups resulting in injury, varying locations of impact should be considered.

5.2 MAXIMUM PRINCIPLE STRAIN

In this study, the results of MPS show that the child category produced significantly lower values than the adolescents and adults, with the adolescents and adult groups being statistically similar. These results indicate that significantly lower brain tissue strains were experienced by the youth, in comparison to the adolescents and adults. This lower brain tissue strain resulting in injury may suggest that children can be injured at lower levels in comparison to the adolescent and adult age groups. Therefore, this data suggests that children may be more susceptible to concussive injury at lower magnitudes compared to other ages under similar conditions.

5.3 RISK OF INJURY

The values for each case were greater than those reported in the literature that are representative of the thresholds for risk of concussive injury. The results of this study show that all age categories produced results that place them above a 50% risk for concussive injury threshold for all variables. For linear acceleration, a range of risk values were found within literature to be from 90g-146g which were much lower than the averages in this study which ranged from 211.1-319.1g (Duma et al, 2005; Brolinson et al, 2006; Schnebel et al; 2007; Frechede and McIntosh, 2009; Gurdjian et al, 1966; McIntosh et al, 2014). This was also seen for rotational acceleration with literature ranges from 3000-8020rad/s^2 yet the values from this study reached from 15232-28814 rad/s^2 (Willinger and Baumgartner (2003); Frechede and McIntosh, 2009; Ommaya et al, 1967; McIntosh et al, 2014). Furthermore, for peak linear and rotational acceleration, all age categories produced results with an associated risk of mTBI above 80% (Zhang et al, 2004). For rotational velocity, a range of 22-28.3 rad/s was presented in literature to represent risk of concussive injury (Rowson et al, 2012; Pincemaille et al, 1989; McIntosh et al, 2014) and the values reported in this study ranged from 24.0-36.3 rad/s. In regards to the MPS values produced from impact, all results ranging from 0.464-0.630 were associated with a high level of risk for concussive injury within literature (0.14-0.24) (Zhang et al, 2004; Kleiven 2007, Willinger and Baumgartner, 2003). As all values exceeded reported thresholds for risk of injury, this may be representative of the environment of the impact (falls to non-compliant surfaces), which has been reported to create higher magnitudes of response (Dawson et al, 2013).

The results of this study are above the reported range for risk of concussive injury and closer to the reported average of PCS concussions which may be attributed to the fact that these were all impacts to a non-compliant surface (Oeur et al, 2014; Post et al, 2014b).

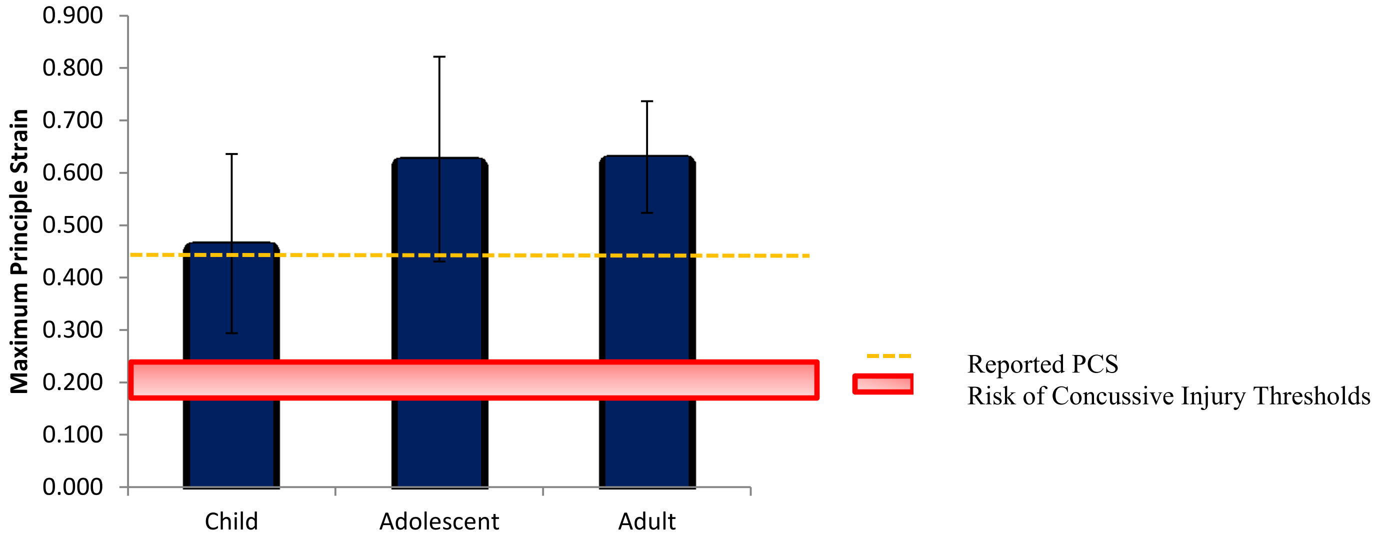

Figure 5. Average MPS values for each age group. A range of data reported from literature representing risk of concussive injury is shown by the red zone (Zhang et al, 2004; Kleiven, 2007), and reported PCS cases are represented by black dotted lines (Oeur et al, 2014; Post et al, 2014).

Maximum principle strain has demonstrated a closer association to predicting risk of concussive injury (Zhang et al, 2004; Kleiven 2007; Willinger et al, 2003). Evaluating MPS values responsible for creating concussive injury may provide a more accurate prediction of risk across varying ages. Currently, thresholds for risk of injury presented in literature are based upon adult data and represent the risk of a tissue response from trauma causing concussive symptoms. The values from all ages in this study showed responses greater than the thresholds for risk of concussive injury however, lower MPS values resulting in concussive injury were seen for the children in comparison to the adolescent and adult age groups. This may suggest that youth might be more susceptible to injury at lower brain tissue strains. For this reason, it should not be assumed that the risk of injury for adults is representative for that of youth. Understanding that differences in age may contribute to unique risks of injury could be beneficial when evaluating league regulations as well as certifying safety equipment that is appropriate for the risk of each age.

As this study examined falls to concrete, the conclusions of this study are specific to that environment. The cases of this study represent the high end of magnitudes as all cases are representative of a low compliance impacts. Previous studies have shown that varying responses can be produced by varying mechanisms such as collisions and projectiles as well as compliances therefore these results should be interpreted as such. Additionally, the finite model used was a scaled adult model as no child model currently exists. For this reason, the physiological of children may not be fully represented in this model.

5.5 Conclusion

The objective of this study was to determine if differences exist in impacts causing concussive injury between three age categories. The results of this study indicate a trend that children produce lower results across all variables. This supports the hypothesis that when examining impacts causing concussions in differing ages, significant differences will be discovered for dynamic response and brain tissue deformation. Reconstructions of injury events showed a clear distinction in the children's response as they produced the lowest results for all variables compared to the adolescents who produced the largest magnitudes of response. This dissimilarity between the youth, adolescent, and adults can contribute to the understanding in risk of concussive injury at varying ages showing that children may be at an increased risk of injury at lower magnitude events. This information should be considered in issues such as the revising of protocols for game regulation and pass/fail criteria for protective equipment to account for the risk of injury pertaining to each age category. Understanding these differences in risk of injury between ages could aid in the management of pediatric concussion to ultimately decrease the incidence of concussions in all ages.

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Appendix A: Complete Result Tables

Table 4. Results for Child Cases

Case	Location	Velocity	Linear Acceleration	Rotational Acceleration	Rotational Velocity	MPS
1	Front	2.65 (0.02)	157.8 (0.6)	5446 (171)	14.4 (0.5)	0.356 (0.019)
2	Front	2.10 (0.05)	145.7 (2.3)	8126 (644)	16.1(2.1)	0.311 (0.004)
3	Front	2.91 (0.04)	188.1 (3.9)	8808 (1006)	42.9 (0.6)	0.414 (0.026)
4	Front	2.81 (0.03)	169.1 (2.2)	6371 (291)	15.5 (0.1)	0.389 (0.005)
5	Front	1.73 (0.04)	158.0 (12.3)	5079 (1599)	14.4 (0.6)	0.318 (0.029)
6	Side	3.05 (0.01)	267.8 (5.4)	36060 (260)	17.8 (0.8)	0.704 (0.050)
7	Side	1.75 (0.03)	123.3 (1.6)	15372 (120)	16.9 (0.2)	0.395 (0.019)
8	Side	4.05 (0.09)	354.0 (56.3)	42073 (1257)	25.3 (0.6)	0.853 (0.092)
9	Rear	2.78 (0.03)	228.6 (6.9)	9252 (449)	19.0 (3.1)	0.455 (0.013)
10	Rear	3.54 (0.02)	319.2 (7.7)	13232 (400)	25.8 (4.1)	0.567 (0.024)
11	Rear	2.90 (0.01)	210.8 (11.1)	17728 (2727)	56.6 (7.8)	0.353 (0.040)

Table 5. Results for Adolescent Cases

Case	Location	Velocity	Linear Acceleration	Rotational Acceleration	Rotational Velocity	MPS
1	Front	3.12 (0.03)	269.3 (8.5)	16781 (3176)	53.1 (1.0)	0.535 (0.027)
2	Side	4.90 (0.01)	384.7 (5.6)	43891 (621)	21.1 (0.7)	0.657 (0.025)
3	Side	3.72 (0.01)	344.6 (3.4)	28402 (269)	36.5 (0.7)	0.787 (0.143)
4	Side	1.85 (0.01)	151.0 (5.7)	14543 (283)	30.0 (0.3)	0.307 (0.008)
5	Side	3.56 (0.02)	295.8 (6.0)	30473 (906)	20.7 (0.5)	0.513 (0.011)
6	Side	3.54 (0.02)	324.1 (8.4)	32263 (279)	38.4 (1.7)	0.538 (0.022)
7	Side	4.73 (0.03)	446.7 (8.2)	47911 (4115)	29.0 (0.2)	1.026 (0.140)
8	Rear	3.44 (0.03)	312.5 (19.6)	23471 (1070)	32.4 (0.1)	0.511 (0.011)
9	Rear	4.21 (0.06)	412.6 (2.4)	24349 (185)	39.9 (0.3)	0.689 (0.007)
10	Rear	3.88 (0.02)	386.1 (3.3)	26061 (287)	61.4 (8.3)	0.697 (0.015)

Table 6. Results for Adult Cases

Case	Location	Velocity	Linear Acceleration	Rotational Acceleration	Rotational Velocity	MPS
1	Front	3.11 (0.02)	256.1 (6.2)	18114 (466)	23.1 (0.2)	0.553 (0.029)
2	Side	4.06 (0.00)	185.5 (8.0)	17368 (563)	34.8 (1.0)	0.572 (0.024)
3	Side	1.23 (0.00)	291.0 (4.7)	24327 (372)	27.4 (0.6)	0.870 (0.016)
4	Rear	3.62 (0.01)	233.1 (0.3)	15318 (415)	50.1 (0.7)	0.529 (0.023)
5	Rear	3.71 (0.00)	284.5 (6.8)	19168 (422)	27.9 (0.3)	0.619 (0.023)
6	Rear	4.77 (0.02)	485.3 (7.4)	31624 (489)	22.1 (0.5)	0.788 (0.028)
7	Rear	3.14 (0.01)	242.8 (2.4)	15980 (175)	30.9 (1.8)	0.516 (0.019)
8	Rear	3.83 (0.03)	238.7 (1.4)	18580 (876)	25.8 (0.2)	0.627 (0.033)
9	Rear	3.63 (0.02)	298.2 (3.8)	19412 (211)	23.9 (0.7)	0.618 (0.026)
10	Rear	3.83 (0.01)	324.4 (0.7)	20651 (31)	27.3 (0.4)	0.624 (0.030)
11	Rear	3.82 (0.01)	311.8 (2.4)	20084 (242)	27.3 (0.1)	0.615 (0.027)

Appendix B: Full Event Description

Table 7: Child and Adolescent Case Descriptions

Age Group	Case Number	Gender	Age	Child's weight (kg)	Child's height (cm)	Fall - Drop Height-MADYMO input (cm)	Inbound Velocity from MADYMO Simulation (m/s)	Symptoms			Diagnosed with PCS	Specify type of non-sport related fall
								1 week	4 weeks	8-12 Weeks		
Children	1	M	5	21.5	97	97	2.7-3.6	nausea fatigue	No symptoms	no symptoms	No	Fall down stairs
	2	F	6	20	117	60	2.2-3.2	fatigue	dizzy sad	headache nausea balance	No	Fall from height (e.g. fall from bed or tree)
	3	F	6	29.6	117	117	2.9-4.2	headache balance dizzy fatigue	No symptoms	no symptoms	No	Struck head against household object (e.g. furniture)
	4	M	7	21	91	91	2.8-3.6	No symptoms	No symptoms	No symptoms	No	Slipped/fell/tripped on floor/ground
	5	M	5	16	107	30	1.7-2.1	nausea fatigue sensitive to light	No symptoms	headache nausea fatigue	No	Slipped/fell/tripped on floor/ground
	6	F	5	27	117	119	3.1-4.3	Sensativity to noise difficulty remembering	difficulty concentrating difficulty remembering	no symptoms	No	Struck head against wall/door
	7	M	7	26	122	25	1.8-2.1	No symptoms	no symptoms	no symptoms	No	Fall from height (e.g. fall from bed or tree)
	8	F	9	35	135	135	4.1-5.7	headache fatigue difficulty remembering vision	headache fatigue	headache fatigue	No	Struck by object
	9	M	5	17.3	107	107	2.8-5.4	headache fatigue irritable	no symptoms	no symptoms	No	Slipped/fell/tripped on floor/ground
	10	M	7	25.6	119	120	3.6-535	balance fatigue sad	difficulty remembering	difficulty remembering	No	Slipped/fell/tripped on floor/ground
	11	F	8	25.2	122	121	2.9-4.8	headache nausea dizzy	headache nausea	headache sensitive to noise difficulty remembering	No	Struck head against household object (e.g. furniture)
Adolescent	1	M	9	25	135	135	3.2-4.5	no symptoms	no symptoms	Headache nausea balance fatigue difficulty concentrating	No	Slipped/fell/tripped on floor/ground
	2	M	9	29	122	150	4.9-5.1	no symptoms	no symptoms	no symptoms	No	Recreational Play (gym, recess)
	3	M	9	34	145	143.92	3.7-6.2	nausea irritable	nausea balance fatigue sensative to light sad vision slow movements drowsy	headache dizzy irritable slow thinking	Yes	Struck head against household object (e.g. furniture)
	4	M	10	30.5	130	92	1.9-2.5	no symptoms	no symptoms	no symptoms	No	Fall from height (e.g. fall from bed or tree)
	5	F	11	69.1	165	150	3.5-5.8	headache balance dizzy fatigue sensitive to noise nervous difficulty concentrating difficulty remembering vision slow movements slow think drowsy mental fog		Headache balance dizzy fatigue sensitive to noise nervous difficulty concentrating difficulty remembering vision slow movements slow think mental fog	No	Slipped/fell/tripped on floor/ground
	6	M	12	45	168	165	3.5-5.8	No symptoms	no symptoms	no symptoms	No	Slipped/fell/tripped on floor/ground
	7	F	13	62.5	163	163	4.8-5.4	headache nausea balance dizzy fatigue sensative to light sensative to noise irritable sad difficulty concentrating difficulty remembering vision emotional slow answers slow drowsy mental fog	sensitivity to light difficulty concentrating difficulty remembering	headache dizzy irritable difficulty concentrating difficulty remembering	Yes	Slipped/fell/tripped on floor/ground
	8	M	9	50	137	122	3.4-5.4	fatigue	drowsy	no symptoms	No	Fall from height (e.g. fall from bed or tree)
	9	M	11	61.7	163	163	4.2-6.8	no symptoms	no symptoms	fatigue irritable difficulty concentrating drowsy	No	Slipped/fell/tripped on floor/ground
	10	M	12	31.6	137	137	3.8-5.6	Fatigue drowsy	no symptoms	no symptoms	No	Fall from height (e.g. fall from bed or tree)

**If no symptoms were recorded, this meant that the patient reported no symptoms 1 week following the impact or that the symptoms they were having one week post injury they also reported as experiencing prior to impact. For this reason, the symptoms that were described as occurring before impact were then excluded from the study. In this study, PCS was identified as 3 or more symptoms at exactly 4 weeks post injury.*

Appendix C: Testing Location Descriptions

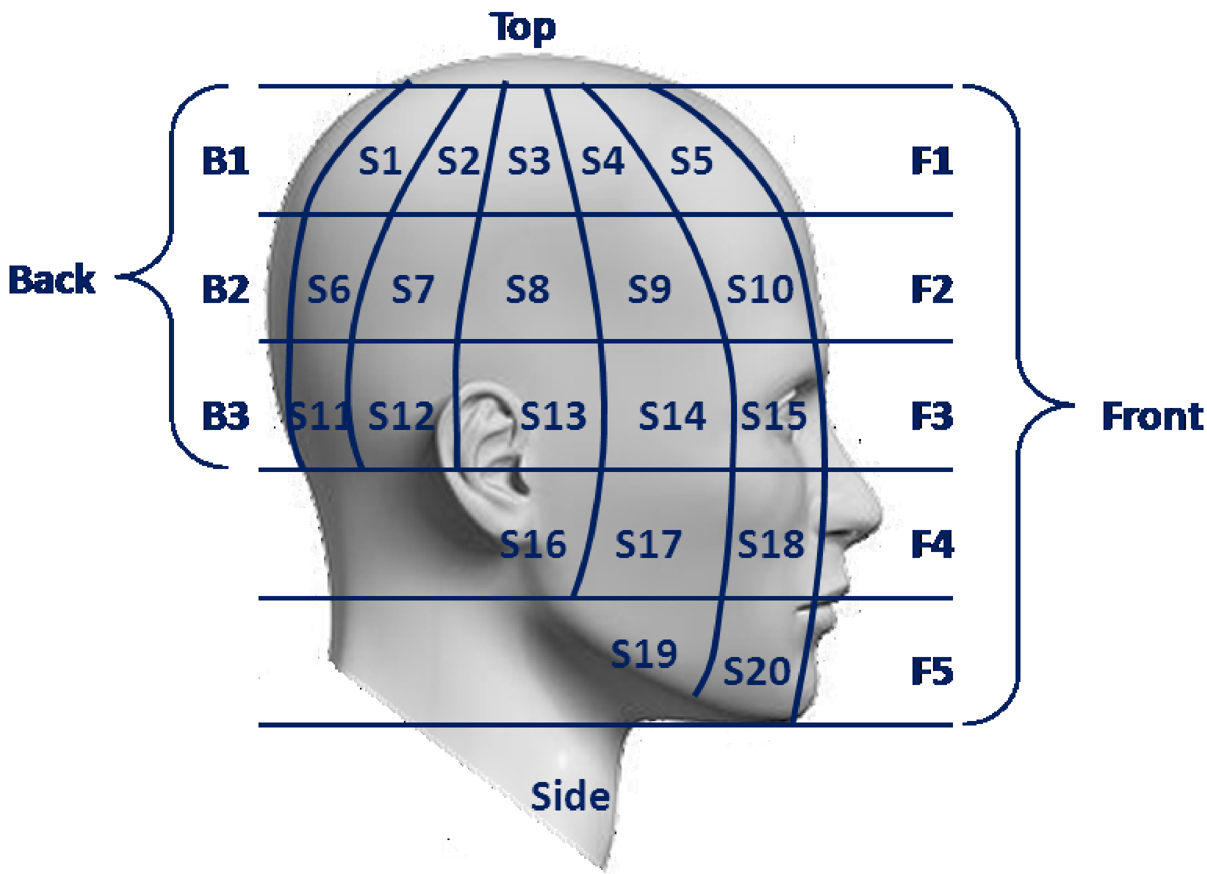

Figure 6: Visual representation of impact locations from Neurotrauma Impact Laboratory form

**If Direction of Impact was not recorded, reconstruction direct of impact was chosen based on impact side creating an option of front, side or back with no 45 degree angle. The orientation of impact was deemed straight if no other orientation was indicated.*

Child Location of Impact Description

General Location	Child Case	Patient/Parent Identified Location	Reconstruction Impact Location	Impact Side	Direction of Impact	Orientation of Impact
Front	1	F1	F1	-	-	-
	2	F1,2,3,4,5/S10,15,18,20	Midline S15/F3	Right	Front	Straight
	3	F2	F2	Left	Front	Straight
	4	F1/S10	Top Right S/10 Bottom Left F1	Midline	Front	-
	5	F1	F1	Midline	Front	Upward
Side	6	S4	S4	Left	-	Straight
	7	S1	S1	Right	Side	-
	8	S3/S4	Midline S3/S4	Right	Side	-
	9	S8	S8	Side	Side	Straight
Back	10	B2	B2	Midline	Back	-
	11	B2/B3	Midline B2/B3	-	Back	Straight

Adolescent Location of Impact Description

General Location	Adolescent Case	Patient/Parent Identified Location	Reconstruction Impact Location	Impact Side	Direction of Impact	Orientation of Impact
Front	1	F2/F3	Midline F2/F3	Left	front	Downward
Side	2	S2	S2	Left	Unknown	Unknown
	3	S6/B2	Midline S6/B2	Right	Back 45	Straight
	4	S1/S2/S3/S6/S7/S8	Midline S2/S7	Right	Side	Unknown
	5	S1/S2/S6/S7	Intersection of S1/S2/S6/S7	Left	Back 45	Straight
	6	S6	S6	Left	Back 45	Unknown
	7	S6	S6	Right	Back 45	Straight
Back	8	B2	B2	Left	Back	Downward
	9	B2	B2	Midline	Front	Straight
	10	B1/B2/B3	B2	Midline	Back	Straight

Adult Location of Impact Description

General Location	Adult Case	Reconstruction Impact Location
Front	1	Left side of face and eye
Side	2	Right Medial Parietal Region
	3	Left side of head
Rear	4	Back of head
	5	Right occiput
	6	Back of head
	7	Back of head
	8	Upper right occiput
	9	Posterior aspect of skull, occipitoparietal region
	10	Occiput
	11	Occiput

Appendix D: Sample Size Calculation

This calculation by done using the following formula for the sample size n:

$n = (Z_{\alpha/2}+Z_{\beta})^2 * 2 * \sigma^2 / d^2$,

$Z_{\alpha/2}$ is the critical value of the Normal distribution at $\alpha/2$ (e.g. for a confidence level of 95%, α is 0.05 and the critical value is 1.96), Z_{β} is the critical value of the Normal distribution at β (e.g. for a power of 80%, β is 0.2 and the critical value is 0.84), σ^2 is the variance represented by the pilot youth data and d is the difference between the means of the child pilot data and the adult data. This calculation was run for a pilot of linear, rotational and MPS. It was determined that for linear acceleration, 3 samples were needed, for rotational acceleration 17 samples were needed and for MPS 3 samples were needed. However, once all data was collected, significant main effects were found across all ages for each variable.