The Effect of Isometric Cervical Strength, Head Impact Location, and Impact Mechanism on Simulated Head Impact Measures in Female Ice Hockey Players

by

Brittany Pennock

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ABSTRACT

Head injuries in sport have become a growing concern due to the negative acute and chronic health effects manifested from concussion injuries. Ice hockey is a sport associated with a high rate of concussions, although most research has focused on concussions in men's hockey. Comparatively, women's hockey has not only seen a drastic increase in participation rates, but female hockey players also exhibit a higher concussion rate than male players, despite the "no body contact" rule that is founding characteristic of women's hockey. In fact, female hockey players may be more at risk for concussions than their male counterparts. The concerning prevalence of concussions in women's hockey has been identified, yet the factors contributing to the high risk of concussions are still unclear.

Among others, factors such as cervical muscle strength, head impact location, and impact mechanism have all been discussed in the literature as potential variables influencing the risk of concussion in athletes. The influence of these factors on head impact biomechanics, however, have not been thoroughly investigated. Furthermore, women experience high rates of concussion that have been potentially linked to decreased cervical muscle strength; however, there is little research that has characterized cervical muscle strength among female hockey players and limited research that has developed a set of normative data for female hockey players. Consequently, the purpose of this study was twofold. The first purpose was to develop normative data on the isometric cervical muscle strength and anthropometrics of female hockey players. The second purpose was to examine the effect of neckform torque, head impact location, and impact mechanism on simulated head impact measures of peak linear acceleration, shear force, and injury risk in female hockey players.

To address the first purpose, the isometric cervical strength of a sample of female hockey players (n= 25) was measured in flexion, extension, and side flexion. An average of the muscle

strength in these three directions was then calculated to develop an average overall isometric cervical strength measure for each athlete. Overall cervical strength measures of 58.64 N, 76.01 N, and 108.27 N (SD=17.52 N) represented the 10^{th}, 50^{th}, and 90^{th} percentiles, respectively, of the normally distributed dataset created from the sample. These measures were then scaled and transformed into torque measures to be appropriately modelled on a mechanical neckform to address Part II of the simulation study. The 10^{th}, 50^{th}, and 90^{th} percentile isometric cervical strength measures corresponded to torque measures of 1.36 Nm (*weak*), 2.94 Nm (*average*), and 4.62 Nm (*strong*), respectively, as established through calibration and transformation of the data. To address the second purpose, three neckform torques (weak, average, and strong), three helmet impact locations (front, rear, and side), and two impact mechanisms (direct and whiplash+impact) were tested at 16 different drop speeds using a dual-rail vertical drop system. The outcome measures included peak linear acceleration, shear force, and Gadd Severity Index, as these are variables commonly used to assess concussions in athletes.

A three-way ANOVA revealed a statistically significant main effect of impact mechanism on peak linear acceleration $F(1, 270)=55.60$, $p<.05$, $\eta^2=.17$; peak shear force $F(1, 270)=63.49$, $p<.05$, $\eta^2=.19$; and Gadd Severity Index $F(1, 270)=68.18$, $p<.05$, $\eta^2=.20$. Specifically, there was greater peak linear acceleration and peak shear force during the whiplash+impact mechanism as compared to the direct impact mechanism. There was also a significant main effect of impact location on peak shear force $F(2, 270)=13.85$, $p<.05$, $\eta^2= .09$, where frontal head impacts experienced significantly lower shear force than side and rear impacts. Furthermore, there was a statistically significant two-way interaction effect between impact location and impact mechanism on measures of peak linear acceleration $F(2, 270)=10.40$, $p<.05$, $\eta^2= .07$ and peak shear force $F(2, 270)=4.90$, $p<.05$, $\eta^2=.04$.

Existing research has used video footage to recreate real-time head impacts via simulation-based impact testing. The current research, however, is one of the first studies to incorporate human cervical muscle strength measures into simulation-based concussion testing. Cervical muscle strength is speculated to influence concussion risk, especially in female athletes who typically have weaker cervical strength than their male counterparts. Therefore, modelling the isometric cervical strength of female hockey players in simulation-based head impact testing may prove to be an effective strategy to increase the generalizability of the simulated measures to this specific population. Furthermore, the normative data characterizing the anthropometric and strength measures in female hockey players obtained from Part I may be used in future research to examine the role of head and neck anthropometrics and isometric cervical muscle strength in concussion injuries specific to the sport of women's hockey. In addition to isometric cervical strength, this research also examined the role of head impact location and impact mechanism on head kinematics in female hockey players, an area that has yet to be explored using simulation testing. Results from the head impact simulation testing in Part II will provide a foundation upon which future research can build when examining risk factors influencing concussion risk in women's hockey and may inform future epidemiological studies on which risk factors to examine in a real-world application.

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Table of Contents

ACKNOWLEDGEMENTS ... 4
List of Figures ... 9
List of Tables ... 10
List of Equations ... 10
List of Abbreviations ... 11
Chapter 1- Introduction ... 12
Chapter 2- Literature Review ... 18
Head Injury Classifications ... 18
Focal injuries. ... 18
Diffuse injuries ... 19
Concussions: General Overview ... 19
Definition. ... 19
Symptomology. ... 21
Anatomy of the Head and Neck ... 21
Skeletal structures. ... 22
Muscular structures. ... 24
Functional properties. ... 25
Biomechanics of Concussions ... 26
Shear forces. ... 28
Acceleration thresholds. ... 32
Subconcussive impacts. ... 33
Injury severity. ... 34
Concussion Risk Factors ... 36
Defining strength, torque, and stiffness. ... 36
Cervical muscle strength. ... 40
Head impact location. ... 43
Impact mechanism. ... 44
Concussions in Women's Hockey ... 46
Risk factors related to females. ... 48
Protective Equipment ... 52

Testing Protocols Using Impactors 55
Headform and neckform. 55
Impactor. 57
Characteristics of the impact. 57
Outcome measures. 58
Real-life data in simulations. 59
Research Problem 60
Chapter 3- Part I: Human Neck Strength Testing 63
Purpose 63
Research Question 63
Method 63
Participants 63
Inclusion and exclusion criteria. 63
Instruments 64
Nautilus neck strength machine. 64
Procedures 65
Participant recruitment 65
Preliminary measures. 65
Neck strength testing protocol. 68
Data Analysis 69
Results 69
Discussion 72
Implications 75
Neck strength transformation 77
Chapter 4- Part II: Head Impact Testing 79
Purpose 79
Research Question 79
Method 79
Instruments 79
Headform. 79
Accelerometers and software 81
Mechanical neckform 82

Helmet. 83
Drop system. 84
Force platform. 86
Procedures 87
Anthropometric accuracy. 87
Drop simulations. 87
Data Analysis 91
Results 92
Research Question 1: Anthropometric Accuracy 92
Research Question 2: Peak Linear Acceleration 92
Research Question 3: Peak Shear Force 97
Research Question 4: Gadd Severity Index (GSI) 101
Discussion 102
Anthropometric Accuracy 102
Peak Linear Acceleration 103
Acceleration threshold 103
Main effect of impact mechanism 104
Two-way interaction and simple main effects 105
Peak Shear Force 107
Main effects of impact location and impact mechanism. 108
Two-way interaction and simple main effects. 110
Gadd Severity Index 111
Practical Implications of Part II Results 112
Chapter 5- Discussion and Conclusion 117
The Integration of Human Data and Simulation 117
Conclusions 119
Limitations 121
Future Directions 123
References 125
Appendix A: Informed Consent Form 143
Appendix B: ParQ Form 145
Appendix C: Pre-Screening Questionnaire 147

Appendix D: Warm-Up and Cool-Down 150
Appendix E: Body Segment Parameters for Estimated Head Mass 155
Appendix F: Neckform Calibration and Data Transformation Procedures 157
Appendix G: Descriptive Summary Tables for Impact Simulations 162

List of Figures

Figure 1. The Atlas (C1) and Axis (C2) of the vertebral column… 24
Figure 2. Posterior view of the muscles of the neck that move the head. 25
Figure 3. The Wayne State Tolerance Curve. 35
Figure 4. Impact locations as defined by NOCSAE standards. 50
Figure 5. The Nautilus neck strength machine. 65
Figure 6. The CROM device. 67
Figure 7. Histogram illustrating the normal distribution of overall cervical strength measures in a sample of female hockey players. 70
Figure 8. The NOCSAE headform. 81
Figure 9. Mechanical neckform attached to the circular endplate. 83
Figure 10. The dual-rail vertical drop system. 85
Figure 11. The AMTI force platform. 86
Figure 12. Adjusting the torque of the neckform to manipulate the neck stiffness. 89
Figure 13. Equipment setup for the whiplash+impact mechanism. 91
Figure 14. Main effect of impact mechanism on peak linear acceleration. 93
Figure 15. Two-way interaction of impact mechanism and impact location on peak linear acceleration. 94
Figure 16. Peak linear acceleration across impact locations for direct head impacts.. 95
Figure 17. Peak linear acceleration across impact locations for the whiplash+impact mechanism 96
Figure 18. Peak linear acceleration between impact mechanisms for each impact location. 97
Figure 19. Measures of peak shear force for two different impact mechanisms. 98
Figure 20. Measures of peak shear force across impact locations. 99
Figure 21. Two-way interaction between impact location and impact mechanism on measures of shear force. 99
Figure 22. Measures of peak shear force for direct head impacts.. 100
Figure 23. Measures of peak shear force during each impact location. 101
Figure 24. Measures of GSI for two different impact mechanisms. 102
Figure F1. Neckform calibration setup………………………………………………………159
Figure F2. The relationship between torque and force for the mechanical neckform…………161

List of Tables

Table 1. Normal Values for Cervical Range of Motion.. 64
Table 2. Shapiro-Wilk Test of Normality for Overall Isometric Cervical Strength 70
Table 3. Isometric Cervical Strength Values of Female Ice Hockey Players 71
Table 4. Shapiro-Wilk Test of Normality for Head and Neck Anthropometric Measures........... 71
Table 5. Distribution of Head and Neck Anthropometric Measures 72
Table 6. Summary of the Torque Measures.. 78
Table 7. Approximate Measures of the Medium-Sized NOCSAE Headform.............................. 80
Table 8. Hybrid 50^{th} Percentile Neck Parameters.. 82
Table 9. Simulee Inbound Velocities and Drop Heights 88
Table E1. Body Segment Parameter Data……………………………………………………157
Table F1. Average Force Measures Generated for Varying Torque Measures………………...160
Table F2. Conversion and Scaling Process……………………………………………………..162
Table G1. Summary of Descriptive Statistics for Head Impact Simulations…………………..164

List of Equations

Equation 1-8. The Relationship Between Shear Force and Angular Acceleration………………30
Equation 9. Gadd Severity Index……………………………………………………………...…37
Equation 10. Resultant Acceleration…………………………………………………………..…83
Equation 11. Relative Error……………………………………………………………..….…88
Equation 12. Shear Force………………………………………………………………….…..…89
Equation 13. Torque Calculation……………………………………………………………..….90
Equation F1. Calibration Equation ……………………………………………………….….161
Equation F2. Z-Score Calculation……………………………………………………………...162
Equation F3. Data Scaling……………………………………………………………………...162

List of Abbreviations

GSI	Gadd Severity Index
SI	Severity index
WSTC	Wayne State Tolerance Curve
NOCSAE	National Operating Committee for the Standards on Athletic Equipment
HIC	Head Injury Criterion
mTBI	Mild traumatic brain injury
EPP	Expanded polypropylene
VN	Vinyl nitrate
PCS	Post-concussion syndrome
CSF	Cerebral spinal fluid
CROM	Cervical range of motion
Whiplash+Impact	Whiplash + head impact (when describing the second impact mechanism)

Chapter 1- Introduction

Ice hockey is a popular high-speed, high-intensity sport played by both males and females (Wilcox et al., 2014). Participation in women's ice hockey, hereafter referred to as women's hockey, has recently exploded, having seen a 900% increase over the past 15 years (Decloe, Meeuwisse, Hagel, & Emery, 2014). This increase in participation has been paralleled with an increase in the calibre and competitiveness of the sport. Consequently, despite rules prohibiting intentional body contact in women's hockey (Hockey Canada, 2003), injury rates of female hockey players have also increased (Decloe et al., 2014, Schick & Meeuwisse, 2003). In fact, approximately 50% of injuries sustained during games is a result of player contact (Agel et al., 2007). The most common injuries sustained by female hockey players are those involving the head and neck (Detis et al., 2010; Schick & Meeuwisse, 2003), with concussions reported as the number one injury by some sources (Agel et al., 2007; Schick & Meeuwisse, 2003). This is an alarming finding because of the profound effect that concussions can have on an athlete's physical, mental, and cognitive health (McCrory et al., 2017).

A concussion can be described as a mild traumatic brain injury resulting from forces applied directly to the head or transmitted to the head from an indirect force (McCrory et al., 2017). Concussions can cause acute and chronic physical, cognitive, and emotional symptoms that may negatively affect an individual's daily life (King, Brughelli, Hume, & Gissane, 2014). The biomechanical foundation of concussive injuries is still being explored; however, linear and angular accelerations are generally accepted as the forces responsible for producing the microstructural brain damage that leads to a concussion (Greenwald, Gwin, Chu, & Crisco, 2008). Controversy exists as to whether linear or angular acceleration contributes most to concussions. It is suggested, however, that using a combination of both accelerations is more representative of the impact forces when examining concussions than using either acceleration individually (Greenwald et al., 2008).

Female hockey players have been documented to have a greater risk of sustaining a concussion than their male counterparts (Brainard et al., 2012; Wilcox et al., 2015). These findings are unexpected since females experience less exposure than males, defined by total ice time participation (games and practices) (Schick & Meeuwisse, 2003). Various sex-specific reasons have been proposed to help explain the increased concussion risk in female athletes, although the exact reasoning behind this observed trend remains unclear (Brainard et al., 2012). For example, Brainard et al. (2012) identified that physiological factors such as neck strength, range of motion of the neck, body center of mass, and brain morphology may help to explain the concussion risk in female athletes. Furthermore, research has suggested that head impact location may influence the brain tissues response based on the resultant accelerations applied to the head during an impact (Walsh, Rousseau, & Hoshizaki, 2011; Zhang, Yang, & King, 2001). Additionally, since the biomechanical response of the head varies for different types of impacts, the mechanism of injury may influence the forces applied to the head and consequently the likelihood of those forces resulting in a concussion injury (Meaney & Smith, 2011). Exploring some of these potential risk factors further may provide a better understanding on how certain factors influence the relative risk of concussion injuries in a female-specific population.

Specific to women's hockey, a few of the proposed factors that warranted further investigation included the cervical muscle strength of female players, the location of head impact, and the mechanism of the injury. Insufficient cervical muscle strength in females may limit the ability of the neck to control the head during impacts, resulting in higher accelerations experienced by the head and an increased risk of sustaining a concussion (Eckner, Oh, Joshi, Richardson, & Ashton-Miller, 2014; Mihalik et al., 2011). In relation to impact location, previous research has found higher measures of peak linear acceleration during side head impacts compared to other head impact locations (Walsh, Rousseau, & Hoshizaki, 2011). Interestingly, side impacts are more frequent in female hockey players compared to male hockey players (Brainard et al., 2011), prompting further speculation into the influence of impact location on concussion risk. Finally, the impact mechanisms leading to concussions can vary. In general, concussions result from a direct head impact or inertial loading that indirectly transfers accelerations to the head (King, Yang, Zhang, Hardy, & Viano, 2003; Meaney & Smith, 2011). Direct head impacts, which typically produce a combination of linear and angular acceleration, have been more closely linked to focal head injuries than diffuse brain injuries due to the higher levels of peak linear acceleration produced (Meaney & Smith, 2011). Meanwhile, inertial loading typically generates high measures of angular acceleration, causing greater shear stress on the intercranial tissues, which has been linked to concussion injuries (King et al., 2003; McLean & Anderson, 1997). According to researchers, however, concussions can result from both direct head impacts and inertial loading since linear and angular accelerations are present during both impact mechanisms (Rowson & Duma, 2013). Subsequently, concussion testing should be interested in each type of mechanism when examining head impacts.

Concussion testing has previously involved one of two general approaches. One testing approach involves collecting head impact data using head impact telemetry (HIT), which allows researchers to measure the forces sustained from head impacts during real-time activities (Wilcox et al., 2014). A HIT system involves instrumenting helmets with accelerometers that measure the linear and angular accelerations experienced by the head during impacts. The other method commonly used in concussion testing involves simulating head impacts using surrogate devices and anvil impactors in a laboratory setting (Beckwith, Greenwald, & Chu, 2012). A horizontal or vertical anvil impactor may be used during simulation testing and the accelerations experienced by the surrogate headform can be collected via accelerometers mounted strategically in the headform. Each of these methods of concussion testing provides valuable information; although there are also limitations to each testing approach. On-field assessments, such as the HIT systems, are often limited by lack of resources and financial sustainability, while simulation testing may be limited by the inability to accurately reproduce real-life parameters. To help reduce the inherent limitations in certain testing methods, a few researchers have combined data from on-field assessments via real-time videos to reconstruct simulated head impacts and create a more comprehensive testing protocol (Beckwith et al., 2012; Pellman, Viano, Casson, & Waeckerle, 2003).

Although previous research has combined real-life data with head impact simulations, most of this research has only replicated impact velocity and location from videotapes (Pellman et al., 2003). Research has yet to use specific anthropometrics or strength measures from specific target populations to adjust the surrogate devices to achieve more realistic head impact biomechanics. For example, since females have significantly weaker neck musculature than males (Salo, Ylinen, Mälkiä, Kautiainen, & Häkkinen, 2006), it would be unrealistic to use a surrogate neckform with the same stiffness when targeting specifically males or females. Tailoring impact simulations to a specific target population would be facilitated by using normative data from the population to adjust the surrogate devices. Normative data, however, may be limited for specific populations. Therefore, by increasing the availability of the normative data for specific populations, research can increase the generalizability of results when using impact simulations for concussion research.

A review of the literature reveals a lack of normative data on factors such as head and neck anthropometrics, and cervical muscle strength in female hockey players. This information would be invaluable when examining concussion risk in female hockey players, as cervical strength has been identified as a factor potentially influencing concussion risk in the population (Brainard et al., 2012; Collins et al., 2014). Moreover, there is limited research on concussions in female athletes, despite the increasing prevalence of head injuries in women's sports. The limited research on risk factors potentially predisposing female hockey players to concussions makes it difficult to develop strategies to help decrease this risk. Examining the influence of cervical strength, impact location, and impact mechanism on head impact biomechanics may provide a foundation to better understand concussions in women's hockey. To address these identified gaps in concussion research, the purpose of this study was twofold. The first purpose was to develop normative data of the isometric cervical strength of female hockey players. The second purpose was to examine the effect of isometric cervical strength, helmet impact location, and impact mechanism on simulated head impact measures of peak linear acceleration, shear force, and injury risk in female hockey players.

This study provides a novel contribution of data related to concussions in a vulnerable population. There is currently very limited normative data reporting the anthropometric and cervical strength measures of female hockey players. By generating a robust set of normative data with this information, future research can examine the relationship between cervical muscle strength, head and neck anthropometrics, and the incidence of concussions in women's hockey. Furthermore, this research is the first known study to model human cervical muscle strength measures of a specific population on a surrogate neckform during simulation impact testing. Since simulation-based impact testing provides an acceptable means of measuring concussion risk via biomechanical variables such as peak linear acceleration, shear force, and SI, incorporating human strength measures to the simulated testing increases the generalizability of the results to the target population. Moreover, this research has identified key biomechanical impact characteristics that have been previously linked to increased concussion risk (Brainard et al., 2012; Carlson, 2016; Walsh et al., 2009; Wilcox et al., 2015) and contextualized these risk factors with anthropometric measures specific to female hockey players. Consequently, this study may provide a foundation upon which future research can build when applying human measures to simulated head impact testing to examine the influence of specific risk factors on head impact biomechanics in a female-specific population. Additionally, the results of this research may provide further insights into the mechanism between cervical msucle strength and factors associated with increased concussion risk in female hockey players.

Chapter 2- Literature Review

Head Injury Classifications

Despite the use of protective equipment in contact sports, head injuries can still occur, especially in a sport like ice hockey. Ice hockey is a fast-paced contact sport with inherent opportunity for injury risk due to the high acceleration-deceleration in skating, rapidly changing directions, high-speed shooting, and a low-friction ice surface (Biasca, Wirth, Maxwell, & Simmen, 2005). Consequently, head injuries are still one of the most common injuries in the sport for both male and female (Emery & Meeuwisse, 2006; Simmons, Swedler, & Kerr, 2017). Head injuries are typically classified as either focal or diffuse brain injuries based on the mechanical forces causing the injury and the resulting tissue damage (Andriessen, Jacobs, & Vos, 2010).

Focal injuries. Focal brain injuries are produced by direct loading, which is generated from a direct impact to the head (Andriessen et al., 2010; Biasca et al., 2005; Meaney & Smith, 2011). These direct forces cause compression of the tissues directly underneath the location of impact (coup) or the tissues that are opposite to the location of impact (Andriessen et al., 2010). Brain tissue damage is localized to the area of impact in focal injuries. The direct forces producing focal injuries may result in lacerations to the scalp, but depending on the magnitude of force and the location of the impact, are more likely to cause traumatic brain injuries such as skull fractures, cerebral contusions, and epidural hematomas (Andriessen et al., 2010; Biasca et al., 2005; Kleiven, 2013). The use of helmets in sport, however, has effectively decreased the prevalence of traumatic focal injuries in sports such as hockey (Rousseau, Post, & Hoshizaki, 2009). Although often classified as mild traumatic brain injuries (mTBIs), diffuse brain injuries in sport are equally as concerning as traumatic focal injuries, particularly because head protection has proven to be ineffective in preventing diffuse brain injuries and the associated effects.

Diffuse injuries. Diffuse brain injuries typically result from rapid acceleration-deceleration of the head, causing widespread white matter damage (Andriessen et al., 2010). During rapid acceleration-deceleration, also known as inertial loading, the movement of the head causes the brain to move within the cranium. Although the cerebral spinal fluid (CSF) is designed to act as a shock-absorber for the brain, the varying tissue properties and fixation of the brain within the cranium cause certain brain segments to move at different rates than others during inertial loading (Biasca et al., 2005; Andriessen et al., 2010). This non-uniform movement of the brain applies shearing and tensile forces to the tissues, which are not effectively managed by the properties of the CSF (Biasca et al., 2005). Since brain tissue is one of the softest biological materials (Meaney & Smith, 2011), it deforms easily in response to shearing, creating diffuse trauma throughout the intercranial tissues. Although not yet fully understood, concussions are a well-recognized subset of diffuse brain injuries and commonly occur in sports. Gaining a better understanding of concussions as they relate to sports is the first step in reducing the risk of neurological impairment in athletes.

Concussions: General Overview

Definition. The universal definition of a concussion is inconsistent in the literature due to the complex pathophysiology and unique manifestation of concussion symptoms for every individual. In the early 1990s, a concussion was defined as a traumatically induced physiological disruption of brain function with a short period of altered or loss of consciousness (The American Congress of Rehabilitation Medicine, 1993). It is now known, however, that altered or loss of consciousness is not required to be diagnosed with a concussion (King et al., 2014). Most researchers use operational definitions of a concussion to ensure that the term is being addressed appropriately in the research. For example, Covassin, Moran, and Elbin (2016) operationalized concussion as “a complex pathophysiological process affecting the brain, induced by traumatic biomechanical forces” (p. 190). Meanwhile, Browne and Lam (2006) broadly defined a concussion as a “traumatically induced alteration in mental status that may or may not be associated with loss of consciousness” (p. 163).

To help clarify the definition of a concussion, an expert panel at the 2016 International Conference on Concussion in Sport in Berlin developed a general definition of a sport-related concussion and some common defining features. According the expert panel, a sport-related concussion can be defined as “a traumatic brain injury induced by biomechanical forces” (p. 2), typically resulting from a direct blow to the head or an inertial force transmitted to the head (McCrory et al., 2017). A sport-related concussion results in the rapid onset of transient neurological impairment that leads to functional disturbance, rather than a structural injury, with signs and symptoms typically resolving spontaneously after seven to ten days. The spontaneous resolution of symptoms has lead experts to believe that concussions cause neuronal dysfunction, rather than cell death (Giza & Hovda, 2001). Although the neurometabolic cascade that occurs during a concussion is complex, researchers identify that the neuronal dysfunction experienced may occur due to ionic shifts, altered metabolism, impaired neuronal connectivity, or changes in neurotransmission (Giza & Hovda, 2001). These alterations in cell function, however, will typically return to normal with time and proper post-concussive management. The clinical signs and symptoms, however, can be prolonged in some cases, resulting in post-concussive syndrome (PCS).

Symptomology. Concussions manifest diverse clinical signs and symptoms, which can affect an individual's physical health, mental health, behavioural patterns and mood, cognition, and sleep patterns (Daneshvar, Nowinski, McKee, & Cantu, 2011). King, Brughelli, Hume, and Gissane (2014), described many of the signs and symptoms of a sport-related concussion in a systematic review of the literature. Accordingly, common physical symptoms reported by concussed individuals include headache, dizziness, nausea and vomiting, photosensitivity, and drowsiness. Affective symptoms can include irritability, sadness, and anxiety. Cognitively, concussed individuals may experience difficulty concentrating, feelings of being in a "fog", and difficulty remembering (King et al., 2014). Moreover, sleep patterns may also be affected by a concussion, as the individual may experience an inability to sleep, or may sleep more than usual (King et al., 2014). Due to the multifaceted nature of concussions, every individual will experience different symptoms. Complicating concussion diagnosis further, self-reported symptomology requires subjects to willingly, truthfully, and accurately report symptoms (Alla, Sullivan, Hale, & McCrory, 2009; McCrea, Hammeke, Olsen, & Leo, 2004). Inability to proficiently provide details of concussion symptoms can lead to an under-diagnosis or inappropriate rehabilitation for concussions. Symptoms experienced may reflect the region of the brain or cervical structures that were injured in concussive blow, which is why understanding the basic knowledge of the head and neck anatomy is useful when examining concussions.

Anatomy of the Head and Neck

Movements of the head are controlled by the muscles of the neck, but the cervical kinematics are also largely influenced by the shape and structure of the cervical vertebrae. (Bogduk & Mercer, 2000). Since a concussion can occur from inertial loading, in addition to direct head impacts, understanding each of these mechanisms requires foundational knowledge of the musculoskeletal structures of both the head and neck. Due to the complexity of the anatomy of the head and neck, only a general overview of the musculoskeletal structures that are pertinent to concussion injuries will be explored in this literature review.

Skeletal structures. The skull is the boney structure of the head that is made up of the cranium, which houses the brain, and 12 facial bones. The cranium is formed by eight separate bones, including the frontal bone, two parietal bones, two temporal bones, the occipital bone, the sphenoid bone, and the ethmoid bone (Tortora & Neilson, 2012). Since impacts to the head can occur from any direction, all regions of the head are susceptible to impacts causing concussions. The cranial bones are fixed structures, held together by sutures, to facilitate the protection of the enclosed brain. The inner surface of the cranium attaches to layers of meninges, which are inner membranes that stabilize the position of the brain, blood vessels, and nerves (Tortora & Neilson, 2012). Although the meninges act to help protect the brain through stabilization of the structures within the cranium, the brain is essentially a floating structure within a sea of CSF. Consequently, excess or rapid movement of the head, such as with a player-on-player collision in ice hockey, can cause the brain to shift in the CSF within the cranium, impacting the inner walls of the cranium (Andriessen et al., 2010). If the brain impacts the inner surface of the cranium with enough force, damage to the intercranial tissues may occur. Damage to the intercranial tissues resulting from the brain impacting the side of the cranium causes a cascade of cellular events that disrupts normal brain cell function, which we identify as a concussion (Graham, Rivara, Ford, & Spicer, 2014). The movement of the head is controlled by the structures of the neck, which have the potential to either increase or decrease the magnitude of forces applied to the head and the enclosed brain.

Structurally, the cervical spine supports the skull, acts as a shock absorber for the brain, and serves a protective role by protecting the brainstem, spinal cord, and various neurovascular structures (Nordin & Frankel, 2012). Biomechanically, the neck facilitates the transfer of weight and bending moments of the head (Nordin & Frankel, 2012). The cervical spine is composed of the seven cervical vertebrae which are the smallest and most variable of all the vertebrae, and have the greatest degree of range of motion at their joint surfaces (Swartz, Floyd, & Cendoma, 2005; Tortora & Nielsen, 2012). The first two cervical vertebrae are very different than the other five cervical vertebrae (Figure 1). The first cervical vertebra (C1) is called the atlas. This vertebra articulates superiorly with the occipital condyles, forming the atlantooccipital joint. The atlas supports the head and allows for flexion and extension of the neck and head. The second cervical vertebra (C2) is called the axis (Swartz, et al., 2005; Tortora & Nielsen, 2012). The axis rests within the facets of the atlas and features a boney process, the odontoid process, which projects through the vertebral foramen of the atlas to allow right and left rotation of the neck and head. The high degree of mobility of the cervical spine is a structural characteristic that causes the cervical structures to be stressed to their limits during an inertial load, which can cause concussions (Morin, Langevin, & Fait, 2016). Consequently, the cervical musculature surrounding these mobile joints plays an important role in decreasing the high degree of movement of the neck in response to perturbation.

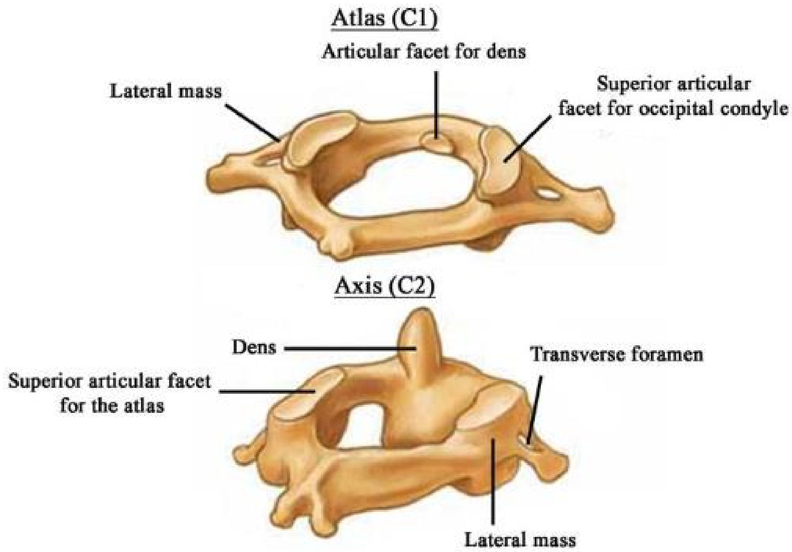

Figure 1. The Atlas (C1) and Axis (C2) of the vertebral column. The top and bottom images illustrate the main features of the atlas and axis from an anterior view, respectively. Adapted and modified from "The neck: Neck injuries in military scenarios" by K. Bridwell, 2016, pp. 221-256. In "Military injury biomechanics", by M. Franklyn and P. Vee Sin Lee (Chapter 11). Philadelphia, PA: Elsevier Inc. Copyright © Dr. Keith Bridwell, 2016.

Muscular structures. Overlying the vertebrae are several muscles, some of which are more prominent than others due to their relative size and anatomical depth (Figure 2). A large, superficial muscle of the neck is the sternocleidomastoid muscle. This muscle is responsible for flexing the neck when activated bilaterally, and laterally rotating and flexing the head to the opposite side of the contracting muscle when activated unilaterally (Tortora & Nielsen, 2012). The posterior fibres of the sternocleidomastoid can also assist in extending the head. The anterior, middle, and posterior scalene muscles are the muscles of forced inhalation, as they originate on the cervical spine and act to elevate the ribs (Tortora & Nielson, 2012). Another large, superficial muscle is the trapezius muscle, which acts to elevate the scapula and extend the neck (Tortora & Nielson, 2012).

Furthermore, the semispinalis capitis, splenius capitus, and longissimus capitis muscles are smaller than the previous muscles and collectively contract to help extend the head (Tortora & Nielsen, 2012). These muscles also act individually to rotate the head to the opposite side of the contracting muscle (Tortora & Nielsen, 2012). Finally, the spinalis capitis is a very small muscle that is often absent but helps to extend the head when it is anatomically present (Tortora & Nielsen, 2012). The musculature of the neck is believed to be important in helping to reduce the risk of concussions and the severity of subconcussive blows because 80% of the mechanical load placed on the head and neck is managed by the neck muscles, while bones and ligamentous structures manage the remaining 20% of the load (Schmidt et al., 2014). It is important to recognize, however, that many of these cervical muscles work together to control the movement of the head during an impact, regardless of the direction of force application.

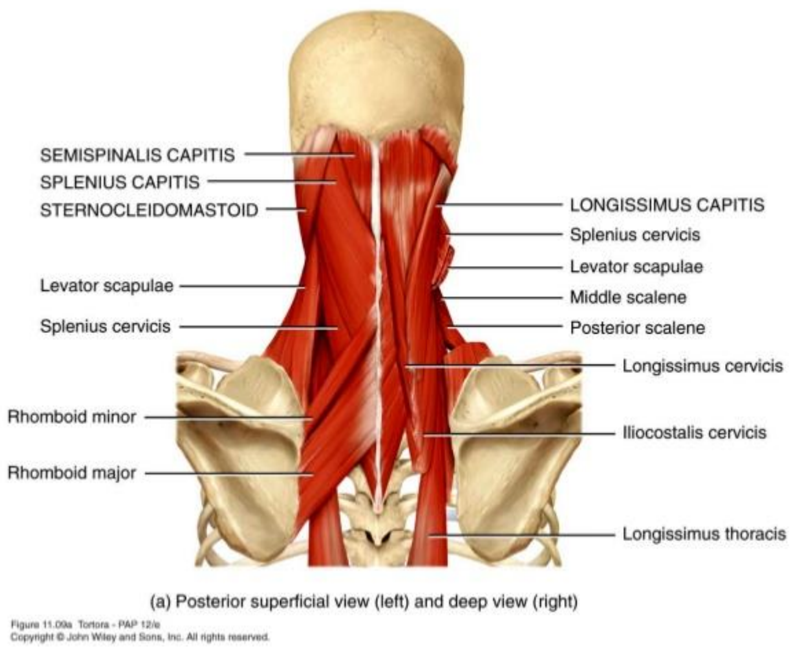

Figure 2. Posterior view of the muscles of the neck that move the head. The left side of the diagram (a) illustrates the superficial cervical muscles, while the right side (b) identifies the deep view of the cervical musculature. Adapted and modified from “Principles of human anatomy” by G. J. Tortora and M. T. Nielson, 2012, (12th ed.), p. 361. Hoboken, NJ: John Wiley & Sons, Inc.

Functional properties. Functionally, the neck moves through all three planes of motion (i.e. frontal, sagittal, transverse) around all three axes of rotation (i.e. mediolateral, antero-posterior, longitudinal). Neck flexion-extension occurs at the atlantooccipital joint in the sagittal plane around the mediolateral axis; rotation of the neck occurs at the articulation of the atlas and axis vertebrae in the transverse plane around the longitudinal axis; and lateral flexion-extension occurs at the superior and inferior articular facets of the last five cervical vertebra in the frontal plane around the anteroposterior axis (Tortora & Nielsen, 2012). An impact of sufficient force from any direction in any plane of motion can result in a concussion. According to Swartz, Floyd, and Cendoma (2005), the cervical spine moves through approximately 80 to 90 degrees of flexion, 70 degrees of extension, 20 to 45 degrees of lateral flexion, and up to 90 degrees of rotation to both sides. It is important to note that concussions can be experienced within the normal range of motion of the cervical spine. (Cholewicki et al., 1998; Panjabi et al., 1998). These anatomical and functional properties of the head and neck provide a foundation for understanding the mechanisms underlying concussions through direct and indirect loading. The next task is to understand the biomechanics of a concussion as it relates to the direct or inertial forces applied to the head during an impact.

Biomechanics of Concussions

According to King, Yang, Zhang, Hardy, and Viano, (2003) brain deformation, or strain, is the primary cause of concussions; however, it is difficult, if not impossible, to measure brain deformation *in vivo* upon impact in humans. Consequently, linear and angular head accelerations are most often used when exploring the behaviour of the head and neck in concussions, and are generally accepted as the primary mechanical factors causing this type of head injury (Greenwald, Gwin, Chu, & Crisco, 2008; Guskiewicz & Mihalik, 2011; Meaney & Smith, 2011; Rowson & Duma, 2013). During an impact, linear and angular head accelerations produce movement of the head, subsequently producing shear, tensile, and compressive forces within the brain tissues (Andriessen et al., 2010). If the head accelerations produce significant enough forces applied to the brain tissues, the impact may lead to a diffuse brain injury, causing a concussion. Linear acceleration of the head results from an impact force directed through the center of mass of the head and is measured in the unit ‘g’, which quantifies acceleration as a multiple of gravity (9.81 m/s^2). Meanwhile, angular acceleration is produced when a force vector does not pass through the centre of mass of the head, causing an off-centre rotation (McLean & Anderson, 1997). Angular acceleration is measured in radians per second squared (rad/sec^2).

Controversy exists as to whether linear or angular acceleration is more causative ofconcussions. Original research suggested that linear acceleration is most responsible for concussions (Gurdjian & Webster, 1945). Emerging evidence, however, has disputed this claim. Instead, it has been suggested that linear acceleration results in focal injuries of the head, rather than diffuse brain injuries which are more associated with concussions. Linear acceleration is typically greater in direct impacts, which are more likely to result in focal brain injuries, as previously discussed (Andriessen et al., 2010; Kleiven, 2013). Conversely, diffuse brain injuries, including concussions, are believed to be produced more by angular acceleration due to the shearing forces experienced by the intercranial tissues and resulting in the widespread neurological disruptions (Andriessen et al., 2010; Biasca et al., 2005; Kleiven, 2013).

In support of these emerging claims of the instrumental role of angular acceleration in concussive mechanisms, an early animal study conducted by Gennarelli, Thibault, and Ommaya (1972), revealed that pure translational head impacts in animals did not result in a concussion. Meanwhile all animals that were subjected to head rotation, rather than pure translation, sustained a concussion. More recent literature has supported these observations (Kleiven, 2013; Meaney & Smith, 2011; Rowson et al., 2012). Angular acceleration has been found to produce greater tissue shearing than linear acceleration, which has been found to contribute to greater brain deformation than pure translational forces (Kleiven, 2013; Meaney & Smith, 2011). As such, several researchers claim that angular acceleration is main mechanical factor contributing to concussions (Kleiven, 2013; Meaney & Smith, 2011; Rowson et al., 2012).

Although, linear and angular acceleration rarely, if ever, exist independently of one another during head impacts (Clark, Post, Hoshizaki, & Gilchrist, 2016). It has, therefore, been suggested that observing linear and angular accelerations independently when examining concussions is irrelevant. Instead, both accelerations should be considered in the biomechanical foundation of concussions. Greenwald, Gwin, Chu, and Crisco (2008), measured linear and angular head acceleration, impact duration, and impact location in 449 football players using instrumented helmets to investigate whether a combination of biomechanical factors was more useful in predicting a concussion than a single factor. Results of the study suggested that using a combination of factors, including both linear and angular acceleration, was more useful in predicting a concussion than using only one of the biomechanical factors. Similarly, Rowson and Duma (2013) used compared the predictive capability of a new metric using both linear and angular acceleration as opposed to a single biomechanical parameter and found that resultant effects of linear and angular head acceleration was a better predictor of sustaining a concussion than either measure individually. Consequently, my thesis work examined how linear acceleration and the shear impact forces producing angular acceleration were influenced by cervical muscle strength, head impact location, and impact mechanism, rather than only measuring a single outcome measure.

Shear forces. Shear force is “a force applied parallel to a surface, causing internal deformation in an angular direction” (Hamill, Knutzen, & Derrick, 2015, p. 40). Based on this definition, shear head impact forces cause rotation of the head, thereby producing angular acceleration of the head. Shear force and angular acceleration are mathematically inter-related, although shear force has not been examined extensively in concussion research. The following equations can be used to describe the relationship between shear force and angular acceleration:

Based on Newton's Second Law of Motion:

$$F = m \times a \tag{1}$$

where:
F = external impact forces
m = mass of particles
a = linear acceleration due to impact

$$\text{Let, } a = r \times \propto \tag{2}$$

where:
r= position vector
α = angular acceleration

We can rewrite Equation 1 as:

$$F = m \times r \times \alpha \tag{3}$$

We are interested in computing the sum of moments of inertia and torques producing angular impact accelerations.

To do this, let *I*= the sum of moments of inertia of particles and *T*= the sum of all torques.

$$I = \int r^2 \times dm \tag{4}$$

$$T = \int F \times dr \tag{5}$$

In order to substitute Equation 4 and Equation 5 into Equation 3, first multiply both sides of Equation 3 by the position vector, *r*.

$$F \times r = m \times r^2 \times \alpha \tag{6}$$

Since $F \times r$ = Torque and $m \times r^2$= moment of inertia, we can substitute Equations 4 and 5 into Equation 6 as follows:

$$\int F \times dr = \alpha \int r^2 \times dm \tag{7}$$

Therefore, we can isolate angular acceleration as:

$$\alpha = \frac{\int F \times dr}{\int r^2 \times dm} \tag{8}$$

This mathematical relationship demonstrates the association between the external applied force, which would be shear in off-centre impacts, and the angular impact acceleration generated. Based on Equation 8, angular acceleration generated from an off-centre impact may be influenced by the magnitude of the force and the position vector. The position vector would be the radius of the system. This relationship can be applied to head impacts with regards to the influence of the magnitude and location of the external shear force on the subsequent angular acceleration of the head. Concussion risk may consequently be increased due to variations in these factors that result in higher shear forces and angular accelerations (Meaney & Smith, 2011).

The angle of head impact is one factor that can affect the shear forces transmitted to the head during contact. According to Halldin and Kleiven (2013), the steepness of the impact and the friction present at the moment of impact are two factors that influence the shear forces applied to the head. Greater steepness and higher levels of friction cause greater shearing, generating potential for greater intercranial tissue damage (Halldin & Kleiven, 2013). Few studies have examined the behaviour of the head during impacts at varying angles (Oeur, 2012; Walsh, Rousseau, & Hoshizaki, 2011). Furthermore, those studies that have examined angled impacts have measured the resulting accelerations, rather than the shear forces causing the head rotation.

Walsh, Rousseau, and Hoshizaki (2011) simulated horizontal head impacts using a linear pneumatic impactor and a Hybrid III headform at five different impact locations and four impact angles. When examining the effect of impact location and angle of impact on linear and angular head acceleration, Walsh et al. (2011) demonstrated that impacting the headform at a 45-degree rotation produced greater angular acceleration than impacting a neutrally positioned headform; although, both head orientations resulted in angular accelerations associated with a high probability of sustaining a concussion based on acceleration threshold measures proposed by Zhang, Yang, and King (2004). Comparatively, linear head acceleration was found to be greater with the headform in the neutral position as compared to a rotated position (Walsh et al., 2011). Other research has supported this finding, indicating that linear acceleration is greater when the head is impacted at neutral, rather than angled positions (Carlson, 2016). This evidence subsequently supports the relationship between angled impacts and increased angular acceleration, which may be related to a higher risk of sustaining a concussion.

Most research, however, has focused on angular acceleration of the head during impacts, failing to examine the shear force applied to the head. It is potentially valuable to explore the shear forces applied to the head during impact, since shear force is the source of angular acceleration. Carlson (2016) has been one of the few researchers to examine the shear forces transmitted to a surrogate head during simulated head impacts. Carlson demonstrated that the shear forces transmitted to the headform were significantly greater during angled impacts, as opposed to impacts onto a 0-degree incline. Since shear force generates angular acceleration, it can be assumed that greater shear forces would caused a greater angular acceleration of the head, thereby increasing the risk of concussions due to the shearing of intercranial tissues. Since most existing research measures the angular acceleration of the head during impact (Brainard et al., 2012; Mihalik et al., 2010; Walsh et al., 2011; Rowson et al., 2012; Rowson & Duma, 2013;), there is potential to expand this research by focusing on shear forces rather than the resulting angular accelerations. Gaining a better understanding of shear forces during head impacts would allow researchers to improve understanding of the source of one of the biomechanical measures related to concussions.

Acceleration thresholds. Although the literature on head impact biomechanics generally agrees that linear and angular head acceleration are the biomechanical measures used to characterize concussions, the threshold at which a concussion is likely sustained from these accelerations remains unclear (Guskiewicz & Mihalik, 2011). According to Guskiewicz and Mihalik (2011), the reason for the disparity in acceleration thresholds that may induce concussions is because there are many factors that influence the ability of the body to dissipate forces applied directly or indirectly to the head, including the cerebrospinal fluid levels and function, the athlete's vulnerability to brain tissue injury, the relative musculoskeletal strengths and weaknesses, and the athlete's anticipation of the impending impact.

Gurdjian (1972) proposed that a linear acceleration of 80-90 g for greater than four milliseconds would result in a concussion. Comparatively, Zhang, Yang, and King (2004) suggested that a peak resultant linear acceleration of 66 g, 82 g, and 106 g at the centre of gravity of the head corresponds to a 25%, 50%, and 80% probability of sustaining a concussion, respectively. These thresholds were derived from typical impact durations of 10-16 milliseconds. Similarly, peak resultant angular accelerations were proposed to be 4600 rad/s^2, 5900 rad/s^2, and 7900 rad/s^2 for 10-30 millisecond impacts in association with a 25%, 50%, and 80% probability of sustaining a concussion, respectively (Zhang et al., 2004). Zhang et al. (2004) used video-recordings of football impacts resulting in confirmed concussion diagnoses and reconstructed these head impacts in the laboratory using surrogate devices. Head kinematics, including linear and angular acceleration, were measured during the head impact simulations and used in a sophisticated finite element head model to help determine injury tolerance levels for concussions (Zhang et al., 2004). Although these tolerance thresholds were established from data characterizing a male sample, the threshold values proposed by Zhang et al. (2004) were used in the current research to quantify the likelihood of the simulated head impacts producing linear acceleration measures large enough to result in a concussion. Many researchers use the acceleration thresholds proposed by Zhang et al. in their concussion research (Carlson, 2016; Clark, et al., 2016; Ouer, 2012; Walsh et al., 2011) because the thresholds were established using similar equipment and protocols used in most simulation-based concussion research. Future research, however, should use similar protocols to establish acceleration thresholds specific to females, as the threshold tolerance may differ for males and females. Furthermore, proposed thresholds for shear force magnitude linked to concussions do not currently exist in the literature, so they could not be applied in the current research.

Subconcussive impacts. The majority of research on sports-related concussions has been concerned with impacts resulting in concussions, while little research has concerned itself with those head impacts that generate head accelerations below the threshold for eliciting concussion injuries (i.e. subconcussive blows). These subconcussive impacts occur more frequently in sport than concussive impacts, yet do not manifest the clinical signs and symptoms that are observable with concussions. Although subconcussive impacts are not detectable clinically, they have still been associated with pathophysiological changes and damage to the nervous system (Bazarian, Zhu, Blyth, Borrino, & Zhong, 2011; Breedlove et al., 2012; Broglio, Eckner, Paulson, & Kutcher, 2012; Johnson, Neuberger, Gay, Hallett, & Slobounov, 2014; Talavage et al., 2014).

An examination of the neurological performance and health of high school football players using neurocognitive testing and functional magnetic resonance imaging (fMRI) has revealed significant neurocognitive (Talavage et al., 2014) and neurophysiological (Breedlove et al., 2012) changes due to subconcussive impacts. Bazarian, Zhu, Blyth, Borrino, and Zhong (2011) also identified significant changes in brain white matter in high school football and hockey athletes who experienced multiple subconcussive blows. There were abnormalities in the diffusion tensor imaging of the athletes with the multiple subconcussive blows, which more closely resembled imaging of a concussed athlete than a non-athlete healthy control (Bazarian et al., 2011). These findings, however, were based on a male population, as are many sport-related concussion research studies. Physiological differences, including possible differences in brain morphology between males and females (Brainard et al., 2012; Wilcox et al., 2015), makes it difficult to apply these findings directly to females. Nonetheless, it remains important for both genders to consider the potential negative effects of all levels of head impacts, including impacts that are considered subconcussive, as these induce clinically important changes that manifest after repeated exposure. The likelihood of sustaining a concussion from an impact can be estimated using various injury severity indices derived from the Wayne State Tolerance Curve, including Gadd Severity Index.

Injury severity. Using the biomechanical factors related to head impacts, including head accelerations and impact duration, the relative severity of resulting injury (i.e., a concussion) can be estimated. There are various head injury tolerance criteria that exist (Greenwald et al., 2008), which are used in predicting injury severity resulting from a head impact. Some of these tolerance criteria include the Wayne State Tolerance Curve (WSTC; Gurdjian, Roberts, & Thomas, 1966), Head Injury Criteria (HIC; Versace, 1971), and the Gadd Severity Index (GSI;

Gadd, 1966). The WSTC, illustrated in Figure 3, was developed to better understand the relationship between linear head acceleration and head injury in automotive crashes. This criterion uses linear acceleration and impact duration to predict the acceleration threshold that induces skull fractures; these forces are theorized to correlate with head impacts that induce moderate to severe concussions (Greenwald et al., 2008). The HIC and the GSI were derived from the WSTC. The National Operating Committee for the Standards on Athletic Equipment (NOCSAE) uses GSI as a tolerance criterion in helmet testing (NOCSAE, 2017).

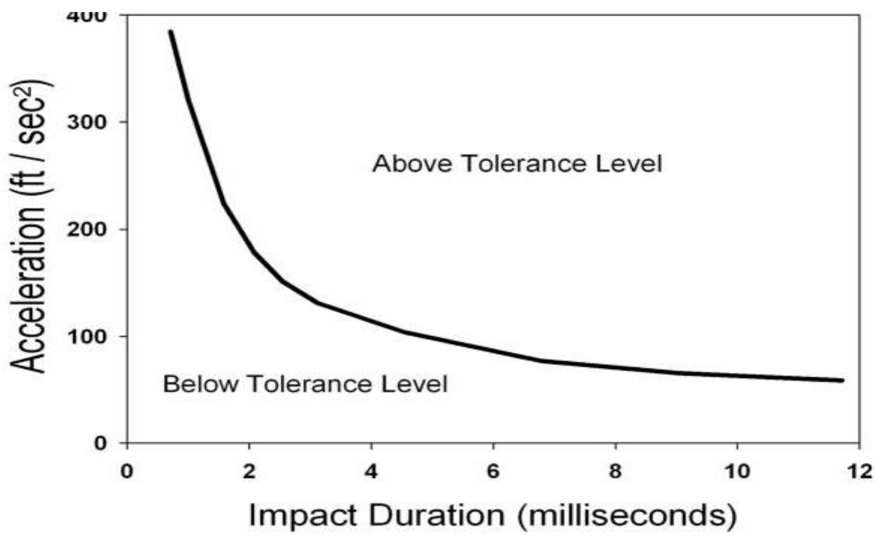

Figure 3. The Wayne State Tolerance Curve. This injury tolerance curve uses peak linear acceleration and impact duration to predict the threshold of skull fracture. Adapted from “Head impact severity measures for evaluating mild traumatic brain injury risk exposure”, R. Greenwald, J. Gwin, J. Chu, and J. Crisco, 2008, *Neurosurgery, 63*(4), p. 789-798.

These tolerance curves, however, are most effective at predicting traumatic brain injuries, such as skull fractures. A tolerance criterion specific to mild traumatic brain injuries, such as concussions, is lacking in the literature. Therefore, it is difficult to define an accurate injury severity threshold that would likely lead to a concussion. Nonetheless, the GSI is frequently used in understanding injury severity in sport because of its by NOCSAE in helmet tolerance testing. The GSI uses an algorithm (Equation 9) to measure the severity of an impact based on peak linear head accelerations, and the index cannot exceed acceptable levels (NOCSAE, 2017).

$$\text{GSI} = \int_{t0}^{t1} A^{2.5} dt \qquad (9)$$

where:

A= head acceleration impulse function
t_1= impulse duration

Gadd (1966) reported that an index of 1000 on the GSI represents the upper limit for severe brain injury, causing possibly life-threatening injuries. Meanwhile, the National Highway Traffic Safety Administration had identified 700 as the tolerable upper limit for severe head injury (Eppinger et al., 1999). Therefore, if these indices are proposed upper limits for severe brain injury, it is likely that the upper limits for concussion injury severity are more conservative than the proposed values. There are many factors that may influence the risk of injury resulting from a head impact. Concussion risk factors, such as those described in the subsequent section, have the potential of increasing or decreasing the injury severity of impacts by influencing the forces applied to the head during direct and/or indirect loading. The GSI, however, at least provides an indication of the injury severity and the relative risk of head trauma from an impact.

Concussion Risk Factors

Defining strength, torque, and stiffness. Existing literature documenting properties of the cervical musculature and the behaviour of the neck during direct and indirect loading has used the terms strength, torque, and stiffness rather loosely and without a consistent definition. In fact, many previous studies have not clearly defined these terms. Clarifying the operational definition of important terminology, such as the strength, torque, and stiffness, is crucial to understanding the theoretical concepts and practical applications of concussions. Strength can be defined as “the amount of force a particular muscle or group of muscle can produce” (Richards, 2008, p. 31) and can be influenced by such factors as the body segment inclination, muscle insertion, the angle of pull of the muscle, and the type and speed of contraction. Although, Richards (2008) explains that it is usually the effective moment of the muscle that is being measured, rather than the direct muscle force, because force is always acting at a certain distance from an axis of rotation. Moment is synonymous with torque and is said to be the rotational equivalent to linear force production (Richards, 2008). Muscle torque, therefore, is the product of the force produced by the muscle and the shortest perpendicular distance from the muscle attachment to the axis of rotation (Nordin & Frankel, 2012).

Muscle force and muscle torque are often used interchangeably in the literature. It is inappropriate, however, to do this despite their close relation to one another. Rezasoltani, Ylinen, Bakhtiary, Norozi, and Montazeri (2008) compared strength and torque measures of the cervical extensor muscles and found that, although force and torque measures were highly correlated, they could not be directly compared to one another because of the varying length of the moment arm. Determining the length of the moment arm when measuring neck torque, however, has proven to be difficult. Previous studies have used different moment arms when measuring neck torque (Harms-Ringdahl & Schuldt, 1988; Pollock et al., 1993; Portero et al., 2010; Staudt & Duhr, 1994). The axis of rotation has been identified as being halfway between the spinous process of C7 and the upper margin of the manubrium sterni (Harms-Ringdahl & Schuldt, 1988) and at the level of the thyroid cartilage (Pollock et al., 1993) in previous research. Staudt and Duhr (1994) used the entire length of the neck as the moment arm to calculate neck torque in their research. Other studies that have calculated neck torque did not indicate the length or location of the moment arm (Portero et al., 2010; Schmidt et al., 2014). Inconsistencies in the length of the moment arm when measuring muscle torque may subsequently overestimate or underestimate the torque generated by muscles, making it difficult to make accurate comparisons across different research studies.

According to Rezasoltani et al. (2008), "there is no single joint by which the lever arm can easily be measured. There are rotation axes distributed in all segments of the cervical spine…" (p. 380). Salo, Ylinen, Mälkiä, Kautiainen, and Häkkinen, (2006) also justified their use of cervical strength measures, expressed in Newtons, when measuring the isometric cervical strength of male and female participants, indicating that "the cervical spine has no clearly defined single axis or lever arm for the movement of the flexion and extension" (p. 497). A review of the literature highlighted the need to develop more universal methods for calculating neck torque. Therefore, although strength and torque are closely related, they cannot be used interchangeably, and it is important to clearly define how each measure is to be calculated to ensure the appropriate measure is being used. Concussion research has typically measured cervical muscle strength as opposed to neck torque to help limit the potential inconsistencies related to accurately defining a standard moment arm within the cervical structures (Broennle, Kivi, & Zerpa, 2017; Collins et al., 2014; Hildenbrand & Vasavada, 2013; Mihalik et al., 2011).

Another term that is frequently used when discussing the behaviour of the neck in response to an external perturbation is neck stiffness. The conventional definition of muscle stiffness is the resistance to deformation under an applied load (McLean & Anderson, 1997). Portero et al. (2015) further define musculotendinous stiffness as the "stiffness of the series elastic components of a muscle or group of muscles" (p. 2). A review of the literature suggests that cervical muscle strength and torque are related to neck stiffness. Not only are stronger muscles able to produce greater absolute force, but they can also produce greater tensile stiffness (Schmidt et al., 2014), and generate torque more quickly than weaker muscles (Eckner, Oh, Joshi, Richardson, & Asthon-Miller, 2014). Portero et al. (2013) have also demonstrated that increased muscle torque significantly increased the musculotendinous stiffness. According to

Schmidt et al. (2014), “stiffness of the cervical region is proportional to both muscle activity and force generated through muscular contraction” (p. 2064). Neck stiffness is also increased through preparatory muscle activity (Schmidt et al., 2014).

The descriptions above illustrate the way in which the terms have been used in literature involving humans and therefore, biological tissues. When replicating the response of the neck to external loading on mechanical neckforms, the terms must be contextualized to application. There have been few simulation studies which have referred to the manipulation of the compliance of the neckform. Those studies which have discussed the mechanical properties of the neckform with respect to its compliance during loading and unloading have used the term *stiffness* (Carlson, 2016; Haldin & Kleiven, 2013). The use of the term stiffness, as opposed to strength or torque, helps to describe the resistance to deformation of a mechanical neck model due to the absence of force-producing muscles, which provide active stiffness to a human neck (Seigler et al., 2015). Seigler et al. (2015) explained that passive protection of the cervical region from the ligaments, bones, discs, and connective tissue provides the “stiffness characteristics” to the spine and therefore only passive stiffness can be applied in a mechanical neck model.

A review of the literature, described above, led to the following operational definitions of neck strength, torque, and stiffness in the current research. Neck strength, or cervical muscle strength, was defined as the amount of force generated by the cervical musculature, which theoretically represented the effective moment of the cervical muscles. The term cervical muscle strength was used in Part I to help define the neck strength testing on a human sample. Meanwhile, neck torque was defined as the product of muscle force times the perpendicular distance from its line of action to the axis of rotation. Although neck torque was not measured on humans, it was used to adjust the stiffness of the mechanical neckform during the head impact simulations in Part II. Finally, stiffness was operationally defined as the resistance to deformation in response to an applied load and referred to the behaviour of the mechanical neckform in response to loading and unloading during impact in Part II.

Cervical muscle strength. It is theorized that athletes with stronger cervical musculature have a decreased risk of sustaining a concussion because they are able to mitigate the resultant accelerations applied to the head during direct and indirect loading (Eckner et al., 2014; Mihalik et al., 2011; Naish, Burnett, Burrows, Andrews, & Appleby, 2013; Schmidt et al., 2014). Since 80 percent of mechanical loading during impact is managed by the muscles of the neck, Schmidt et al. (2014) speculated that athletes who do not have sufficient cervical muscle strength are not able to generate the necessary preparatory and reactive forces to mitigate head acceleration. According to Eckner, Oh, Joshi, Richardson, and Ashton-Miller (2014), stronger muscles are able to generate more absolute force, torque, and greater tensile stiffness more rapidly than weaker cervical musculature. Contrarily, smaller and weaker cervical musculature causes the neck to be more compliant, thereby resulting in a greater overall linear and angular head displacement from an external perturbation (Eckner, et al., 2014). Collins et al. (2014) further postulated that head and neck anthropometrics, such as head and neck circumference, also play a role in concussion risk. The literature supporting neck muscle strength as a potential protective factor against concussions, however, remains inconclusive.

In a study examining the isometric cervical strength and head and neck anthropometrics of 6704 high school athletes, Collins et al. (2014) found that male and female athletes who had been diagnosed with a concussion had a significantly smaller mean neck circumference, significantly smaller mean neck circumference to head circumference ratio, and significantly less mean cervical muscle strength than those athletes who did not sustain a concussion, $p<.05$. There was an inverse relationship observed between cervical muscle strength and concussion risk; concussion risk was reduced by five percent for every one-pound increase in cervical muscle strength (Collins et al., 2014). Comparing male and female athletes, it was observed that females had a smaller mean neck circumference (M=32.55 cm, SD=2.35 cm) and a smaller neck to head circumference ratio (M=0.59, SD=0.04) than males' neck circumference (M=36.10 cm, SD=2.53 cm) and circumference ratio (M=0.64, SD=0.04). Female athletes also demonstrated less overall cervical muscle strength (M=8.28 lbs, SD=4.53 lbs) than male athletes (M=10.56 lbs, SD=5.41 lbs). From this comparison, one can infer that female athletes had weaker cervical muscles and a larger head in proportion to neck girth, which could result in greater difficulty controlling the movement of the head upon impact. Furthermore, when comparing female athletes separately, it was noted that the only significantly different anthropometric measure between athletes with and without concussion was the overall neck strength. Female athletes who sustained a concussion demonstrated weaker cervical muscle strength (M=7.41 lbs, SD=4.39 lbs) than those who did not sustain a concussion (M=8.28 lbs, SD=4.53 lbs). These findings support the literature suggesting cervical muscle strength to be a risk factor for concussions in female athletes.

The research results from Collins et al. (2014) were supported by other research examining the effect of cervical muscle strength on head impact biomechanics, which found a positive relationship between increased cervical muscle strength and decreased head acceleration in response to external perturbations (Gutierrez, Conte, & Lightbourne, 2014; Viano, Casson, & Pellman, 2007). In a study of 46 athletes who play contact sports, Eckner et al. (2014) examined the influence of cervical muscle strength and cervical muscle activation on resultant peak linear and angular velocity after inertial loading. Results of the study identified that greater isometric cervical strength and greater anticipatory response to inertial loads resulted in lower peak linear and angular velocities of the head (Eckner et al., 2014). These results highlight the controversy in the literature as to whether cervical muscle strength, and/or ability to develop tension in the cervical musculature quickly in response to an impact provides a greater protective factor against concussions.

Furthermore, Schmidt et al. (2014) examined isometric cervical strength, muscle size, and response to external perturbation in high school and collegiate football players. Study results identified that cervical strength alone was not able to mitigate the severity of the head impact. Rather, greater cervical stiffness reduced the odds sustaining higher magnitude head impacts. Similarly, Mihalik et al. (2011) found that cervical strength did not significantly reduce linear and angular head acceleration in a sample of youth hockey players. Cervical muscle strength as a protective factor would theoretically require the neck muscles to always be tense, or the athlete always anticipating a hit, which is not always the case (Mihalik et al., 2011). Other researchers also agree that cervical muscle strength alone cannot mitigate the risk of concussions because the muscles of the neck need to be contracted at the moment of impact to decrease the head displacement significantly enough to serve as a protective factor (Broglio, Eckner, & Kutcher, 2012; Mansell, Tierney, Sitler, Swanik, & Stearne, 2005).

Subsequently, it appears as though cervical muscle strength alone is not able to significantly reduce the risk of sustaining a concussion. Although, cervical muscle strengthening can increase the muscles' ability to develop stiffness in response to a perturbation (Eckner et al., 2014), increases in peak force and rate of force development (Sale, 1998), and can also lead to neuromuscular adaptations, such as improved muscle coactivation, improved proprioception, and greater stabilisation of the deep cervical flexors (Naish et al., 2013). Thus, despite the disparity in the research examining the influence of cervical muscle strength on concussions risk, it can be generally assumed that "interventions aimed at increasing athletes' neck girth, strength, and stiffness still hold promise as a means of reducing their risk of sport-related concussion" (Eckner et al., 2014, p. 567).

Head impact location. The location of head impact has also been shown to affect the head and neck biomechanics during impact due to the influence of linear and rotational acceleration applied to the head (Wilcox et al., 2015). There has been limited research investigating the influence of head impact location on head impact biomechanics; however, existing research seems to support similar relationships. Zhang, Yang, and King (2001) used three-dimensional finite element modelling to simulate head impacts to the front and lateral aspects of the head. Results of the study indicated that lateral impacts produced larger localized skull deformation and induced larger shear stress within the brain than frontal impacts (Zhang, Yang, & King, 2001). Similarly, previous research by Hodgson, Thomas and Khalil (1983) found that concussions occurred more frequently due to impacts to the temporo-parietal region, as compared to the frontal or occipital region in anesthetized monkeys. Gennarelli et al. (1982) also suggested that lateral head movement produces more severe diffuse brain damage than movements in front and rear movements. Furthermore, Walsh, Rousseau, and Hoshizaki (2011) identified that peak linear acceleration was largest in lateral head impacts when compared to front and rear impacts during simulated impacts using the Hybrid III headform. Thus, although the existing research regarding head impact location is limited, the results are consistent. Impacts to the lateral aspects of the headform typically result in greater peak linear acceleration, greater shear and tensile forces, and have been found to have greater overall impact magnitudes. Based on these findings, it is theorized that side head impacts contribute to a greater risk of sustaining a concussion upon impact when compared to other impact locations.

Impact mechanism. A concussion can be caused by two primary mechanisms: direct loading or inertial loading (King et al., 2014; King, Yang, Zhang, Hardy, & Viano, 2003; Meaney & Smith, 2011). Direct loading occurs when the head directly strikes, or is struck by, an external object or surface. In ice hockey, examples of direct loading include contact with the boards, ice, puck, or another player in hockey (Graham et al., 2014; King et al., 2003). Inertial loading occurs when a force is applied to another part of the body (usually the torso) and the force is transmitted to the head and neck causing an acceleration-deceleration response (King et al., 2003; Meaney & Smith, 2011). In other words, inertial loading occurs from impulsive head motions during which the head does not strike an object, such as during whiplash (Meaney & Smith, 2011). Both mechanisms generate different levels of linear and angular acceleration, which may influence the severity of the impact and the type of injury sustained. According to King, Yang, Zhang, Hardy, and Viano (2003), direct loading produces large linear accelerations causing more focal head injuries, while inertial loading produces greater angular accelerations which contributes to a higher risk of diffuse brain injuries. As discussed, both accelerations are almost always present in any impact; therefore, both impact mechanisms have the potential to cause this type of injury. There is limited research, however, comparing the risk of sustaining a concussion from direct loading versus inertial loading in practical and simulated settings.

To date, few studies have examined concussions resulting from inertial loading in applied or simulation-based settings. Ommaya (1966) exposed Rhesus monkeys to a severe extension/flexion whiplash-type head acceleration with a subsequent minor head impact. The angular accelerations of these head movements were upwards of 10,000 rad/s, but the whiplash movements appeared to result in acute subdural haematomas, gliding contusions of the brain, and spinal cord injuries rather than concussions (Ommaya, 1966). Other research manually-induced inertial loading on surrogate headforms that were anthropometrically similar to six-month old infants by shaking them, with the goal of mimicking "Shaken Baby Syndrome" (Duhaime et al., 1987). Results of the study revealed that the angular motions created by shaking the surrogate headform were well below the levels of accelerations indicative of brain strain. Other than these rather dated studies, applied concussion research has typically focused on head injuries resulting from direct head impacts. Although, a recent study examined the effect of neck stiffness, impact location, and impact mechanism on peak linear head acceleration during simulated head impacts (Pennock, Kivi, & Zerpa, 2017). The results of the study revealed that a whiplash mechanism produced significantly higher measures of peak linear acceleration than direct head impacts during the simulation testing, further supporting the potential role of impact mechanism in concussion risk (Pennock et al., 2017).

Understanding the mechanism of inertial loading and how it relates to concussions is potentially invaluable to appreciating the influence of impact mechanism on concussion risk in sports. Even without direct and visible head impact, significant acceleration/deceleration forces can have negative effects on brain tissue (Barth, Freeman, Broshek, & Varney, 2001). When the head undergoes rapid acceleration or deceleration resulting from an impact to the body, the brain rebounds due to the sudden change in motion, causing the brain to impact the inner lining of the skull (Barth et al., 2001; Hynes & Dickey, 2006). Subsequent shearing of the brain tissue occurs due to the rotational forces placed on the head from the inertial loading.

The literature suggests that inertial loading, as compared to direct head impacts, results in greater angular acceleration of the head and shearing of the intercranial tissues (Meaney & Smith, 2011). There are few applied studies, however, that have compared these two mechanisms through practical or simulation-based research. Since the influence of inertial loading in concussions is suggested in the literature but there is limited research to provide evidence for this role, gaining a better understanding of the relationship between inertial loading and concussions is imperative. Furthermore, concussions are a large concern in sports that prohibit intentional body contact, such as women's hockey. If intentional body contact, informally known as body checking, is not permitted in women's hockey, it may be assumed that fewer direct hits to the head occur. Instead, perhaps inertial loading is responsible for a large majority of head injuries in the sport as many of the collisions or impacts may be unexpected, limiting an athlete's ability to reduce the inertial loading applied to her head. Since previous concussion research has primarily examined direct head impacts (Beckwith et al., 2012; Brainard et al., 2012; Clark et al., 2016; Oeur, 2012; Walsh et al., 2011), examining other impact mechanisms, such as inertial loading, may provide valuable information on the influence of impact mechanism on head impact accelerations leading to concussions. Further research on the relationship between impacts involving inertial loading and head impact biomechanics may provide a better understanding of the influence of impact mechanism on concussions in sport.

Concussions in Women's Hockey

Participation in women's ice hockey has grown exponentially in the last 15 years (Decloe et al., 2014). The growth of the sport, however, has been paralleled with a gradual increase in injuries in female hockey players (Brainard et al., 2012). Several sources report injuries to the head and neck, including concussions, to be the most common injury sustained by female hockey players (Agel et al., 2007; Detis et al., 2010; Schick & Meeuwisse, 2003). Women's ice hockey differs from men's ice hockey in one significant official rule: the permission of intentional body contact. In women's hockey, intentional body contact is prohibited (Hockey Canada, 2003). Any intentional body contact in women's hockey is penalized, the severity of which depends on the characteristics of the contact. Although, despite rules prohibiting intentional body contact in women's hockey, female players sustain more concussions than male players (Brainard et al., 2012; Covassin, Moran, & Elbin, 2016). This unexpected trend is still unclear and is receiving increasing amounts of research to understand some of the factors influencing the concussion risk in women's hockey.

In a study by Brainard et al. (2012), two male and two female NCAA ice hockey teams wore instrumented helmet units equipped with single-axis linear accelerometers. The helmets, which were worn over two playing seasons, were designed to measure the magnitude and location of head impacts sustained by players. Results of this study revealed that the female players received fewer head impacts per season, lower angular accelerations, and lower overall magnitude impacts than male players. Since female hockey players are found to have a higher incidence of concussions than male players (Agel et al., 2007; Covassin, Swanik, & Sachs, 2003; Forward et al., 2014), these findings dispute the common theory that magnitude and/or frequency of impact alone contributes to concussion risk (Brainard et al., 2012). Supporting the research by Brainard et al., Covassin, Moran, and Elbin (2016) analyzed the injury profiles of 1702 concussed NCAA athletes and found that female ice hockey players had a 1.1 times greater risk of sustaining a concussion than male ice hockey players. Covassin et al. also found that female athletes generally had longer recovery times from their concussions than male athletes.

The reasoning behind the elevated risk for concussions in female ice hockey players remains unclear. Researchers have attempted to identify potential variables that may increase a female's predisposition to head injuries. Brainard et al. (2012) proposed that neck strength, body center of mass, brain morphology, weight and/or speed of the athlete, and lack of experience receiving and delivering body checks (intentional or unintentional) are all variables that may contribute to risk of concussion after head impact. When analyzing concussion risk in high school athletes, Collins et al. (2014) proposed that head and neck anthropometrics may play a role in concussion risk. Furthermore, Wilcox et al. (2015) identified physiological differences (e.g. anthropometrics) and hormonal differences as potential sex-specific risk factors for concussions. Wilcox et al. also emphasized that impact location, anticipation of impact, and cervical muscle strength are all important factors when analyzing head impacts, in addition to traditional acceleration measures. The factors that will be explored further in the current research include cervical muscle strength, head impact location, and impact mechanism.

Risk factors related to females.

Cervical muscle strength. Females have demonstrated weaker cervical musculature than males (Cagnie, Cools, De Lose, Cambier, & Danneels, 2007; Garcés, Medina, Milutinovic, Garavote, & Guerado, 2002; Salo et al., 2006). Research comparing the isometric cervical strength of males and females has reported males to have approximately 40% greater isometric strength in their cervical flexors and extensors than females (Cagnie et al., 2007; Garcés et al., 2002). This sex-based difference in cervical strength is consistent throughout the literature, including in an athletic population (Hildenbrand & Vasavada, 2013). According to Hildenbrand and Vasavada (2013), sex differences in cervical muscle strength begin to develop between the ages of 12 to 17 years, and sex is said to be more influential on cervical muscle strength than age, body weight, and body mass index (Peolsson & Öberg, 2001). To help explore the influence of cervical muscle strength as a protective factor in women’s hockey, it would be invaluable to have normative cervical muscle strength data for this specific population. This data, however, is scarce. Previous research has measured cervical isometric flexion and extension in female hockey players (Kivi et al., 2017), although increasing this normative data to include cervical isometric side flexion and a larger sample size would be beneficial.

Related to cervical muscle strength, neck stiffness and the ability of the muscles to store elastic energy is also greater in male athletes, which may help to explain why female athletes have a higher rate of concussions than male athletes (Schmidt et al., 2014). Neck stiffness relates to the ability of the cervical structures to resist deformation in response to an applied load and is reported to be linearly proportional to isometric muscle strength (Eckner et al., 2014). Therefore, the ability of a muscle or group of muscles to develop tensile stiffness would theoretically decrease the linear and angular displacement of the head and neck upon impact. Since cervical muscle strength is related to the ability of the muscles to develop tensile stiffness (Eckner et al., 2014), cervical muscle strength would prove to be an appropriate factor to explore in concussion research. Therefore, investigating the relationship between cervical muscle strength and concussion risk in women's hockey may contribute valuable information to the understanding of concussion disposition in this specific population.

Impact location. In ice hockey, there are multiple possible locations on the head that can be impacted during contact. According to NOCSAE and illustrated in Figure 4, there are six main head locations that are tested during head impact simulations, including front, front boss, rear, rear boss, side (left and/or right), and top (NOCSAE, 2017). The front and rear locations are defined by the intersection of the midsagittal and reference planes, the front boss is located 45 degrees from the front impact location, the rear boss is located 45 degrees from the rear impact location, and the side impact location is defined by the intersection of the coronal and reference plane. Each of these locations represents a realistic impact location in ice hockey, as direct or indirect loading can be sustained from any direction and angle. As discussed previously, lateral head impacts have been shown to lead to greater peak linear head acceleration and result in greater intercranial tissue strain than either front or rear impacts (Walsh et al., 2011; Zhang et al., 2001). The question remains whether impact location is a risk factor that may be contributing to the concussion risk in female hockey players.

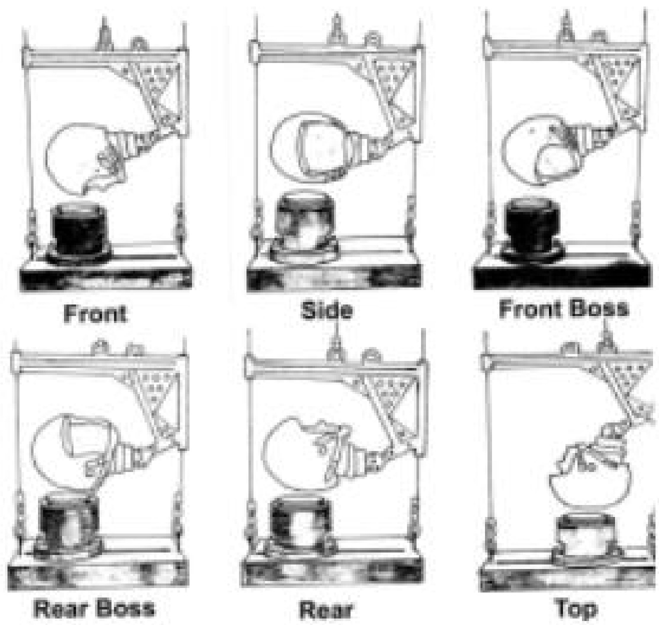

Figure 4. Impact locations as defined by NOCSAE standards. Adapted from "Standard test method and equipment used in evaluating the performance characteristics of headgear/equipment", National Operating Committee on Standards for Athletic Equipment (2017), Overland Park, USA: NOCSAE.

In a study analyzing head impact magnitude, frequency, and location in male and female NCAA hockey players, Brainard et al. (2012) found that female players received more frequent impacts to the right and left sides of the head as compared to males; however, males received more lateral impacts that were in the top one percentile of linear and angular acceleration ($\geq$80.4 g and $\geq$8375 rad/s^2). In other words, female hockey players received more lateral impacts, which were lower in magnitude than the impacts sustained by males, however female hockey players have been shown to sustain more concussions (Agel & Harvey, 2010; Brainard et al., 2012; Forward et al., 2014. These findings emphasize the influence that impact location may have on concussion risk, since high magnitude impacts were not needed to cause concussions in females. Other than the research conducted by Brainard et al., the research exploring the influence of impact location on concussion risk in specific populations, such as women's hockey, is scare. It is difficult to draw conclusions on the relationship between head impact location and concussions in women's hockey based on results from a single study. Although, based on the consistent trends in other research suggesting side impact locations to lead to greater concussion risk (Walsh et al., 2011; Zhang et al., 2001), exploring this observation further in relation to women's hockey has the potential to increase the evidence supporting impact location as a potential risk factor. Clarifying the role of head impact location on head kinematics related to concussions in simulated head impact research may then encourage future studies to examine this construct in a more applied context for women's hockey.

Impact mechanism. When examining the general injury profiles of female hockey players, previous literature has identified contact with other players, either intentional or unintentional, to be the most common source of injury in women's hockey (Decloe et al., 2014; Forward et al., 2014; Schick & Meeuwisse, 2003; Wilcox et al., 2014). This finding exists despite the official rule prohibiting intentional body contact in women's hockey. Nonetheless, given the unpredictable nature of body contact, it is valuable to understand how exactly injuries are sustained upon contact. As previously mentioned, two primary mechanisms cause concussions: direct impact to the head, or an impact applied to the body that is subsequently transmitted to the head, also referred to as a whiplash injury (Meaney & Smith, 2011).

Specific to women's hockey, a direct impact injury can result from a blow applied directly to the head; for example, a projectile to the head or an elbow to the head. Meanwhile, a whiplash injury could occur as a result of an open-ice hit in hockey, in which there is player-on-player contact, but no direct impact to the head. An additional potential impact mechanism, which combines the previous two mechanisms, involves a whiplash-type mechanism resulting in subsequent impact of the head. This proposed mechanism, combining whiplash with head contact, is a realistic impact mechanism in women's hockey as the unsuspecting player may be less prepared for the hit and not have the anticipatory response to contract the cervical musculature to help control the movement of the head. Consequently, head may contact the boards, ice surface, or another player following the whiplash-type movement of the head and neck. This additional mechanism, however, has not been explored in the research. Since each impact mechanism results in different amounts of linear and rotational acceleration acting on the head, the risk of sustaining a concussion should also vary.

Unfortunately, impact mechanism in women's ice hockey has not been well explored in the literature. In fact, to date there is no previous research documenting the behaviour of the head and neck during whiplash injuries as it relates to concussions in sport. Similarly, there is no data that has examined the head impact accelerations when the two primary impact mechanisms are combined, which limits the ability to identify the likelihood of sustaining a concussion during this impact mechanism. Due to the unpredictable nature of body contact in women's ice hockey, there is value in determining whether direct head impact, whiplash-type mechanisms, or a combination of the two mechanisms produces the greatest linear and angular acceleration of the head during impact, as this information could help coaching staff and clinicians be more mindful of potential concussive impacts during competition.

Protective Equipment

Helmets are a type of protective headgear worn in several sports, including ice hockey, to help reduce the risk of head injuries. Helmets are primarily designed to reduce the risk of a severe TBI such as a skull fracture, by decreasing the energy transfer to the head from the impact (Clark, et al., 2016; Daneshvar et al., 2011). Although, helmets have not proven to be adequately effective in reducing the risk of concussions in sport (Clark et al., 2016). Hockey helmets are currently designed with a semirigid outer shell and an inner protective foam liner made from either expanded polypropylene (EPP) or vinyl nitrate (VN) (Rousseau, et al., 2009). The outer shell is designed to spread the energy from the impact over a larger surface area to prevent focal injuries, while the interior liner acts to absorb the energy produced by the impact (Rousseau et al., 2009). Hockey helmets are considered multi-impact because they are designed to effectively manage the forces from repeated impacts. The linear material deforms upon impact and resumes its original state without losing any of its protective properties (Rousseau et al., 2009). There is a threshold, however, at which the helmet's materials begin to lose their effectiveness in managing impact forces. In a study examining the effect of neck stiffness, impact location, and peak linear acceleration during simulated head impacts, Carlson (2016) identified that, for the CCM V08 hockey helmet, the materials began to change its properties after approximately 90 head impacts. Although the helmet may still be protective after this point, its intended effectiveness has decreased.

Another component of hockey helmets which is rarely discussed is the role of the full cage facial shielding in protecting the head from concussive forces. Women's hockey mandates that players of all ages and skill levels must wear a full-face mask, either a cage or visor (Hockey Canada, 2003). There is limited research, however, examining the influence of a full-face cage on the accelerations experienced by the head during direct or indirect impact. Most research has found no difference in concussion rates between players with and without facial shielding (Benson, Rose, & Meeuwisse, 2002; Stevens, Lassonde, de Beaumont, & Keenan, 2006; Stuart, Smith, Malo-Ortiguera, Fischer, & Larson, 2002). Although, the presence of a cage alters the geometry of the helmet, which has the potential to influence the amount of force applied to the head upon impact. For example, Lemair and Pearsall (2007) assessed the differences in peak linear acceleration experienced by a surrogate headform during simulated head impacts when wearing visors versus full cages. Results of the study revealed that full cages significantly reduced the peak linear acceleration of the head upon impact, likely because a cage helps to distribute some of the forces radially and away from the head's centre of mass. The influence of full cages on the shear force applied to the head, however, has yet to be examined. Since full cage facial shields protrude outwards from the helmet, the tangential forces applied to the head during impact would be applied at a greater distance from the head's center of mass. This approach would theoretically result in an increase in the shear force and angular acceleration experienced by the head, potentially increasing an athlete's risk for sustaining a concussion. Helmet testing, however, occurs without the presence of a facial shield and is designed to test the helmet's ability to resist only linear accelerations (NOCSAE, 2017), since helmets are not currently designed with the goal of preventing concussions.

According to Clark, Post, Hoshizaki, and Gilchrist (2016), it is difficult to accurately test helmets for concussion protection within a laboratory setting because it is challenging to fully replicate the complex loading scenarios experienced in sports. Furthermore, helmets are designed to mitigate linear acceleration forces such as those with the potential to cause a skull fracture, but are not proficient at decreasing angular accelerations applied to the head during loading (Clark et al., 2016). Since helmet testing only examines linear acceleration, it is unclear how well helmets are able to decrease angular head acceleration, which is a large contributor to concussions (Rousseau et al., 2009). Since helmet protection is not guaranteed to prevent concussions resulting from head impacts in sport, it is invaluable to determine other factors relating to concussions that could be modified to minimize concussion risk, such as cervical muscle strength.

Testing Protocols Using Impactors

Headform and neckform. Using head impact simulations to investigate sports-related concussions is a relatively new but increasingly popular method of examining the head and neck kinematics related to head impacts. Previous research using head impactors to examine head impact biomechanics have used comparable protocols, including equipment, procedures, and outcome measures. Most previous simulation-based research has used the Hybrid III surrogate headform and neckform (Beckwith, Greenwald, & Chu, 2012; Clark et al., 2016; Ouer, 2012; Pellman, Viano, Casson, & Waeckerle, 2003; Walsh et al., 2011). Other studies have used the NOCSAE surrogate headform for simulated head impacts (Carlson, 2016; Cournoyer, Post, Rousseau, & Hoshizaki, 2016; Rowson & Duma, 2011; Sproule & Rowson, 2017; Zerpa, Carlson, Elyasi, Przysucha, & Hoshizaki, 2016). Surrogate headforms are equipped with accelerometers in different arrays to capture the linear and angular accelerations of the headform upon impact (MacAlister, 2013). The NOCSAE headform, although used less frequently in research than the Hyrbid III headform, is gender-neutral and is designed to represent the 50th percentile of an adult head. It is also considered to be more anatomically correct than the Hybrid III, including appropriate facial features and bone structures (MacAlister, 2013).

Although there are advantages to the gender neutrality of the NOCSAE headform, it may also present challenges in gender-specific research. There is emerging research addressing the limitation of the lack of female-specific surrogate devices in simulation-based concussion research. Despite females having a higher risk for concussions and whiplash injuries, a neckform representing the 50^{th} percentile for a female does not exist. Furthermore, the only headform representing the 50^{th} percentile of a female was designed via finite element modelling, but not manufactured for use (Östh, Mendoza-Vazquez, Sato, & Svensson, 2017; Vasavada, Danaraj, & Siegmund, 2008). According to MacAllister (2013), a 5^{th} percentile female surrogate headform has been designed, but its use in research is limited. The lack of a female-specific headform and neckform in concussion research may be problematic given the anthropometric differences between males and females.

When examining the anthropometric differences of the head and neck between males and females, it has been found that the neck is 9-16% larger in males, while the head is only 3-6% larger in males as compared to females (Vasavada et al., 2008). According to Vasavada et al. (2008), a female-specific neck model should be considered when studying sex-based differences in neck-related disorders because, not only is there a large magnitude anthropometric difference between male and female necks, there are geometric differences in the cervical spine between sexes. The geometric differences in the cervical spine of males and females make it unrealistic to scale male neck models to fit female parameters, a technique that has been attempted with surrogate headforms (Meijer, Wisgerhof, Wismans, & Been, 2009). Until female-specific models are designed and readily used in simulation-based concussion research, researchers must use gender-neutral models, understanding that there may be some limitations to the generalizability of the results of the research. In the current research, the head and neck anthropometrics of the female participants were measured and compared to the average parameters used to design the surrogate headform and neckform. This comparison provided an indication of the representativeness of the surrogate models to the female sample.

Impactor. Another defining aspect of simulation-based concussion research is the instrument used for delivering the head impact. In general, previous research has either utilized a drop rig to simulate vertical impacts, such as falls (Carlson, 2016; Clark et al., 2016; Zerpa et al., 2016), or a pneumatic linear impactor to simulate horizontal head impacts (Beckwith et al., 2012; Ouer, 2012; Walsh, et al., 2011). Vertical drop systems have been mono-rail (Clark et al., 2016) or dual-rail (Carlson, 2016; Zerpa et al., 2016), both allowing for a frictionless freefall of the surrogate headform onto an impact surface situated below. Additionally, Clark, Post, Hoshizaki, and Gilchrist (2016) used a pneumatic puck launcher to simulate projectile head impacts, while Pellman, Viano, Casson, and Waeckerle (2003) utilized a custom-designed system to achieve player-on-player impacts, reconstructed from NFL video analysis. The type of impactor used in simulation-based concussion research should align with the type of head impact that is being reconstructed.

Characteristics of the impact. Previous research has often been differentiated by which independent factors are manipulated in the simulated head impacts. Factors such as the impact speed, location, angle, and neck stiffness are strategically chosen to accurately recreate the head impacts and to isolate the desired risk factors being examined. In previous research, the speed at which the head was impacted depended largely on the real-life impact that was being simulated with the impactor. For example, Clark, Post, Hoshizaki, and Gilchrist (2016) used the velocities of 3, 5, and 7 meters per second to simulate falls and collisions, while puck impacts were delivered at velocities of 20, 30, and 40 meters per second. Other studies have impacted the head at 5.5 meters per second (Walsh et al., 2011; Ouer, 2012), 4.5 meters per second (Zerpa et al., 2016), or using a protocol of 18 different impact velocities, ranging from 2.62 to 4.85 meters per second (Carlson, 2016). Each of these impact velocities were strategically chosen to replicate the nature of the impact in real life.

Impact location is another variable that often differentiates simulation-based concussion research. All studies that were reviewed utilized between two and five different head impact locations (Carlson, 2016; Beckwith et al., 2012; Clark et al., 2016; Ouer, 2012; Pellman et al., 2003; Walsh et al., 2011; Zerpa et al., 2016). These impact locations typically involved the locations defined by NOCSAE standards, including front, rear, front boss, rear boss, and side (NOCSAE, 2017). Only certain studies, however, specifically altered the angle of impact at any of the selected locations to examine the differences in linear and angular acceleration experienced by the headform (Carlson, 2016; Ouer, 2012; Walsh et al., 2011).

Another variable that can be manipulated in simulation-based concussion research is the stiffness of the neckform. Although there is empirical evidence suggesting that cervical muscle strength plays a role in concussion risk (Eckner et al., 2014; Mihalik et al., 2011; Schmidt et al., 2014), few studies have specifically isolated characteristics of the neckform during simulated head impacts. Carlson (2016) examined the difference in linear acceleration, shear force, and GSI among three different neck stiffnesses, similar to research conducted by Rousseau and Hoshizaki (2009) assessing the effect of neck compliance and deflection on linear and angular acceleration, and GSI. With limited research using neck stiffness as an independent factor, it is difficult to draw conclusions about the effect of neck stiffness on measures of head impact biomechanics in simulation-based research.

Outcome measures. The outcome measures of previous simulation-based concussion research are comparable. Most studies recorded the peak linear and angular accelerations of impact, and at least one measure of injury severity (either GSI or HIC) (Carlson, 2016; Beckwith et al., 2012; Clark et al., 2016; Ouer, 2012; Pellman et al., 2003; Walsh et al., 2011; Zerpa et al., 2016). Only one study chose not to analyze angular acceleration (Zerpa et al., 2016) and another study measured shear forces rather than angular acceleration (Carlson, 2016). It has been noted, however, that when a combination of independent factors is manipulated within a simulation-based study, the results have typically been analyzed independently. In other words, previous studies have not examined the interaction between independent factors such as impact speed, location, angle, or neck stiffness. Given that concussions in real life are influenced by a complex interaction of external factors, it may be invaluable to examine the interaction of these independent factors in simulation-based concussion research as well.

Real-life data in simulations. With the emergence of simulation-based research, there has been speculation as to how well the simulated head impacts are able to mimic real-life head impacts in sports, including the accuracy of the outcome measures. Some researchers have used real-time videos from sport to recreate the collisions or other head impacts that occur, with the accurate impact speed, head impact location (Beckwith et al., 2012) and impact method (Pellman et al., 2003). Beckwith, Greenwald, and Chu (2012) sought to determine the correlation between on-field measures of head impact biomechanics, such as those obtained from the Head Impact Telemetry System (HITs), and impact simulations performed with surrogate headforms such as the Hybrid III headform. Results from the study revealed that simulation-based concussion research was highly correlated with on-field measures of peak linear acceleration, (r^2= 0.903), peak angular acceleration (r^2= 0.710-0.981), GSI (r^2= 0.846) and HIC (r^2= 0.787), deeming it is an acceptable and often more economical means of examining head impact biomechanics in sport (Beckwith et al., 2012). It has been proposed, however, that combining real-life data with simulated head impacts may provide even greater value to the results obtained in simulation-based research. Human data may increase the anthropometric and structural accuracy of the surrogates, in addition to making the head impacts more realistic.

Research Problem

Due to the growing concern surrounding head injuries in sport, the prevention of concussions has been identified as a research priority (Centers for Disease Control and Prevention, 2003). Research on concussion in sport has focused on contact sports with high concussions rates including American football, ice hockey, and rugby (Koh, Cassidy, & Watkinson, 2003). The problem exists, however, in the sex-based bias of the concussion research in sport, as there is very limited research on concussions in female athletes. This is concerning since females have a higher concussion risk than males and the increased predisposition to injury in females is still not well understood. Women's hockey, in particular, is a sport in which intentional body contact is not permitted, yet the rate of concussions is still high (Schick & Meeuwisse, 2003). It is not clear, however, why female hockey players are at a greater risk for concussions than their male counterparts because there has been limited research exploring potential factors influencing this risk.

Insufficient cervical muscle strength in females is one factor that has been suggested to be contributing to the high rate of concussions in women's hockey because the female players may less capable of effectively controlling their head in response to an external perturbation such as a collision (Brainard et al., 2012; Wilcox et al., 2015). The relationship between cervical muscle strength and head impact biomechanics specific to female hockey players has received little attention in previous research. In fact, there is limited data available documenting the cervical strength of female hockey players, which would provide invaluable information to studying this relationship further. One recent study documented the maximal isometric cervical muscle strength in female hockey players (Kivi et al., 2017), although this research did not measure lateral flexion strength. Lateral flexion strength in female hockey players may prove to be an important construct to include because of the higher frequency of side head impacts that occur in the sport. With this being said, impact location is another factor that has been examined in previous research related to concussions (Walsh et al., 2011; Zhang et al., 2001), as it is suspected to influence the linear and angular acceleration applied to the head. Since female players have been found to receive more impacts to the right and left sides of the heads than males (Brainard et al., 2012), examining the role of impact location in concussions in women's hockey could prove to be meaningful. A final factor that could be influencing the risk of concussions in women's hockey is the mechanism of injury. Concussions can occur via direct head impact, whiplash-type injuries, or a combination of the two mechanisms, although previous research has focused primarily on direct head impacts, which may not be representative of the majority of impacts experienced by female hockey players. Since many collisions in women's hockey occur unexpectedly and may include a whiplash and impact component, further investigation into the role of impact mechanism in concussions may improve the understanding of how concussions most often occur in the sport.

There is limited research examining potential factors contributing to the evident and concerning high rate of concussions in women's hockey. Without knowledge on risk factors such as cervical muscle strength, impact location, and impact mechanism, it is difficult to generate preventative strategies and policies to help reduce the risk of concussions in the sport. For example, if evidence suggests that cervical muscle strength is a primary factor increasing concussion risk in female hockey players, then potential injury prevention strategies could focus on increasing the strength of the cervical musculature in female players. To address this research problem, the current research was divided into two main parts, each with its own purpose. The purpose of Part I, discussed in Chapter 3, was to establish normative data describing strength and anthropometric measures in female ice hockey players. These normative data were then used in Part II of the study, which is discussed in Chapter 4. The purpose of Part II was to examine the effect of cervical muscle strength, helmet impact location, and impact mechanism on head impact biomechanics related to concussions using simulation impact testing.

Chapter 3- Part I: Human Neck Strength Testing

Purpose

The purpose of Part I of the study was to establish normative data describing strength and anthropometric measures in female ice hockey players.

Research Question

The following research questions were used to guide the purpose of Part I:

1. What is the 5^{th}, 10^{th}, 50^{th}, 90^{th}, and 95^{th} percentile of overall isometric cervical muscle strength from a sample of female hockey players?
2. What is the 5^{th}, 10^{th}, 50^{th}, 90^{th}, and 95^{th} percentile of the head mass, head circumference, neck circumference, and neck length of female ice hockey players?

Method

Participants

Competitive female ice hockey players between the ages of 17 and 30 were recruited from the Lakehead University and Confederation College women's hockey teams, and from teams in the Thunder Bay Women's Hockey Association (TBWHA) Senior House Division. A total of 25 participants (age = 22.1±2.6 years; playing experience = 15.8±2.7 years; body mass = 71.7±10.9 kg; height = 165.2±5.2 cm) were recruited to participate in isometric cervical muscle strength testing.

Inclusion and exclusion criteria. Participants were included if they had not been diagnosed with a concussion or other head/neck injury within the past six months to ensure that their ability to perform maximal isometric cervical contractions was not compromised. Any participant who had sustained a head or neck injury prior to six months ago that prevented them from participating in their sport, must have received medical clearance to return to play to be included in the study. Furthermore, the participants had to have played at a caliber equivalent to, or higher than their current caliber for the past three years and had to be an active player at the time of testing. These requirements were to ensure that the sample was representative of the population (i.e. competitive female hockey players). Participants had to be otherwise healthy, as determined by the ParQ form, and free of any other musculoskeletal disorders that limited their ability to perform maximal isometric cervical contractions safely, determined from a pre-screening questionnaire. Finally, to help prevent the risk of injuries, participants were excluded if they had insufficient cervical range of motion based on the normal parameters identified by Swinkels and Swinkels-Meewisse (2014), provided in Table 1.

Table 1

Normal values for cervical range of motion for ages 20-29 years, measured in degrees

Flexion	**Extension**	**Side Flexion (Left)**	**Side Flexion (Right)**	**Rotation Left**	**Rotation Right**
60 (10.92)	75 (10.34)	46 (7.50)	45 (7.47)	78 (7.97)	79 (6.63)

Adapted from "Normal values for cervical range of motion," by R. A. H. M. Swinkels & I. E. J. C. M. Swinkels-Meewisse, 2014, *Spine, 39*, pp. 362-367. Copyright © 2014 Lippincott Williams & Wilkins.

Instruments

Nautilus neck strength machine. The Nautilus neck strength machine (Figure 5) is a commercially available device that enables the user to develop the strength of their cervical musculature through neck flexion, extension, and lateral flexion isotonic exercises. It was chosen maximal isometric cervical strength testing because it is a standard piece of equipment used for cervical muscle strengthening and it has been used in previous research examining maximal isometric cervical strength (Broennle et al., 2017). The device was set up near a wall and bolted to the floor to ensure adequate stability. A strain gauge load cell, attached to the wall and connected perpendicular to the moveable arm of the Nautilus machine, was used to measure the cervical muscle force produced during the maximal isometric strength tests. The load cell was attached to the moveable arm by a lightweight chain, which caused the headrest to become immovable, providing a resistance against which the isometric contractions could be performed.

Figure 5. The Nautilus neck strength machine. The lightweight chain attached the load cell to the headrest and prevented the moveable arm from moving during the isometric contractions.

Procedures

Participant recruitment. Once ethical approval was received from the Lakehead University Research Ethics Board, participants were recruited through a combination of convenience and purposive sampling. With approval from the teams' coaches, players were recruited in-person, through in-person an information session, or via email. All interested players who met the inclusion criteria were asked to email the student researcher for further information about participating in the study.

Preliminary measures. The maximal isometric muscle force production in cervical flexion, extension, and side flexion was measured on a sample of female hockey players during a single testing session lasting approximately 60 minutes. Testing occurred in the Exercise Physiology Laboratory (SB1025) in the Lakehead University Sander's Fieldhouse. The participants were first asked to read and complete an informed consent form (Appendix A), a ParQ form (Appendix B), and a pre-screening questionnaire (Appendix C) asking about head and neck injury history in their sport. Next, to minimize risk of injury, participants completed a dynamic warmup before engaging in any testing measures. The dynamic warmup included a five-minute stationary cycle at a moderate pace to increase core body temperature, followed by dynamic neck stretches (Appendix D). A similar warmup protocol was included in previous research that required participants to perform maximal isometric cervical muscle strength testing (Broennle et al., 2017).

After the completion of the warm-up, the participants' neck range of motion was measured in flexion, extension, lateral flexion, and left and right rotation to ensure that the participants were within the normal range for each movement direction, as defined by Swinkels and Swinkels-Meewisse (2014). Measuring neck range of motion was another method used to help rule out any musculoskeletal disorders or other injuries to the neck that may have limited the participants' ability to perform maximal isometric cervical contractions safely. Range of motion was measured using the Cervical Range of Motion (CROM) device (Figure 6), which is an instrument used in clinical settings to measure cervical range of motion. The CROM device has been used to measure cervical range of motion in previous research (Fernández-Pérez et al., 2012; Kalle, Krakenes, Albrektsen, & Wester, 2007; Osterbauer et al., 1996). The device has demonstrated high concurrent validity with Fastrak motion analysis system on measures of cervical range of motion in all movement directions, with correlation values ranging from $r=0.93$ (flexion) to $r= 0.98$ (extension, left and right rotation), and high between-day test-retest reliability (ICC= 0.89-0.98) in a sample of 20 males and females (Audette, Dumas, Côté, & De Sarres, 2010). The CROM was positioned on the participant's head with the arms of the device resting on the ears and secured behind the head with the head. The position of the device was standardized by ensuring that each of the three inclinometers measured 0-degrees when the participant was sitting in an upright posture and looking straight ahead prior to testing.

Figure 6. The CROM device. This headpiece is worn much like a pair of glasses and secured behind the head with a strap. There are three inclinometers that are strategically positioned in each of the planes of motion to measure range of motion in all directions.

Total height, body mass, and the head and neck anthropometrics of the participants were also measured prior to strength testing. Neck length, neck circumference, and head circumference were measured with a measuring tape using the same anatomical landmarks as Seigler et al. (2015). Neck length was measured from the occipital condyle to the C7-T1 point; neck circumference was measured around the laryngeal prominence; and head circumference was measured just above the level of the ears. These anatomical landmarks are comparable to those utilized in other studies (Collins et al., 2014). Head mass was estimated using the body segment parameter data from deLeva (1996), which was based on the data from Zatsiorsky, Seluyanov, and Chugunova (1990) (Appendix E). These parameters were chosen because they were believed to be most representative of this study's sample of participants.

Neck strength testing protocol. To begin the neck strength testing, the Nautilus neck strength machine was adjusted to appropriately fit each participant. The seat was set at a height to allow the participant's forehead to be positioned on the headrest padding such that the horizontal metal bar located on the headrest was at eye level. The participants performed up to three practice trials at a submaximal effort to familiarize themselves with the testing protocol. Executed in each of the three directions, the participants were instructed steadily increase the amount of force applied over three seconds, until they reached an isometric maximum. The maximum isometric force was held consistent for an additional two seconds. This two-stage protocol was the same as the neck strength testing protocol used in previous research (Broennle et al., 2017) and is an effective strategy for reducing the risk of muscle strains during high effort strength testing. The participants were asked to choose their preferred side for the side flexion movements and the peak isometric force was applied at an angle of 10 degrees for each direction, to align with previous research (Garcés et al., 2002). Verbal encouragement was provided for all trials and the peak force production, in Newtons, was measured by the load cell and captured by LabChart 7 software. Following the neck strength testing, the participants were guided through a series of static neck stretches as a cool down (Appendix D).

Participants performed three maximal isometric contractions in each flexion, extension, and side flexion, completed in that order, with a three-minute break between each trial to reduce the effects of muscle fatigue. The average maximum force value of the three trials was calculated for each movement direction. For the purposes of this study, the average overall cervical muscle strength was calculated by computing the mean of the averaged maximal force measures for flexion, extension, and side flexion. This method of developing an average overall cervical muscle strength was adopted from Collins et al. (2014). Average overall cervical muscle strength was calculated because the design of the mechanical neckform used in Part II of this study only permits the modelling of overall cervical muscle strength, rather than isolated cervical muscle strength in each direction. Therefore, calculating a single measure for overall cervical muscle strength allowed for a more direct transfer of data from human participants to the mechanical neckform used during impact testing in Part II. Although overall cervical muscle strength was utilized in Part II of the study, the 5^{th}, 10^{th}, 50^{th}, 90^{th}, and 95^{th} percentile rankings for flexion, extension, and side flexion were also reported in the normative data.

Data Analysis

Data was analyzed using SPSS Statistics 25. To determine the cervical muscle strength of this sample of female hockey players, the Shapiro-Wilk test of normality was conducted to determine if the overall cervical muscle strength measures of the participants were normally distributed. Next, descriptive statistics were used to calculate the 5^{th}, 10^{th}, 50^{th}, 90^{th}, and 95^{th} percentiles of the cervical muscle strength in flexion, extension, and side flexion, as well as overall cervical muscle strength. Standard deviation of the scores were also calculated. To identify the head and neck anthropometric measures of this sample of female hockey players, the Shapiro-Wilk test of normality was conducted once again to determine if each anthropometric measure was normally distributed. Descriptive statistics were then used to calculate the 5^{th}, 10^{th}, 50^{th}, 90^{th}, and 95^{th} percentiles of head mass, head circumference, neck circumference, and neck length. Standard deviation of each anthropometric measure was also calculated.

Results

For the first research question in Part I, the Shapiro-Wilk test of normality identified that the overall cervical muscle strength of female hockey players was normally distributed, $p>0.05$, as reported in Table 2. Additionally, the histogram in Figure 7 visually illustrates the normal distribution of the data.

Table 2

Shapiro-Wilk test of normality assessing average overall cervical muscle strength, in Newtons

	Statistic	Degrees of Freedom	Significance
Average Overall Cervical Muscle Strength (N)	.965	25	.521

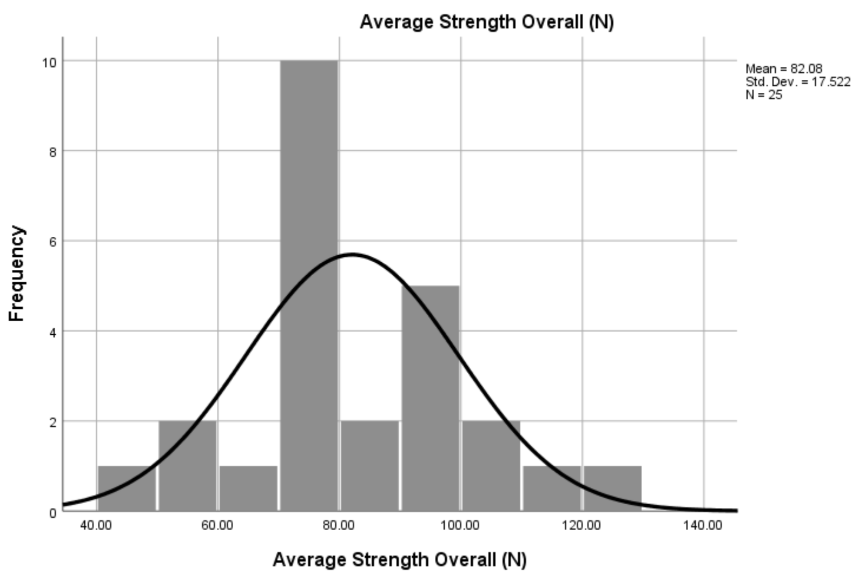

Figure 7. Histogram illustrating the normal distribution of overall cervical muscle strength measures in a sample of female hockey players.

Since the overall cervical muscle strength data exhibited a normal distribution pattern, the 5th, 10th, 50th, 90th, and 95th percentiles were calculated to provide a set of normative data characterizing the cervical muscle strength of female hockey players. Part II of the study only utilized the 10th, 50th, and 90th percentiles of overall cervical muscle strength, however, the 5th and 95th percentiles were also calculated to provide a more robust set of data. Furthermore, since overall cervical muscle strength was calculated from an average of the flexion, extension, and side flexion isometric strength measures, normative data was generated from these data as well.

Table 3 summarizes the percentile rankings for maximal isometric neck flexion, extension, side flexion, and overall cervical muscle strength, along with the standard deviation for each direction. The normative data reveals that, in general, neck extension was the strongest of the three movements, while neck flexion was the weakest movement across all percentile rankings in the sample of female hockey players.

Table 3

Cervical muscle strength values of female ice hockey players, in Newtons

	5th	**10th**	**50th**	**90th**	**95th**	**SD**
Flexion	43.67	54.27	72.62	91.82	101.20	14.94
Extension	53.56	59.67	95.00	136.73	153.49	25.12
Side Flexion	53.74	56.17	76.47	98.66	126.66	19.24
Overall	51.37	58.64	76.01	108.27	121.16	17.52

For the second research question in Part I, the Shapiro-Wilk test of normality revealed that measures of head mass, head circumference, neck circumference, and neck length were all normally distributed, $p>.05$ (Table 4). Since each of the anthropometric measures were normally distributed, the 5th, 10th, 50th, 90th, and 95th percentiles were calculated and summarized in Table 5. Part II of the study only utilized the 50th percentile of head mass and head circumference; however, all of the identified percentiles were calculated and reported.

Table 4

Shapiro-Wilk test of normality assessing head and neck anthropometric measures

	Statistic	**Degrees of Freedom**	**Significance**
Head Mass (kg)	.95	25	.89
Head Circumference (cm)	.98	25	.69
Neck Circumference (cm)	.97	25	.76
Neck Length (cm)	.98	25	.28

Table 5

Distribution of head and neck anthropometric measures for a sample of female ice hockey players

	5^{th}	10^{th}	50^{th}	90^{th}	95^{th}	SD
Neck Length (cm)	6.6	7.0	8.3	9.2	9.8	0.7
Neck Circumference (cm)	30.7	31.4	34.0	36.1	36.6	1.7
Head Circumference (cm)	52.5	53.5	56.1	58.8	59.7	2.0
Head Mass (kg)	3.6	3.9	4.8	5.8	6.5	0.7

Discussion

Cervical muscle strength is a factor that many researchers suggest is linked to concussions in athletes, including female hockey players (Brainard et al., 2012; Collins et al., 2014; Eckner et al., 2014). Stronger cervical musculature may have the ability to mitigate external forces applied to the head, thereby reducing the risk of sustaining a concussion from an impact (Eckner et al., 2014). Despite this purported relationship, there is limited normative data available measuring the cervical muscle strength of female hockey players, making it difficult to assess the influence of cervical muscle strength on concussion risk in the population. The purpose of Part I of this study was to develop a set of normative data describing the cervical muscle strength and anthropometric measures of a sample of female hockey players to establish a more robust dataset upon which further research may be positioned. The 10^{th}, 50^{th}, and 90^{th} percentiles were used to calculate the appropriate neckform torques for Part II of the study, explained in Appendix F. The 5^{th} and 95^{th} percentiles of overall cervical muscle strength were also presented, however, because these are the normative cut-offs most commonly used in anthropometric statistics (Kroemer, Kroemer, & Kroemer-Elbert, 2001).

As mentioned, there is limited existing cervical muscle strength data specific to female hockey players, making it difficult to compare the current results to an established dataset. Kivi et al. (2017) examined the maximal isometric flexion and extension cervical muscle strength of competitive female hockey players in comparison to a control group. Although the current study used similar testing protocols and a comparable sample to the research conducted by Kivi et al., the isometric neck flexion (M=72.62 N, SD=14.94 N) and extension values (M=95.00 N, SD=25.12 N) for the 50th percentile in the current study were slightly lower than those reported by Kivi et al. (flexion M=95.2 N, SD=27.4 N; extension M=121.4 N, SD=43.5 N). The differences between reported isometric cervical muscle strength values may be attributed to the methods used for data analysis, as the current study reported isometric maximal force as an average of three trials while Kivi et al. reported the peak value for isometric force production. The data presented from this study is valuable because it reports a range of percentile values, rather than only mean cervical muscle strength measures. Consequently, the data can be applied to a more diverse range within the population, thereby increasing the representativeness of the strength measures.

Furthermore, epidemiological studies have found that females have lower cervical muscle strength than males (Cagnie et al., 2007; Garcés et al., 2002; Salo et al., 2006), and this appears to be consistent when comparing the isometric cervical strength of the current sample of female hockey players to a sample of male hockey players. Broennle (2011) measured the isometric and isotonic cervical strength of a sample of competitive male hockey players using the Nautilus neck strength machine and testing protocols similar to those used in the current study. The mean isometric cervical flexion and extension strength the male hockey players (flexion: M= 162.9 N, SD= 52.3 N; extension: M= 208.6 N, SD= 41.6 N) was much greater than the mean cervical muscle strength exhibited by the female players in the current study (flexion: M= 72.62 N, SD= 14.94 N; extension: M= 95.0 N, SD= 25.12 N). This comparison emphasizes the difference in cervical muscle strength between male and female athletes, which may strengthen the rationale for examining the influence of cervical muscle strength in concussion injuries for female hockey players.

In addition to cervical muscle strength, previous research has identified the potential role of head and neck anthropometry in concussions (Collins et al. 2014). It has been suggested that athletes with a smaller neck circumference to head circumference ratio, smaller neck circumference, and a greater head mass may be more susceptible to concussions due to the decreased ability to mitigate biomechanical forces applied to the head during an impact (Collins et al., 2014). Various head and neck anthropometric measures were collected in the current study, including neck length, head and neck circumference, and estimated head mass. The 50th percentile of head circumference and head mass were the only anthropometric measures used in Part II of the study; however, the 5th, 10th, 50th, 90th, and 95th percentiles of all anthropometric measures were calculated and reported to generate normative data that can be used in future research.

The anthropometric measures collected in the current study appear to be comparable to data collected by Vasavada et al. (2008) in a sample of adult females, with the exception of the neck length measurements. Differences in reported neck length measures, however, may be attributed to the anatomical landmarks used to measure neck length. The current research used the occipital condyle and C7 as anatomical landmarks when measuring neck length, while Vasavada et al. measured from the left and right tragi to C7, which may help explain why reported average neck length in the current study (M=8.3 cm, SD=0.9 cm) was much lower than the reported measure by Vasavada et al. (M=10.7 cm, SD=.50 cm).

According to Collins et al. (2014), differences in anthropometric measurements may be useful in identifying athletes with an elevated risk of concussion, potentially encouraging the development of screening tools to help identify these more vulnerable athletes. The anthropometric measures of female hockey players from the current study were compared to those of male hockey players measured by Broennle (2011) using comparable landmarks. Interestingly, females appeared to have smaller mean neck circumference (M= 34.0 cm, SD= 1.7 cm) and neck length (M= 8.3 cm, SD= 0.9 cm) than males (neck circumference: M= 39.0 cm, SD=1.6 cm; neck length: M= 11.9 cm, SD= 1.4 cm), raising the concern regarding the potential role of head and neck circumference in the heightened concussion risk for female athletes. Therefore, since existing anthropometric data specific to female hockey players is scarce, the data collected in the current study can be used for research involving looking to explore the relationship between anthropometric measures and concussions in the sport.

Implications

The relationship between cervical muscle strength and concussion risk in female hockey players deserves more attention. If evidence consistently suggests that cervical muscle strength is a modifiable risk factor for concussions in the women's hockey, there is the potential for the development of cervical muscle strengthening intervention strategies to help reduce injury risk in the sport. The first step in examining the relationship between cervical muscle strength and concussions in women's hockey, however, is to establish normative data describing the cervical muscle strength of female hockey players. The normative data developed is valuable because it not only provides data for the maximal isometric cervical muscle strength in each movement direction, but it also combines the data into average overall cervical muscle strength measures. There is a strong rationale for using overall cervical muscle strength measures in addition to strength measures for isolated movements. Some neck muscles are responsible for multiple movements of the neck, such as the sternocleidomastoid muscle, which plays a role in neck flexion, extension, and lateral flexion, depending on which muscle fibres are activated and if activation occurs unilaterally or bilaterally (Tortora & Nielson, 2012). Therefore, when the head experiences an impact or loading from an external force, the various muscles of the neck often act synergistically to help control the movement of the head (Conley, Meyer, Bloomberg, Feedback, & Dudley, 1995). Since there may be multiple muscles acting simultaneously, it is appropriate to examine the collective strength of the muscles responsible for each movement direction.

Furthermore, average overall cervical muscle strength measures, as opposed to individual strength measures, may be required for methodological purposes in future research. For example, the average overall cervical muscle strength was required in Part II for the head impact simulations. To appropriately model the cervical muscle strength of human participants on the mechanical neckform for head impact testing, an overall strength measure was required based on the properties of the neckform. Because the stiffness of the neckform is adjusted by adjusting the torque of the cable running longitudinally through neckform, adjusting the neckform torque results in a uniform stiffness that affects all directions of neck movement. Using this strategy, however, allowed human cervical muscle strength data to be applied to simulation-based concussion research.

Finally, the 50th percentile head circumference and head mass were used in Part II of this research to determine how well the surrogate headform used during the head impact testing represented a human female head. Comparing the head anthropometrics of the human sample and the surrogate provided an indication of the relative representativeness of the surrogate, which is an important consideration when examining a specific population using a gender-neutral surrogate device. Moreover, the normative data for the other anthropometric measures, although not utilized in the current study, can provide invaluable information for future research investigating the role of head and neck anthropometric measures in concussion risk for this population.

Neck strength transformation. In the current research, the 10th, 50th, and 90th percentiles of overall cervical muscle strength were modelled on the mechanical neckform in Part II to examine the effect of cervical muscle strength on peak linear acceleration, peak shear force, and severity index during simulated head impacts. To appropriately model cervical muscle strength on the mechanical neckform, however, the strength measures had to be converted to torque measures. The transformation of neck strength to neckform torque was accomplished by scaling the cervical muscle strength data and using a preestablished calibration equation to generate corresponding torque measures that could be modelled on the neckform. Following this procedure, the 10th, 50th, and 90th percentile of human cervical muscle strength data, which represented *weak*, *average*, and *strong* neck strengths, corresponded to torque measures of 1.36 Nm, 2.94 Nm, and 4.62 Nm, respectively. Due to this transformation process and the mechanical nature of the neckform, the term *neckform torque* was used in Part II to describe the stiffness of the neckform, in comparison to the *cervical muscle strength* measured from human participants in Part I. A detailed explanation of the conversion of overall neck strength to neckform torque measures for the purpose of simulation testing is provided in Appendix C and summarized in Table 6.

Table 6

Summary of the torque measures derived from the cervical muscle strength data

	10^{th} percentile	**50^{th} percentile**	**90^{th} percentile**
Strength (N)	58.64	76.01	108.27
Torque (Nm)	1.36	2.94	4.62
Relative Strength Measure	Weak	Average	Strong

Chapter 4- Part II: Head Impact Testing

Purpose

The purpose of Part II was to examine the effect of neckform torque, head impact location, and simulated impact mechanism on head impact biomechanics in female hockey players using a simulation-based approach.

Research Question

The following questions were used to guide the purpose of Part II of the study:

1. How well does the medium-sized NOCSAE headform accurately represent a female human head?
2. What is the effect of neckform torque, head impact location, and simulated impact mechanism on peak linear acceleration of the head during simulated head impacts?
3. What is the effect of neckform torque, head impact location, and simulated impact mechanism on the peak shear force applied to the head during simulated head impacts?
4. What is the effect of neckform torque, head impact location, and simulated impact mechanism on Gadd Severity Index, during simulated head impacts?

Method

Instruments

Headform. The medium-sized NOCSAE headform (Figure 8) was used for the simulated head impact testing. Although used less frequently in research than the Hybrid III headform, the NOCSAE headform is more anatomically accurate than the Hybrid III headform because of the bone structure and facial features (MacAlister, 2013). It was also designed with the dimensions to represent the 50th percentile of an adult head, presented in Table 7. Although the NOCSAE and Hybrid III headforms differ in their anatomical structure and facial features, the dimensions of the headforms are comparable, making it acceptable to consider Hybrid III parameters for the NOCSAE headform. Furthermore, the NOCSAE headform was originally designed to be mounted on a rigid arm (MacAlister, 2013); however, it has been modified to be mounted on a mechanical neckform to allow for the neckform stiffness to be adjusted during head impact testing. The NOCSAE headform is instrumented with an array of triaxial accelerometers that measure the linear acceleration of the head in the anterior-posterior, superior-inferior, and the left-right directions (MacAlister, 2013). The accelerometers' data can then be used to calculate GSI, which can inform researchers on the severity of the simulated head impact (MacAlister, 2013). The medium-sized NOCSAE headform, specifically, has been used in previous studies examining linear head acceleration in simulation-based concussion research (Carlson, 2016; Rowson & Duma, 2013; Zerpa et al., 2016).

Table 7

Approximate measurements of the medium-sized NOCSAE headform, inches (cm)

Points of Measure	**Medium-Sized Dimensions**
Head Breadth	5.98 (15.2)
Maximum Brow Width (frontal diameter)	5.20 (13.2)
Ear Hole to Ear Hole (bitragion diameter)	5.51 (14.0)
Maximum Jaw Width (bigonial diameter)	4.65 (11.8)
Head Length (glabella landmark to back of head)	7.87 (20.0)
Outside Eye Corner (external canthus) to back of the head	6.81 (17.3)
Ear hole (tragion) to back of head	3.86 (9.8)
Ear hole to outside corner of eye (tragion to ext. canthus)	2.95 (7.5)
Ear hole to top of head (tragion to vertex)	5.24 (13.3)
Eye pupil to top of head	4.53 (11.5)
Ear hole to jaw angle (tragion to gonion)	3.03 (7.7)
Bottom of nose to point of chin (subnasal to menton)	2.80 (7.1)
Top of nose to point of chin	4.88 (12.4)
Head Circumference	22.68 (57.6)
Head weight, including mounting interface	10.8 lb (4.90 kg)

Adapted from "Standard test method and equipment used in evaluating the performance characteristics of headgear/equipment," by National Operating Committee on the Standards for Athletic Equipment, (ND 001-17m17), 2017. Overland Park, USA.

Figure 8. The NOCSAE headform. The headform is designed to represent the 50th percentile of an adult head and has anatomically correct bone structure and facial features. The headform was equipped with a medium-size CCM VO8 helmet.

Accelerometers and software. The accelerometers in the NOCSAE headform are piezoelectric sensors that measure linear acceleration in the x, y, and z directions by converting mechanical energy into electrical energy. The accelerometers are connected to a power supply, PCB model 482A04 integrated circuit, and amplifier, which transmit three analog signals to an A/D Instruments Powerlab 16/30. Each analog channel is assigned an accelerometer reading to capture and convert linear acceleration measures into digital format for the x, y and z directions. The digital data for each channel is scaled to “g” measures using a multiplying factor of 10 mv. A resultant acceleration vector is computed by adding the scaled acceleration data for each channel using Equation 10. For the current study, data were sampled at 20 kHz and high frequency noise was minimize using a low pass digital filter with cut off frequency of 1000 Hz to comply with existing literature.

$$\text{Resultant Acceleration} = \sqrt{x^2 + y^2 + z^2} \tag{10}$$

Where:

x = linear acceleration in the x-direction
y = linear acceleration in the y-direction
z = acceleration in the z-direction

Mechanical neckform. The mechanical neckform (Figure 9) was developed by the Lakehead University Mechanical Engineering department in conjunction with the School of Kinesiology. It was constructed from neoprene rubber with steel discs to represent the intervertebral discs of the cervical spine. The neckform was designed to represent the 50th percentile of a human neck (Spittle, Shipley Jr., Kalep, & Miller, 1992), aligning with the 50th percentile NOCSAE headform. The Hybrid III ATP parameters used to create the neckform are provided in Table 8, both in Standard International (SI) and Empirical (U.S Customary) units of measure with given tolerances.

Table 8

Hybrid 50th percentile neck parameters

	Neck Parameter
Weight (kg)	1.54 $\pm$0.05
Weight (lbs.)	3.40$\pm$0.10
Neck Extension (deg.)	60^o
Neck Flexion (deg.)	50^o
Lateral Flexion, Left (deg.)	45^o
Lateral Flexion, Right (deg.)	45^o
Neck Rotation, Left (deg.)	80^o
Neck Rotation, Right (deg.)	80^o

Adapted from "Hybrid II and Hybrid III dummy neck properties for computer modelling," by E. K. Spittle, B. W., Shipley Jr., I. Kaleps, & D. J. Miller, 1992. Air Force Systems Command, Wright-Patterson Air Force Base, Ohio.

To better simulate the dynamic response of the neck during loading and unloading, a small anterior cutout of the cross-section of the intervertebral disk and a larger posterior cutout were made, which are visible in Figure 9. This design allows the neckform to better mimic the various movements of a human neck during an impact (Spittle et al., 1992). Another important feature of the neck is a galvanized stainless-steel cable that runs longitudinally through the center of the neck. The cable ensures the steel discs and rubber remain firmly together, which maintains the integrity of the structure during impact testing. The cable also allows for the adjustment of the torque of the neck. Adjusting the torque of the neck adjusts the stiffness, which increases or decreases the compliance of the neckform during loading and unloading.

Figure 9. Mechanical neckform attached to the circular endplate. The mechanical neckform features large posterior cutouts and smaller anterior cutouts to help mimic the movements of a true neck. The circular endplate has eight holes drilled around the outer edge to allow the head and neck complex to be attached to the drop carriage.

The headform and mechanical neckform were mounted to a circular endplate (Figure 9). The circular steel bracket features eight holes arranged around the outside edge, which are used to bolt the head and neck complex to the drop carriage on the vertical drop system. Galvanized bolts were used to secure the head and neck complex to the drop carriage in the desired orientation. Ultimately, the circular endplate permitted the manipulation of the position of the head, relative to the contact surface, thereby influencing the impact location on the head.

Helmet. The medium-size CCM VO8 helmet was used to protect the headform during head impact testing (Figure 8). This helmet is lined with vinyl nitrate foam and the dimensions of the circumference of the helmet are adjustable, ranging from 22 ½ to 24 ¼ inches (57.15-61.60

cm), to ensure a snug fit on the wearer's head. These commercially available helmets, worn by both female and male hockey players, have been used in previous head impact research and the deterioration threshold of the helmet has been tested (Carlson, 2016). Carlson (2016) showed that the peak force increased drastically after 189 impacts to the front location, while repetitive impacts to the rear location revealed a gradual increase in peak force after approximately 90 impacts. Since helmet performance appears to deteriorate over time, helmets should be replaced after a large number of impacts. To be conservative, in this study the helmets were replaced with an identical new CCM VO8 helmet after 90 impacts. All helmets also featured a full cage facial shield due to the equipment regulations of women's hockey (International Ice Hockey Federation, 2015). The facial cage used was the Bauer 2100 True Vision face mask, which was replaced with an identical cage each time the helmet was replaced.

Proper helmet fit was maintained for all trials, according to manufacturers fitting instructions. There are limitations to achieving the most optimal helmet fit on a surrogate headform as compared to on a human head. Nonetheless, the helmet was fit most appropriately to the headform by ensuring that the distance between the brim of the helmet and the bridge of the nose was 5.5 cm and all straps were secured, leaving room for one finger between the chin strap and the chin. The helmet fit was remeasured and repositioned if necessary after each impact to maintain consistency.

Drop system. A dual rail drop system was used in the head impact simulations (Figure 10). The drop system was designed by the Lakehead University Mechanical Engineering Department, in collaboration with the Lakehead University School of Kinesiology. The system features a drop carriage, to which the headform was mounted. The position of the headform in the carriage was manipulated simply by orienting the head in the appropriate direction for impact. This subsequently controlled which location of the head was impacted when the drop carriage was released. The drop carriage is attached to the vertical rails of the system and moves along the rails with little friction, allowing a free fall drop. The movement of the drop carriage is remotely controlled by a 110-volt AC winch, which is attached to the carriage using high powered magnets. Once the carriage was raised to the appropriate height, the magnets were demagnetized and released by manually pushing a button on an electronics controller, and the carriage fell onto the AMTI force platform situated below. The combined weight of the headform, neckform, and drop carriage was 30.6 kg and this remained consistent throughout the testing.

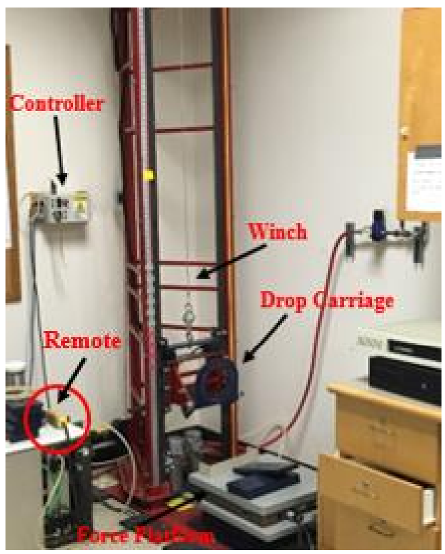

Figure 10. The dual-rail vertical drop system. The remote controller, featured on the left, controlled the winch to raise and lower the drop carriage to the desired height. The controller on the wall features the release button, which allowed the drop carriage to freefall and the headform to impact the force plate below.

The reliability and concurrent validity of the Lakehead University dual rail drop system was investigated previously (Carlson et al., 2016). The system was shown to have strong evidence of reliability (ICC=0.922, $p<0.005$) and strong evidence of concurrent validity on measure of peak linear acceleration (ICC=0.844-0.952, $p<0.005$). In the validity testing conducted by Carlson et al. (2016), the Lakehead University drop system was compared to the

University of Ottawa Neurotrauma Science research lab impact acceleration measures drop system to provide evidence of concurrent validity.

Force platform. The AMTI OR6-5-1 force platform (Figure 11) was used to measure impact forces applied to the head when dropped. An angled plate was mounted on the platform at 13.5 degrees to generate shear forces during the impacts. The force platform was mounted to a frame at the base of the dual rail drop system and simultaneously measured three force components along the x, y, and z axes and three moment components about the x, y, and z axes (Advanced Mechanical Technology, Inc [AMTI], 1987). These values are measured by foil strain gauges configured as wheatstone bridges to form the load cells at the four sides of the platform to produce output forces (AMTI, 1987). The upper limits of platform loading are 2200 lbs (9800 N) of vertical load applied anywhere on the top surface or 1200 lbs (6700 N) of side load applied anywhere in the x or y direction without damage (AMTI, 1987). The impacts that were produced in this study did not reach these upper limits. The data obtained from the platform was acquired by the Power Lab analog to digital converter and processed using LabChart 7 computer software.

Figure 11. The AMTI force platform. A steel plate mounted at an angle of 13.5 degrees was used for all impacts to generate shear force.

Procedures

Anthropometric accuracy. The anthropometric accuracy of the medium-size NOCSAE headform in relation to a human female head was determined prior to simulation testing. To accomplish this, the head circumference and head mass of the NOCSAE headform (Table 7) were compared to the average head circumference and estimated average head mass of the sample of female hockey players (Table 5). The relative error (Equation 11) between NOCSAE headform and the average female head was calculated for each anthropometric measure.

$$\text{Relative Error} = \frac{|H-S|}{H} \tag{11}$$

Where:
H= human head measure
S= surrogate headform measure

Drop simulations. The drop simulations were performed in accordance to the NOCSAE drop test standards protocol. According to these standards, the headform, equipped with properly fitted headgear, is to be positioned in the drop carriage and dropped from a desired height to reach desired freefall velocity (NOCSAE, 2017). The NOCSAE protocol was chosen for the current study because this protocol is designed to provide reliable and repeatable measurements of linear acceleration experienced by a surrogate headform (NOCSAE, 2017). A CCM V08 helmet was mounted on the headform for all simulated head impacts and the helmet was replaced with an identical new helmet after every 90 impacts. At impact, the instantaneous linear acceleration of the head was measured upon impact via the triaxial accelerometers situated in the headform and the resultant acceleration (Equation 10) was used to calculate GSI (Equation 9), aligning with NOCSAE standards (NOCSAE, 2017). Shear force at impact was also measured via the AMTI force platform by computing resultant force using Equation 12.

$$\text{Shear Force} = \sqrt{Fy^2 + Fx^2} \quad (12)$$

Where:

F*y*= Force in the antero-posterior direction (N)
F*x*= Force in the mediolateral direction (N)

The headform was dropped from 16 different speeds, similar to a head drop protocol used in previous research (Carlson, 2016), which resulted in 16 freefall velocities as listed in Table 9. One trial of every combination of the three neck stiffnesses (weak, average, strong), three impact locations (front, rear, and side), and two impact mechanisms (direct and whiplash+impact) was tested at the 16 speeds, resulting in a total of 288 impacts. The following sections provide a detailed description of how each factor was manipulated in the drop simulations.

Table 9

Simulee Inbound Velocities and Drop Heights

Simulee Number	Drop Height (m)	Impact Velocity (m/s)
1	0.35	2.62
2	0.40	2.80
3	0.45	2.97
4	0.50	3.13
5	0.55	3.28
6	0.60	3.43
7	0.65	3.57
8	0.70	3.71
9	0.75	3.84
10	0.80	3.96
11	0.85	4.08
12	0.90	4.20
13	0.95	4.32
14	1.00	4.43
15	1.05	4.54
16	1.10	4.64

Adapted and modified from "The influence of neck stiffness, impact location, and angle on peak linear acceleration, shear force, and energy loading measures of hockey helmet impacts," by S. Carlson, 2016, Master's Thesis, Lakehead University, Ontario, Canada.

Neckform stiffness. To adjust the torque of the neckform, the headform was situated in the device illustrated in Figure 12 and the circular endplate was removed to access the bolt of the longitudinal cable. The stiffness of the neck was manipulated by adjusting the torque of the cable running longitudinally through the neckform using a torque wrench. As illustrated in Figure 12, the correct torque was achieved by adjusting the torque wrench until the appropriate force was applied to a Chatillon gauge, based on Equation 13. The three neckform torques represented the 10th, 50th, and 90th percentile of converted overall cervical muscle strength data from Part I of the study, listed in Table 6.

$$T = F \times d \tag{13}$$

Where :

T= Torque (Nm)
F= Force (N)
d= Perpendicular distance of force from axis of rotation (m)

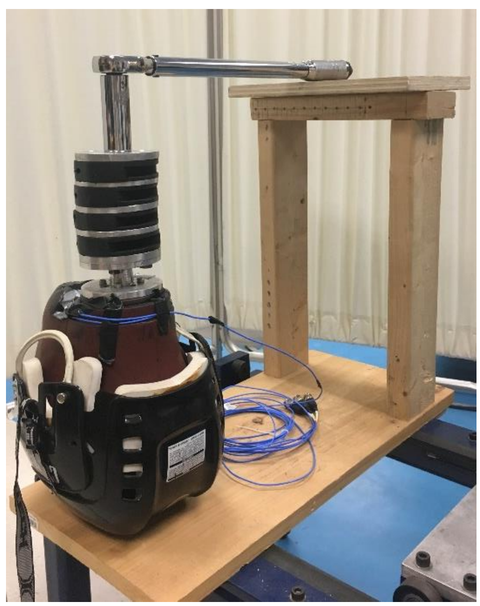

Figure 12. Adjusting the torque of the neckform to manipulate the neck stiffness. The desired torque was achieved when the appropriate force was applied to the torque wrench, based on Equation 10.

Head impact location. Impact location was adjusted by manipulating the position of the headform in the drop carriage. The three head impact locations that were tested included the front, rear, and side of the head. Only three impact locations were tested, as compared to the standard five impact locations identified by NOCSAE (2017), because the front, rear, and side impact locations relate to the flexion, extension, and side flexion strength measures, respectively, collected during the cervical muscle strength testing in Part I of the study. Front boss and rear boss locations were not tested because cervical muscle strength was not measured in the oblique plane in Part I of the study. According to Higgins, Halstead, Snyder-Mackler, and Barlow (2007), when referring to the NOCSAE headform, the front location is situated “in the median plane approximately one inch above the anterior intersection of the median and reference plane” (p. 7); the rear location is situated “approximately at the intersection of the median and reference planes” (p. 7); and the side location is situated “approximately at the intersection of the reference and coronal planes on the right side of the headform” (p. 7). Only the left side of the headform was impacted during testing.

Impact mechanism. The two impact mechanisms included direct head impact and whiplash with subsequent head impact (hereafter referred to as “whiplash + impact”). Direct impact was achieved by allowing the drop carriage to freefall and impact the force platform below, without interruption. The whiplash + impact mechanism was simulated by abruptly halting the freefalling drop carriage prior impact, which caused the head to undergo a rapid deceleration-acceleration whiplash motion and subsequently impact the force platform. The drop carriage was halted when it impacted a base of hockey pucks at a height of 0.36 metres (Figure 13*).* Hockey pucks were used because they did not deform under the force of the freefalling drop carriage and they helped dissipate some of the energy from the system during the abrupt halt, to protect the equipment.

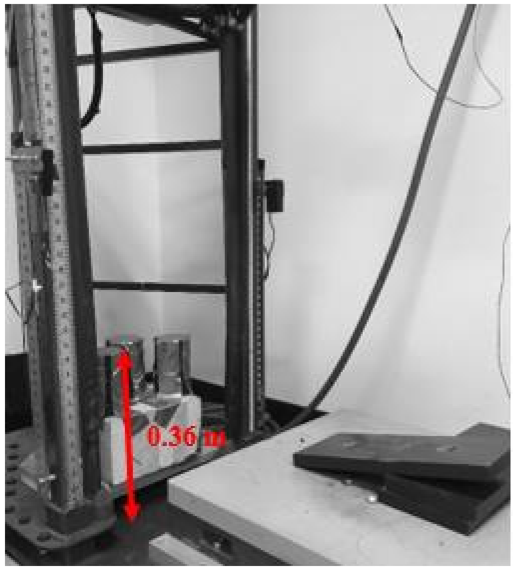

Figure 13. Equipment setup for the whiplash+impact mechanism. The hockey pucks will halt the freefalling drop carriage prior to the headform impacting the force platform, which will create a whiplash effect with a terminal head impact.

Data Analysis

To examine the accuracy of the NOCSAE headform, relative error was used to compare the head circumference and head mass of the medium-size NOCSAE headform to a sample of female hockey players. The surrogate dimensions used were those reported by NOCSAE (2017), while the anthropometric measures of the female hockey players were measured in Part I. A 5% margin of error was considered an acceptable measure of accuracy since the data was analyzed with a 95% confidence interval.

To examine the effect of neckform torque, head impact location, and impact mechanism on peak linear acceleration, peak shear force, and Gadd Severity Index, inferential statistics were used, and each dependent variable was evaluated. Three separate $3_{\text{(neckform torque)}}$ x $3_{\text{(head impact location)}}$ x $2_{\text{(impact mechanism)}}$ completely randomized factorial ANOVAs were conducted to examine the influence of these three factors on peak linear acceleration, shear force, and GSI at an alpha level of $p \leq .05$. If there were no significant three-way interaction effects observed, the significant main effects of the independent factors were analyzed. Next, a two-way ANOVA was conducted to further analyze any significant three-way interaction effects among independent factors. Finally, a series of one-way ANOVAs or independent-samples *t*-tests were conducted to explain the simple main effects observed within the levels of each independent factor. Since neckform torque and impact location were defined by three levels, a Bonferroni post-hoc analysis was used to determine the location of the significant differences within each variable.

Results

Research Question 1: Anthropometric Accuracy

The first research question aimed to determine the representativeness of the medium-sized NOCSAE headform to an average female head. The relative error in head circumference between the surrogate (57.6 cm) and the average female hockey player (M=56.1 cm, SD=2.0 cm) was 2.86%, while the relative error between the headform parameters (M=4.90 kg) and the estimated mass of the average female head (M=4.79 kg, SD=0.73 kg) and was 2.33%. Each of these anthropometric measures, therefore, produced an acceptable margin of error between the NOCSAE headform and the human, since the relative error was <5.0% for both measures.

Research Question 2: Peak Linear Acceleration

The second research question examined the effect of neckform torque, head impact location, and impact mechanism on measures of peak linear acceleration during head impact simulation testing. The descriptive results are summarized in Table G1 in Appendix G. The summary table includes measures of peak linear acceleration, peak shear force, and GSI, and are expressed as mean values and standard deviations (shown in parentheses) for each dependent variable.

A three-way ANOVA revealed no significant three-way interaction effect among neckform torque, head impact location, and impact mechanism on measures of peak linear acceleration, $F(4, 270)=.50$, $p>.05$. There was, however, a statistically significant main effect of impact mechanism on peak linear acceleration with a small effect size, $F(1, 270)=55.60$, $p<.05$, $\eta^2=.17$. Illustrated in Figure 14, the significant main effect of impact mechanism on peak linear acceleration revealed that the whiplash+impact mechanism (M=143.86 g, SD=29.42 g) generated significantly greater peak linear acceleration than direct impacts (M=115.98 g, SD=35.38 g).

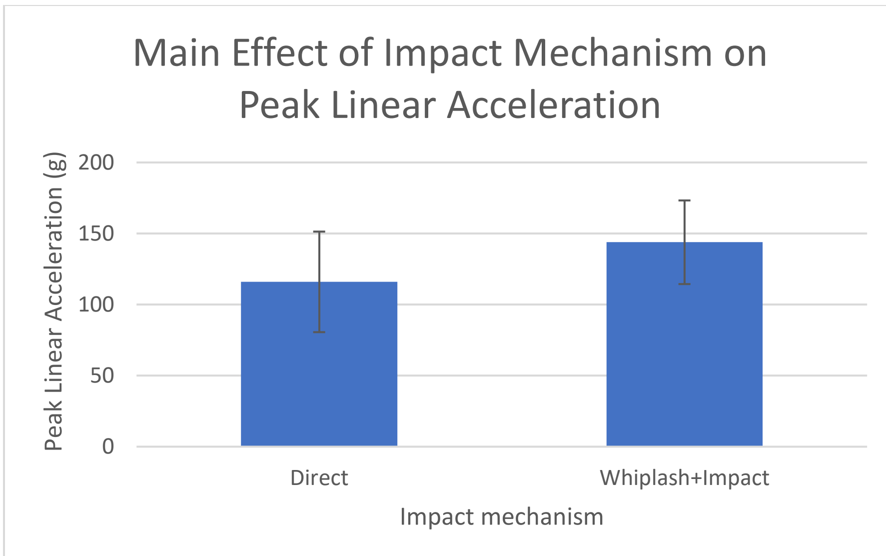

Figure 14. Main effect of impact mechanism on peak linear acceleration. The whiplash+impact mechanism generated significantly greater levels of peak linear acceleration than direct impacts.

Results also revealed a significant two-way interaction effect between impact location and impact mechanism with a small effect size, $F(2, 270)=10.40$, $p<.05$, $\eta^2= .07$. Figure 15 illustrates the interaction effect and the general increase in peak linear acceleration during the whiplash+impact mechanism for all impact locations.

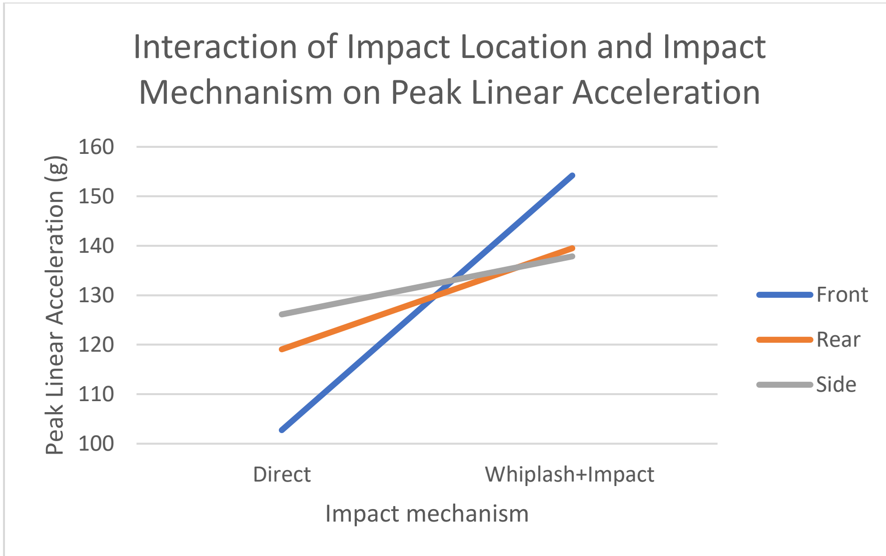

Figure 15. Two-way interaction of impact mechanism and impact location on peak linear acceleration. Linear acceleration was greater during the whiplash+impact mechanism for all impact locations.

To further investigate the two-way interaction effect between impact location and impact mechanism, separate one-way ANOVAs were used to analyze the simple main effects of impact location on measures of peak linear acceleration for each impact mechanism. Significant differences in peak linear acceleration among impact locations for direct impacts were found, $F(2, 141)=5.90$, $p<.05$, $\eta^2=.08$. The mean peak linear acceleration for each impact location for the direct impact mechanism were as follows: front (M=102.74 g, SD=39.24 g); rear (M=119.07 g, SD=28.46 g); side (M=126.13 g, SD=34.12 g). A Bonferroni post hoc comparison revealed that side impacts resulted in significantly higher levels of peak linear acceleration than frontal impacts during direct head impacts, $p<.05$ as depicted in Figure 16.

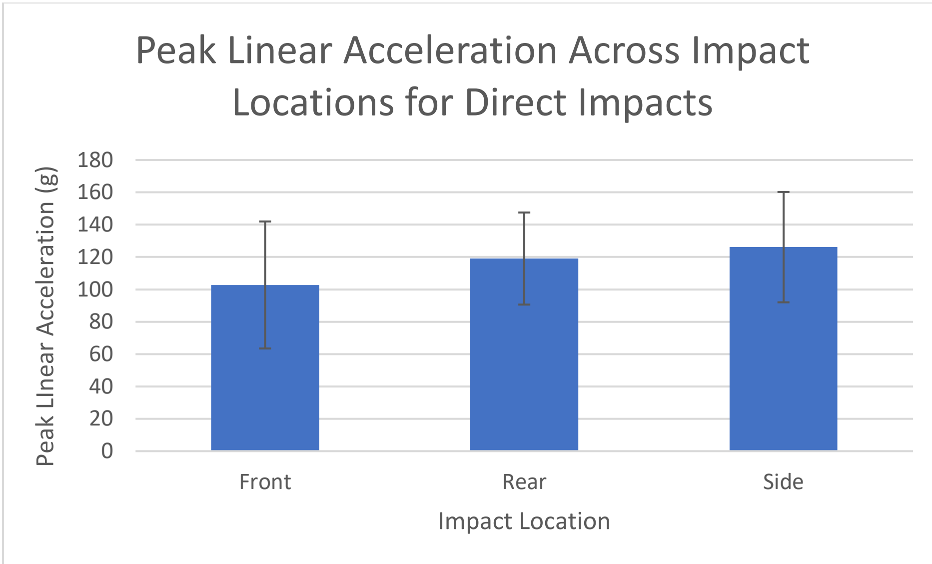

Figure 16. Peak linear acceleration across impact locations for direct head impacts. A Bonferroni post-hoc analysis revealed significantly higher measures of peak linear acceleration in side impacts as compared to frontal impacts.

Furthermore, a significant simple main effect for peak linear acceleration among impact locations for the whiplash+impact mechanism was seen, $F(2, 141)=4.73$, $p<.05$, $\eta^2=.06$. As illustrated in Figure 17, a Bonferroni post hoc analysis revealed significantly higher peak linear acceleration during frontal impacts (M=154.21 g, SD=4.85 g) compared to rear (M=139.50 g, SD=35.79 g) and side (M=137.86 g, SD=23.85 g) impacts for the whiplash+impact mechanism.

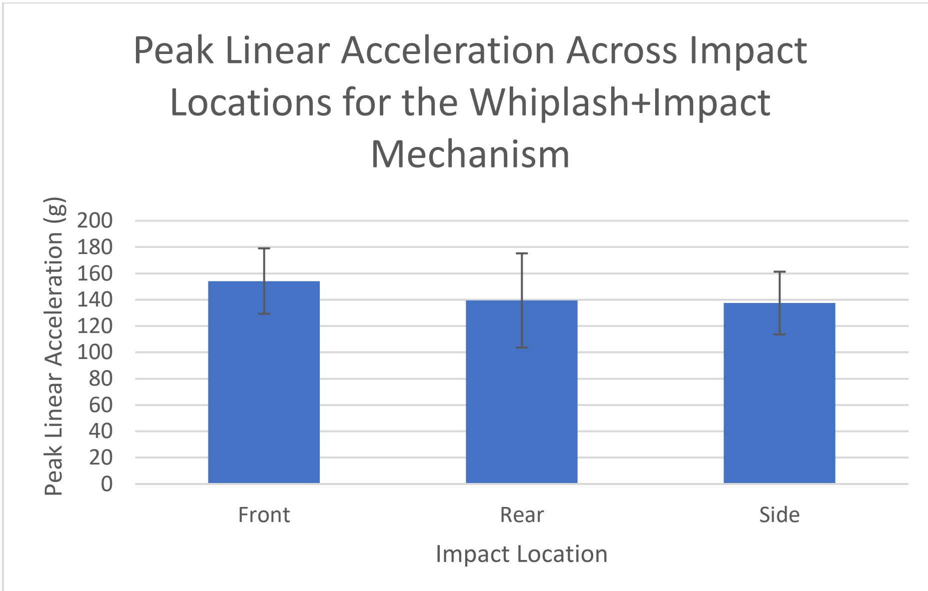

Figure 17. Peak linear acceleration across impact locations for the whiplash+impact mechanism. A Bonferroni post-hoc comparison demonstrated significantly higher measures of peak linear acceleration for frontal impacts as compared to rear and side impacts.

Next, a series of *t*-tests for independent measures were conducted to examine the simple main effects of impact mechanism on measures of peak linear acceleration for each impact location. Results for each impact location were consistent in revealing that the whiplash+impact mechanism produced greater levels of peak linear acceleration for all impact locations, as illustrated in Figure 18. Peak linear acceleration was significantly greater during the whiplash+impact mechanism (M=154.21 g, SD=24.85 g) compared to a direct impact mechanism (M=102.74 g, SD=39.24 g) for frontal impacts, $t(79.48)= -7.68$, $p<.05$, $d=1.57$. This finding was consistent for rear impacts, as the peak linear acceleration was found to be significantly greater during the whiplash+impact mechanism (M=139.50 g, SD=35.79 g) compared to a direct impact mechanism (M=119.07 g, SD=28.46 g), $t(89.46)= -3.10$, $p<.05$, $d=.63$. There was also significantly greater peak linear acceleration during the whiplash+impact mechanism (M=137.86 g, SD=23.85 g) compared to the direct mechanism (M=126.13 g, SD= 34.12 g) for side head impacts, $t(84.08)= -1.95$, $p=0.05$, $d=.40$.

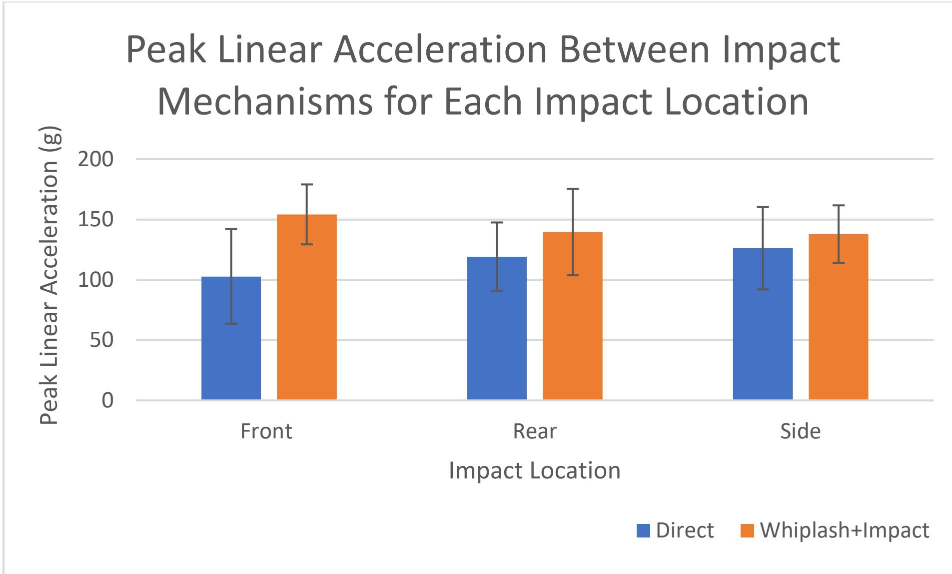

Figure 18. Peak linear acceleration between impact mechanisms for each impact location. The whiplash+impact mechanism consistently exhibited significantly higher levels of peak linear acceleration than direct head impacts.

Research Question 3: Peak Shear Force

The third research question was concerned with the effect of neckform torque, head impact location, and impact mechanism on measures of peak shear force during head impact simulation testing. First, results of the three-way ANOVA did not reveal a significant three-way interaction effect among neckform torque, impact location, and impact mechanism on measures of peak shear force, $F(4, 270)=1.16$, $p>.05$. There was, however, a significant main effect of impact mechanism, $F(1, 270)=63.49$, $p<.05$, $\eta^2=.19$, and impact location, $F(2, 270)=13.85$, $p<.05$, $\eta^2=.09$ when measuring peak shear force. A descriptive analysis of the main effect of impact mechanism revealed that there was significantly higher shear force experienced during direct head impacts (M=2060.67 N, SD=730.25 N) compared to the whiplash+impact mechanism (M=1498.96 N, SD=503.71 N). Figure 19 illustrates the mean difference in peak shear force experienced during impact testing for each of the two impact mechanisms.

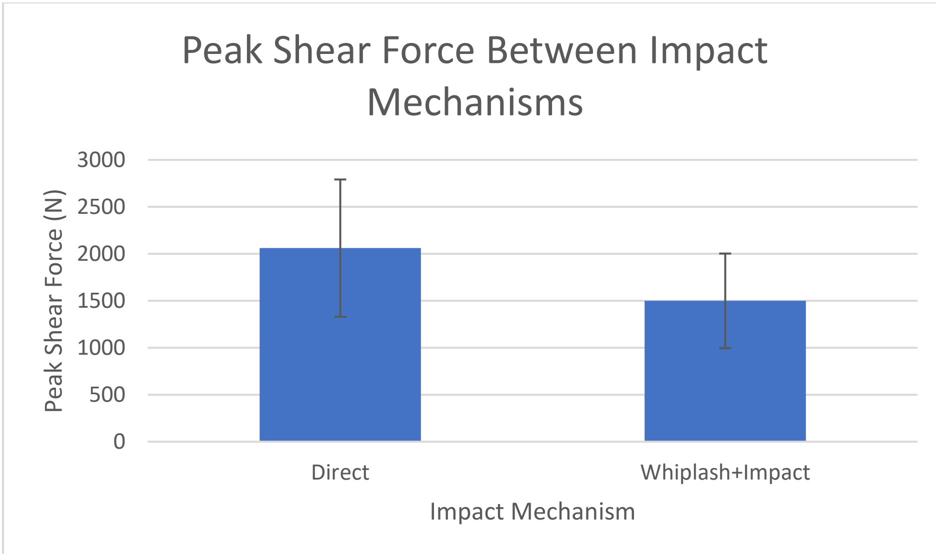

Figure 19. Measures of peak shear force for two different impact mechanisms. Direct head impacts subjected the headform to significantly higher levels of shear force than the whiplash+impact mechanism.

Furthermore, a Bonferroni post hoc analysis of the main effect of impact location on peak shear force revealed that the frontal impacts (M=1538.99 N, SD=491.09 N) experienced significantly lower measures of peak shear force than rear (M=1810.03 N, SD=684.53 N) and side impacts (M=1990.43 N, SD=780.80 N). Illustrated in Figure 20, peak shear force was the lowest in frontal impacts and the greatest in side impacts.

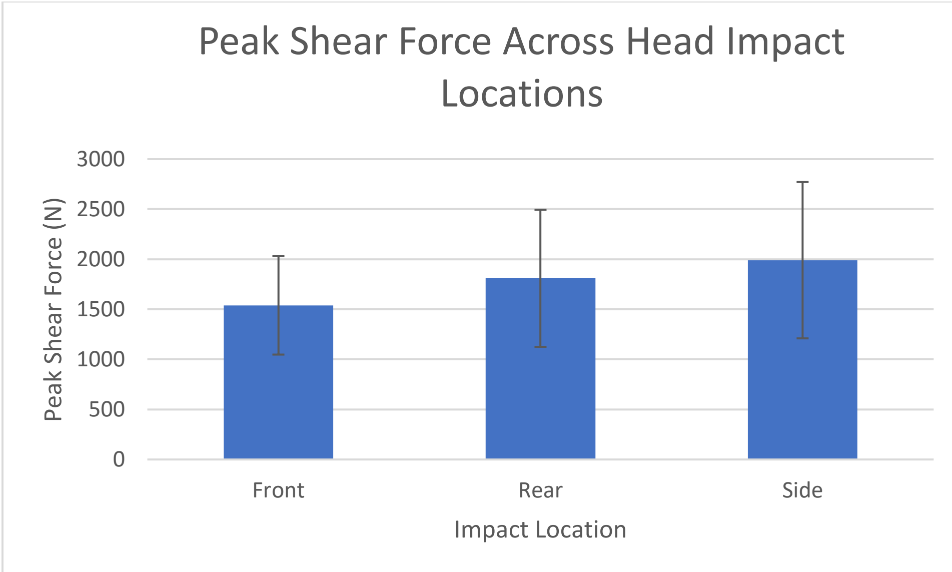

Figure 20. Measures of peak shear force across impact locations. Frontal impacts experienced significantly lower peak shear force measures than rear and side head impacts.

There was also a significant two-way interaction effect between impact location and impact mechanism on measures of peak shear force, $F(2, 270)=4.90$, $p<.05$, $\eta^2=.04$ (Figure 21).

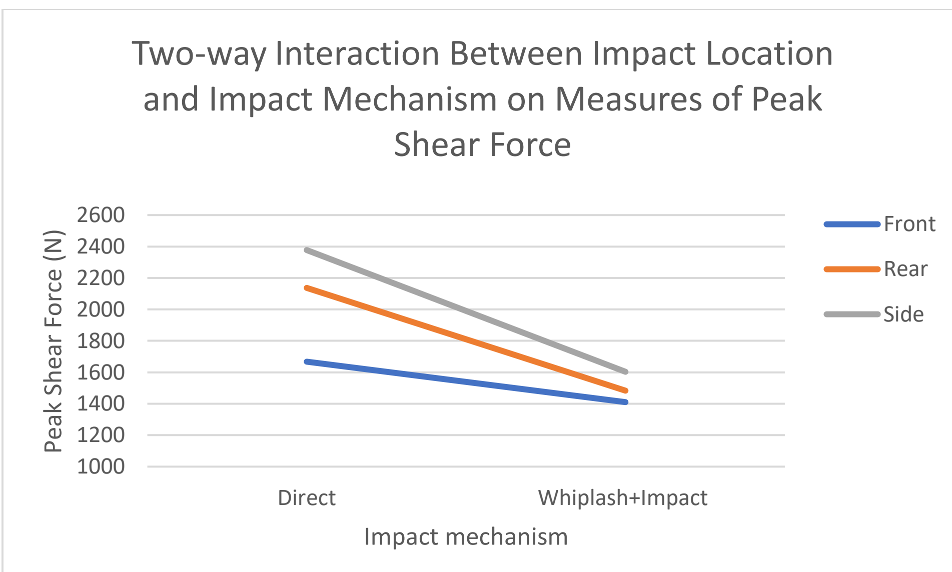

Figure 21. Two-way interaction between impact location and impact mechanism on measures of shear force.

To further analyze the significant two-way interaction, a one-way ANOVA revealed a statistically significant simple main effect on peak shear force among impact locations for direct head impacts, $F(2, 143)= 13.84$, $p<.05$, $\eta^2=.16$. A Bonferroni post hoc comparison of the simple main effect identified significantly lower measures of peak shear force during frontal impacts (M=1667.65 N, SD=565.70 N) compared to rear (M=2136.87 N, SD=677.90 N) and side impacts (M=2377.48 N, SD=759.43 N) for the direct impact impact mechanism (Figure 22). Although there were no statistically significant differences among impact locations for the whiplash+impact mechanism, $F(2, 143) = 1.82$, $p>0.05$, a similar trend was observed in measures of peak shear force for each impact location: front (M=1410.33 N, SD=365.56 N); rear (M=1483.18 N, SD=519.00 N); and side (M=1603.37 N, SD=591.23 N).

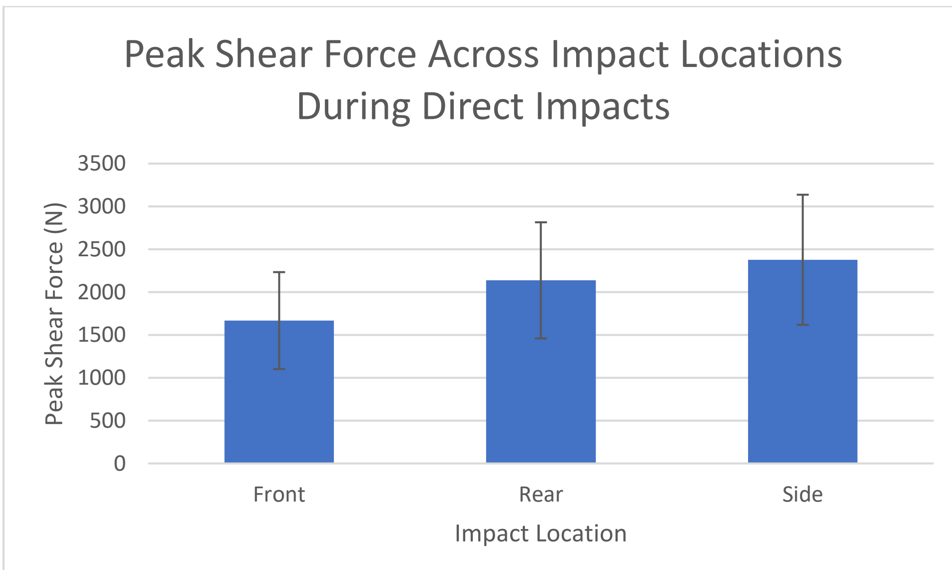

Figure 22. Measures of peak shear force for direct head impacts. A Bonferroni post-hoc comparison identified significantly lower shear force experienced during direct frontal impacts than direct rear or side impacts.

Next, a series of independent samples *t*-tests were conducted to examine the simple main effect of impact mechanism on measures of peak shear force for each impact location. Results revealed that direct head impacts resulted in significantly greater levels of peak shear force than the whiplash+impact mechanism for frontal, $t(80.42)= 2.65$, $p<.05$, $d=.54$; rear, $t(94)= 5.31$, $p<.05$, $d=1.08$; and side head impacts, $t(95)= 5.57$, $p<.05$, $d=1.14$. Figure 23***Error! Reference source*** *not found.* illustrates the higher peak shear force experienced during the direct impacts as compared to the whiplash+impact mechanism for each impact location. The mean peak shear force for each impact location and impact mechanism were as follows: frontal, direct mechanism (M=1667.65 N, SD=565.7 N); frontal, whiplash+impact (M=1410.33 N, SD=365.56 N); rear, direct (M=2136.87 N, SD=677.9 N); rear, whiplash+impact (M=1483.18 N, SD=519.0 N); side, direct (M=2377.48 N, SD=759.43 N); and side, whiplash+impact (M=1603.37 N, SD= 591.0 N).

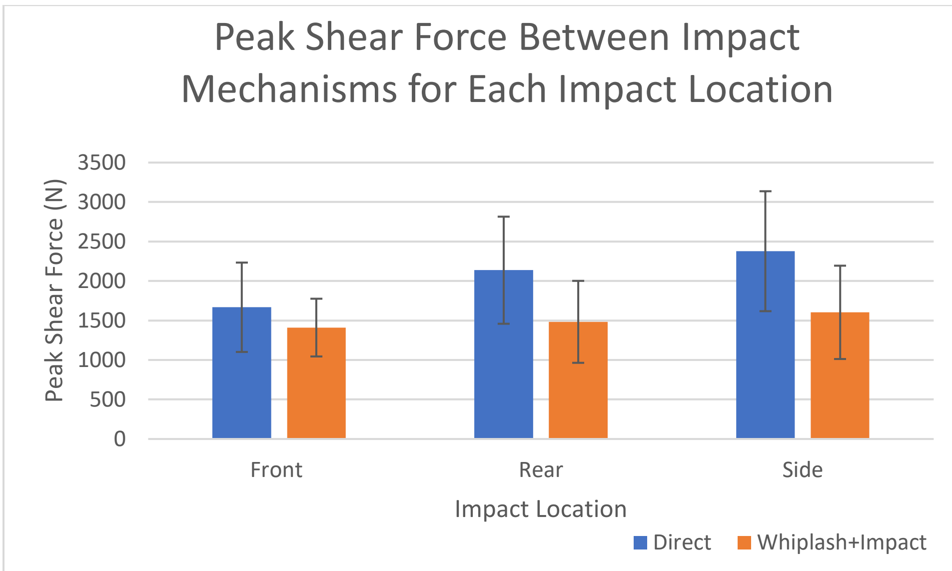

Figure 23. Measures of peak shear force during each impact location. Direct impacts consistently produced significantly higher levels of peak shear force than the whiplash+impact mechanism.

Research Question 4: Gadd Severity Index (GSI)

The fourth research question examined the effect of neckform torque, head impact location, and impact mechanism on Gadd Severity Index. A three-way ANOVA revealed no significant three-way interaction effect among neckform torque, impact location, and impact mechanism on measures of GSI, $F(4, 270)=.35$, $p>.05$. Results did, however, demonstrate a significant main effect of impact mechanism on GSI, $F(1, 270)=68.18$, $p<.05$, $\eta^2=.20$. A descriptive analysis of the significant main effect revealed, specifically, that direct head impacts produced significantly lower mean GSI (M=405.96, SD=217.76) than the whiplash+impact mechanism (M=550.44, SD=328.50), as illustrated in Figure 24. No other significant interaction or main effects of neckform torque, impact location, or impact mechanism on measures of GSI were found.

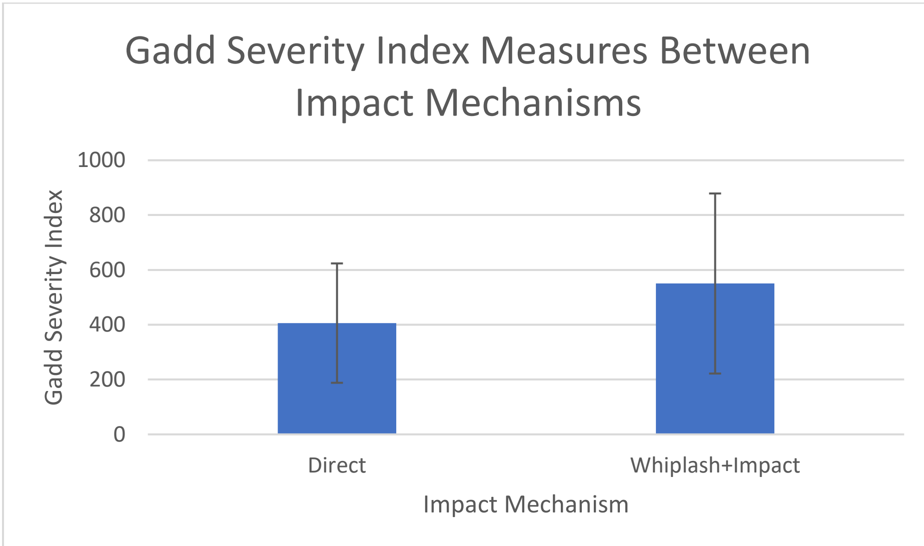

Figure 24. Measures of GSI for two different impact mechanisms. GSI was significantly greater in the whiplash+impact mechanism as compared to direct head impacts.

Discussion

The purpose of Part II of the research was to examine the effect of neckform torque, impact location, and impact mechanism on head impact biomechanics during simulated head impacts. Each of the four research questions used to guide Part II of the study were examined and discussed separately.

Anthropometric Accuracy

Simulation-based concussion testing may be limited by the ability of the simulated impacts to mimic real-life head impacts in sport, including the accuracy of the surrogate devices used to model the human head. Most surrogate headforms, including the NOCSAE headform, are designed to be sex-neutral and represent the 50th percentile of an adult head (McAlister, 2013). This poses a potential barrier when trying to accurately represent males or females specifically.

According to Vasavada et al. (2007), females have head dimensions that are approximately 3-6% smaller than males, including a significantly smaller head circumference. Therefore, to ensure that the medium-size NOCSAE headform could provide an accurate representation of an average female head, this study examined the relative error of the head circumference and head mass between the female sample and the NOCSAE headform. Since results revealed less than 5% relative error between the two sets of dimensions, the NOCSAE headform was deemed an appropriate surrogate to be used in female-specific simulated head impact testing. Nonetheless, to strengthen ecological validity, or the ability of results to be applied to specific real-life settings (George, Batterham, & Sullivan, 2003), surrogate devices that respond to sex-differences in anthropometry should be considered as research continues to innovate human models for simulation testing.

Peak Linear Acceleration

Acceleration threshold. Peak linear acceleration during an impact describes the rate of change of linear velocity over time of the head and occurs when a force is directed through the center of mass of the head (McLean & Anderson, 1997). Linear acceleration is a commonly used measure in concussion research due to its biomechanical relationship to concussions (Clark et al., 2016; Rowson & Duma, 2013). Due to the role of linear acceleration in concussion injuries, research has developed linear acceleration threshold measures to identify tolerance of the human brain to concussion injuries. Using simulated reconstructions of head impacts in football combined with finite element modelling, Zhang et al. (2004) determined that experiencing 66 g, 82 g, and 106 g of linear head acceleration corresponds a 25%, 50%, and 80% risk of sustaining a concussion, respectively. These threshold values proposed by Zhang et al. (2004) were used in the current research to quantify the likelihood of the simulated head impacts producing linear acceleration measures large enough to cause a concussion. Comparing the mean peak linear acceleration for each condition (Table G1) to these threshold measures, each condition presented an 80% probability of resulting in a concussion. This alarming finding draws attention to the potential severity of head impacts sustained during ice hockey, given the high-speed nature of the sport (Wilcox et al., 2014).

Main effect of impact mechanism. Although results did not reveal a significant three-way interaction effect among neckform torque, impact location, and impact mechanism, there was a significant main effect of impact mechanism on measures of peak linear acceleration. Descriptively, both mechanisms produced high enough peak linear acceleration to result in an 80% probability of mTBI; although, the whiplash+impact mechanism produced significantly greater peak linear acceleration than direct impacts. It is difficult to compare the results obtained to previous research because there is no known research that has examined the whiplash+impact mechanism. The results make sense, however, because the linear acceleration experienced by each mechanism separately would become compounded when the two mechanisms are combined. Initially, the inertial loading, or the initial "whiplash" component of the movement, would generate large amounts of linear acceleration, which was observed in previous research conducted by Pennock et al. (2017). Pennock et al. (2017) examined the peak linear acceleration generated during direct head impacts compared to a whiplash impact mechanism across three different neckform torques and three impact locations using similar testing protocols as the current study and identified that a whiplash mechanism produced significantly greater peak linear acceleration measures than a direct impact mechanism. If this initial acceleration generated from the inertial loading is then compounded by the linear acceleration generated during a direct impact, the resulting combined peak linear acceleration would be expected to be significantly greater than that produced by an isolated direct impact. Since direct head impacts may not be primary impact mechanism experienced by female hockey players due to the absence of intentional body checking in women's hockey, these results may be significant in highlighting a potential factor increasing the risk of concussions in the sport. The whiplash+impact mechanism resulted in significantly greater peak linear acceleration than direct head impacts, suggesting that perhaps other, more common impact mechanisms found in women's hockey, including a combined whiplash+impact mechanism, are increasing the players' risk for sustaining a concussion. More attention should be paid to impact mechanisms that result in a combined inertial loading and direct impact loading experienced by the head in an effort to help characterize the risk of these mechanisms in real life.

Two-way interaction and simple main effects. A significant two-way interaction between impact location and impact mechanism on measures of peak linear acceleration was also identified. This finding suggests that the linear acceleration was greater in side impacts than frontal impacts during the direct impact mechanism. These results are comparable to previous simulation-based concussion research (Walsh et al., 2011), as well as initial animal studies (Hodgson et al., 1983). Side head impacts have been found to result in greater levels of acceleration than other impact locations and may also experience more intercranial tissue damage, potentially leading to greater concussion risk (Zhang et al., 2001).

Although these findings do not align with results obtained by Carlson (2016), who found that the front boss impact location resulted in the highest peak linear acceleration followed by the front, side, rear, and rear boss locations, methodological differences may account for some of these differences. Specifically, the current research did not examine front boss and rear boss locations. Also, Carlson (2016) did not use a facial shield for the helmet, which was included in the current research. Full facial cages have the potential to alter the amount of force applied to the head during impact due to the altered geometry of the helmet. Lemair and Pearsall (2007) revealed that full cages significantly reduce the peak linear acceleration experienced by the head during a direct impact. This outcome is likely attributed to the ability of the cage to distribute some of the forces radially, away from the head's center of mass. The chin support in full facial shields is also speculated to reduce forces experienced by the head (Lemair & Pearsall, 2007). The full cage used in the current study, therefore, likely played an important role in reducing the peak linear acceleration in the frontal impacts.

Meanwhile, frontal impacts experienced greater peak linear acceleration than rear and side impacts during the whiplash+impact mechanism. This finding is a more difficult observation to explain and may be related to the relationship between linear and angular accelerations during head impacts. Previous research has noted that linear acceleration decreases with angled impacts, compared to neutral impacts (Rousseau & Hoshizaki, 2009; Walsh et al., 2011), likely due to the increased angular acceleration produced from the tangential forces associated with the angled surface. If the inverse relationship between linear and angular acceleration is consistent in head impacts, then presumably, linear acceleration would be greater in impacts with proportionally less angular acceleration. During the whiplash+impact mechanism, the protrusion of the cage caused the helmet to impact the force platform sooner in the frontal location than rear or side locations. Although not measured in this study, this situation would theoretically reduce the angular acceleration generated in the frontal position during the whiplash+impact mechanism. The reduced angular acceleration may have subsequently contributed to greater peak linear acceleration measures for the frontal location. This explanation, however, is speculative based on findings from previous research (Rousseau & Hoshizaki, 2009; Walsh et al., 2011) and is difficult to confirm without measures of angular acceleration in the current research.

Finally, when examining the significant differences between impact mechanisms when isolating head impact location, it was revealed that the whiplash+impact mechanism produced significantly greater peak linear acceleration than the direct impacts for all impact locations. These findings are similar to the main effect of impact mechanism on peak linear acceleration, as described above. The combination of linear acceleration produced from the initial inertial loading, combined with the linear acceleration generated during the head impact, likely results in a peak linear acceleration much greater than that produced from isolated direct impacts. Overall, the simple main effects observed within two-way interaction suggest that concussion risk may be elevated at certain impact sites or during certain impact mechanisms, depending on the conditions of the impact.

Peak Shear Force

The role of angular acceleration in concussive injuries has been well researched (Brainard et al., 2012; Mihalik et al., 2010; Walsh et al., 2011; Rowson et al., 2012; Rowson & Duma, 2013); however, limited research has examined the force that produces angular acceleration during an impact. Shear force is generated when a tangential or off-centre force is applied to the head and results in some degree of rotation of the head (Kleiven, 2013). Some amount of shear force is almost always present during any head impact, since the forces from most impacts are not transmitted directly through the centre of mass of the head. Greater shear impact forces are associated with greater angular acceleration of the head, which increases the intercranial tissue strain experienced during the impact (Meaney & Smith, 2011), thereby emphasizing the potential value in examining shear forces experienced by the head during an impact. In the current study, shear impact force was measured using a force platform, similar to research conducted by Carlson (2016). Since there are currently no injury thresholds for shear force measures in concussion research, it was not possible to comment on the severity of the peak shear force experienced during this study.

Main effects of impact location and impact mechanism. An examination of the interaction among neckform torque, impact location, and impact mechanism did not reveal any significant differences in peak shear force. There was, however, a main effect of impact location and impact mechanism on measures of peak shear force. Results revealed that peak shear force was significantly lower in frontal impact as compared to rear and side impacts. This finding may be related to the properties of the full facial shield. It has been suggested that the altered geometry of the helmet caused by full facial shields helps to decreases the forces experienced during an impact, as the face mask disperses the forces radially, over a larger surface area (Lemair & Pearsall, 2007). Benson, Rose, and Meeuwisse (2002) also found evidence to suggest that players who wore full facial shields as opposed to half shields experienced fewer concussions and less severe concussions, as measured by the time lost from competition. Comparatively, Carlson (2016), did not observe reduced shear force measures in frontal impacts consistently, when examining the effect of impact location, neckform torque, and impact angle on peak shear force measures during vertical head impact testing. It is important to recognize, however, that Carlson (2016) did not use any facial shielding during the head impact testing. Therefore, the presence of the facial in the current study, which has been speculated to be influential in mitigating shear forces for frontal impacts, may account for the variance in the reported results.

Furthermore, the main effect of impact mechanism revealed that direct impacts resulted in significantly greater measures of peak shear force than the whiplash+impact mechanism. Once again, it is difficult to compare these findings to previous research because no other research has examined a mechanism comparable to the whiplash+impact mechanism in the current research. Based on previous literature detailing the relationship between inertial loading and the generation of angular acceleration (Barth et al., 2001; Hynes & Dickey, 2006; Meaney & Smith, 2011), it was originally expected that the whiplash+impact mechanism would have resulted in greater measures of shear force due the initial inertial loading of the neck and head prior to impact. The observed results, however, made sense based on the head impact characteristics for each mechanism. When observing the head impacts for each mechanism, it was evident that the headform contacted the force platform much more forcefully and with a greater surface area during the direct impact mechanism as compared to the whiplash+impact mechanism. The more forceful impact would have subsequently generated greater shear force measures from the force platform during direct head impacts. The difference in head contact seemed to be related to the equipment setup and the properties of the neckform. During the whiplash+impact mechanism, the stiffness of the neckform limited the range of motion of the neck during the whiplash phase, thereby reducing the subsequent head contact on the force platform. In a real-life scenario of this mechanism, a human neck would likely move through a larger range of motion and result in a more forceful head impact with a surface, similar to a direct impact mechanism. Furthermore, although shear force and angular acceleration are related, the mechanisms by which they are measured in simulation-based concussion research are different. Shear force is measured by a force platform, while angular acceleration is measured via headform accelerometers. It is, therefore, important to consider the way in which each measure is collected before hypothesizing similar trends between the two measures.

Two-way interaction and simple main effects. In addition to the main effects observed for impact location and impact mechanism, results also revealed a significant two-way interaction effect between impact location and impact mechanism. The simple main effect of impact mechanism revealed that the direct impact mechanism experienced significantly greater peak shear force than the whiplash+impact mechanism for all impact locations. Similar to the rationale for the main effect of impact mechanism on measures of shear force, direct impacts likely resulted in greater shear force because of the more forceful contact with the force plate compared to the whiplash+impact mechanism. Moreover, when examining the simple main effects of impact location, it was revealed that frontal impacts resulted in significantly greater measures of peak shear force than rear and side impacts, but only for the direct impact mechanism. Although the speculated reasoning for this finding as it relates to the role of the full facial shield is described above, it is interesting to note that this finding is not statistically significant for the whiplash+impact mechanism. This, however, may potentially relate to the lower levels of shear force experienced during the whiplash+impact mechanism. If the shear force measures are lower due to the reduced contact of the headform with the force plate, then the mean difference in shear force among the impact locations may be minimal. Nonetheless, these results suggest that the peak shear force experienced at impact may be influenced by the combination of impact location and impact mechanism, rather than a single isolated factor. It is, therefore, invaluable to examine the relationship among potential risk factors in a multifaceted sport like ice hockey.

Gadd Severity Index

Various severity index measures have been developed from tolerance criteria associating linear head acceleration with time duration of impact to help identify risk of severe brain injury. Since impact measures specific to concussions are lacking in the literature, the established severity index measures, including GSI, are commonly used in the assessment of mTBIs (Greenwald et al., 2008). GSI, defined by Equation 6, represents the risk of severe brain injury based on the linear acceleration of the head over the duration of the impact. Traditionally, a severity index exceeding 1000 indicates a severe brain injury, likely causing endangerment to life (Gadd, 1996). Recognizing that concussions occur at a lower threshold that has yet to be established is important in concussion research. The results of the current research did not reveal any significant interaction effects among neckform torque, impact location, and impact mechanisms for measures of GSI. There was, however, a significant main effect of impact mechanism on GSI. Specifically, the whiplash+impact mechanism resulted in significantly greater measures of severity index than the direct impact mechanism. This finding was to be expected due to the significantly higher peak linear accelerations observed during the whiplash+impact mechanism. Moreover, the impact duration was also greater during the whiplash+impact mechanism as compared to the direct impacts. Since calculations of GSI are based on both peak linear acceleration and impact duration, it was logical to observe significantly greater GSI measures during the whiplash+impact mechanism. Additionally, there were multiple trials during which the GSI exceeded the injury threshold for severe brain injury, all of which occurred during the whiplash+impact mechanism. This finding only reinforces the need to explore impact mechanisms other than direct head impacts, due to the potential injury risk that they pose to athletes.

Practical Implications of Part II Results

Results obtained from simulation-based concussion research are valuable, especially when contextualized with the sport to which they apply. The practical implications of the effect of each of the independent variables on measures of peak linear acceleration, peak shear force, and GSI may provide greater insight into concussion risk factors involved in women's hockey. A significant trend identified in the research was the influence of impact mechanism on all outcome measures. The whiplash+impact mechanism produced greater measures of peak linear acceleration and GSI than the direct impact mechanism. Although, most existing research has only examined direct head impacts, failing to explore alternative mechanisms that have the potential to produce concussive injuries, such as the whiplash+impact mechanism. Due to the physical nature and enclosed setting of the ice rink, there is the potential for players to impact the boards, ice, or another player before their head makes contact with the surface, making this impact mechanism realistic in the sport of ice hockey (Wilcox et al., 2014). It is also known that significant acceleration/deceleration forces can have negative effects on brain tissue even without direct and visible head impact (Barth et al., 2001). It was noted that measures of peak shear force did not follow a similar trend as the peak linear acceleration and GSI, although, this finding may have been influenced by equipment setup, as previously described. Based on the potential practical influence of these findings, it is, therefore, important for concussion research to extend beyond direct head impacts to become more aware of the influence of varying mechanisms on the accelerations and forces experienced by the head.

Another key finding related to women's ice hockey was the effect of impact location measures of peak linear acceleration and peak shear force. Specifically, observing the increased peak linear acceleration for side impacts, as well as the mitigation of peak linear acceleration and peak shear force during frontal impacts may have strong practical implications. First, impact location has been identified by previous research as a potential risk factor for concussions, particularly identifying side impacts as a high-risk location (Hodgson et al., 1983; Walsh et al., 2011; Zhang et al., 2001). This finding was consistent in the current simulation research, but also aligns well with findings from Brainard et al. (2012). In research examining the frequency, magnitude, and location of head impacts sustained by male and female collegiate hockey players, Brainard et al. observed that the frequency of impacts to the right and left sides of the head as significantly greater in females compared to males. If female hockey players are being impacted more frequently in a high-risk area, then the likelihood of sustaining a concussion from the impacts may be increased.

Furthermore, the moderation of peak linear acceleration during direct impacts and peak shear force during both impact mechanisms for frontal impacts raises the question of the potential protective role of a full facial shield. Full facial shields have been believed to increase the angular acceleration of the head during impact due to the increased radius of the helmet, thereby increasing concussion risk (Graham et al., 2014). Since angular acceleration was not measured in the current research, it is not possible to comment on this relationship. The results of the current study, however, suggest that facial shielding may play a role in reducing other measures, including the linear acceleration and shear force. Female hockey players of all ages are mandated to wear full facial shields (Hockey Canada, 2003). Therefore, with more research and the proper design, facial shields may have the potential of helping to reduce the injury risk associated with certain types of head impacts in the sport.

Despite some of the practical implications maintained by the current research, it was interesting that there was no significant effect of neckform torque on any of the outcome measures. This finding was consistent with research conducted by Carlson (2016), who did not observe any significant interaction or main effects of neckform torque on measures of peak linear acceleration and GSI. Carlson did, however, observe a significant three-way interaction effect of neckform torque, impact location, and impact angle on peak shear force. Furthermore, Jeffries (2017) revealed a three-way interaction effect among neckform torque, impact location, and facial shielding conditions on measures of peak linear acceleration and energy loading during simulated head impact testing. Additionally, Rousseau and Hoshizaki (2009) analyzed the influence of neck compliance on measures of peak linear acceleration and revealed a statistically significant effect of neck compliance on peak linear acceleration at impact speeds of 5 m/s and 9 m/s. Specifically, the results revealed that the peak linear acceleration was significantly greater during the "stiff" neck condition compared to the "soft" neck condition for 5 m/s impacts, and significantly greater in the "stiff" condition compared to the "soft" and "median" conditions for 9m/s impacts (Rousseau & Hoshizaki, 2009). These results contradicted the hypothesis that a stiffer, less compliant neckform would mitigate linear acceleration experienced by the head during impacts.

Based on the results of simulation-based research that found significant effects of neck compliance on peak linear acceleration (Carlson, 2016; Jeffries, 2017; Rousseau & Hoshizaki, 2009), as well as the popular theory that cervical muscle strength may play a significant role in concussion risk (Eckner et al., 2014; Mihalik et al., 2011; Schmidt et al., 2014), the current research expected to observe a significant effect of neckform torque on the outcome measures. Although not statistically significant, the results of the current study followed a similar pattern to the findings by Rousseau and Hoshizaki (2009), in which peak linear acceleration increased with greater neckform torques (see Table G1 for a summary of descriptive statistics). The reason underlying this trend remains unclear; however, it may be related to the material properties of the mechanical neckform and its response to dynamic loading. It is also important to recognize that previous simulation-based research that observed significant effects of neckform torque on peak linear acceleration conducted horizontal head impacts (Jeffries, 2017; Rousseau & Hoshizaki, 2009), rather than vertical head impacts, as executed by Carlson (2016) and the current study. This observation may indicate that perhaps the protective ability of neck stiffness is different in types of impact mechanisms. Specifically applied to the sport of ice hockey, perhaps the role of cervical muscle strength may be more influential during horizontal impacts, such as player-on-player collision, as opposed to a fall to the ice.

Furthermore, limitations in the material properties and design of the mechanical neckform may have altered the dynamic response of the neckform during impacts, thereby influencing its behavior in response to loading. The Hybrid III neckform is the more commonly used in concussion research (Allison, Kang, Bolte, Maltese, & Arbogast, 2014; Beckwith et al., 2012; Kendall et al., 2012; Walsh et al., 2014), while the mechanical neckform used in the current research was custom-designed to allow for the adjustment of neckform torque. Although the design of the mechanical neckform was based on the Hybrid III neckform, the structure of the two neck models are slightly different. There are four vertebrae in the Hybrid III neckform that are offset towards the front of the neck and are notched to mimic the response of flexion and extension (Ashrafiuon, Colbert, Obergefell, & Kaleps, 1996). Comparatively, the vertebrae in the custom designed mechanical neckform feature a large posterior cutout and a small anterior cutout and are not uniformly notched as with the Hybrid III neckform. These structural characteristics could notably alter the dynamic response of the neckform during loading and unloading. It is also recognized that the mechanical neckform may not exhibit the same response to an impact as would a human neck, which should be considered when discussing the practical significance of the current results. Nonetheless, by using human cervical muscle strength data to adjust the stiffness of the neckform during simulation testing, the applicability of the simulation results was increased with the incorporation of data that was representative of the target population.

Chapter 5- Discussion and Conclusion

The Integration of Human Data and Simulation

Concussion research in athletes typically assumes one of two forms: on-field assessment using real-time data or simulated reconstruction of head impacts. Each of these methods provide useful information to researchers regarding the nature of concussions in sport. There are, however, limitations inherent in both research methods that may affect the ecological validity of the results. Particularly, there has been speculation as to how well simulation-based research can recreate head impacts experienced in sport, including the accuracy of the response of the surrogate devices as compared to a human. To help reduce some of the potential limitations inherent in simulation-based research, this study aimed to combine real-life data to head impact simulations by modelling the cervical muscle strength collected from a sample of female hockey players on the mechanical neckform during impacts. By using human data to adjust the stiffness of the mechanical neckform, the results of the study, specifically pertaining to the influence of neckform torque, could be more confidently generalized back to female ice hockey players.

Few studies have applied human data to simulation-based concussion research. Pellman et al. (2003) used video surveillance of NFL games to recreate the impact velocity, direction, and head kinematics during simulation testing. Similarly, Beckwith et al. (2011) used information extracted from video recordings to replicate the average impact velocity and head impact location for football players to compare the head kinematics recorded by the HIT system and those obtained from the Hybrid III headform. However, it doesn't appear as though any other previous study has modelled any specific human characteristics, such as cervical muscle strength, on the head or neckform during simulation testing.

The surrogate headforms and the neckforms utilized during head impact simulation testing typically represent the 50th percentile of human adult anthropometrics (MacAlister, 2013). Since there are notable differences in head and neck anthropometry between males and females (Vasavada et al., 2008), it is unclear as to how accurately the neutral surrogate devices represent female body segment parameters when used in sex-specific research. To explore the potential differences in the current research, the mean head mass and head circumference of female head were compared to dimensions of the medium-sized NOCSAE headform, which was designed to represent the 50th percentile adult head (MacAlister, 2013). The results revealed that the relative error of the mean head mass (kg) and the mean head circumference (cm) between study sample and the NOCSAE headform were 2.33% and 2.86%, respectively. These measures of relative error are within a 5% margin, suggesting that the headform used in simulation testing was an acceptable representation of a sample female hockey players.

Unlike the NOCSAE headform, which is designed with accurate facial features and bone structure (MacAlister, 2013), the mechanical neckform used in this study was not designed to be anatomically accurate. Since it was known that the mechanical neckform was not an accurate anthropometric representation of a female human neck, the process of modelling human neck strengths on the mechanical neckform was used to increase the ecological validity of the results collected. Since there is limited existing research combining real-life data with simulation-based techniques when examining concussion, the current research helps to fill this gap in literature by strategically combining the two research methods to gain a better understanding of the influence of cervical muscle strength, impact location, and impact mechanism on measures of peak linear acceleration, shear force, and GSI in female hockey players.

Conclusions

This study included two separate, but related components. Part I of the study, discussed in Chapter 3, developed normative data for cervical muscle strength and anthropometric measures in a sample of female hockey players. The overall cervical muscle strength data was then transformed to torque measures using a pre-established calibration equation and conversion procedure, to allow the cervical muscle strength measures to be appropriately modelled in Part II of the study, which is discussed in Chapter 4. The purpose of Part II was to examine the effect of neckform torque, impact location, and impact mechanism on head impact biomechanics using head impact simulation testing. The neckform torques established from the data in Part I were modelled on the mechanical neckform in Part II to provide an accurate representation of the cervical muscle strength in female hockey players during the simulation testing. This project aimed to combine human data with simulation-based concussion research to increase the practical application of the results obtained from simulation testing and make the outcomes more ecologically valid.

The overall cervical muscle strength data in Part I exhibited a normal distribution curve from which the 10^{th}, 50^{th}, and 90^{th} percentile strength measures were calculated as 58.64 N, 76.01 N, and 108.27 N, respectively. These cervical muscle strength values were scaled and converted to torque measures using the z-score technique, resulting in torque values of 1.36 Nm, 2.94 Nm, and 4.62 Nm to represent the 10^{th}, 50^{th}, and 90^{th} percentile of neckform stiffness, respectively. In addition to being used in Part II, the normative data collected for female hockey players can be used in future research examining head injuries in the sport. Since limited normative data for female hockey players currently exists, the measures collected from this research can also act as a foundation upon which future descriptive research can build.

Furthermore, results of Part II revealed a statistically significant main effect of impact mechanism on measures of peak linear acceleration, peak shear force, and GSI, as well as a significant main effect of impact location on peak shear force. There was also a statistically significant two-way interaction of impact location and impact mechanism on peak linear acceleration and peak shear force. Finally, there was a significant simple main effect of impact location on measures of GSI. These results documenting the effect of three potential risk factors on head impact biomechanics provide some information on the influence of these factors on concussion risk in women's hockey.

Although results of this study cannot confidently identify risk factors affecting the concussion risk in female hockey players, some of the trends revealed in the data help to identify factors that may contribute to the risk of sustaining a concussion. Practically, results of this study indicate the impact mechanisms other than direct head impacts have the potential to cause concussions, which emphasizes the need for future research to investigate alternative mechanisms further. Additionally, impact location appeared to be influential in injury risk. Coaching staff and players should be aware of the potential risk associated with impacts to sides of the head, as well as the potential protective factor that full facial shields, specifically cages, have in the mitigation of concussive forces. Finally, neckform torque did not appear to influence concussion risk, although players should not rule out the possible benefit of having strong neck muscles to help oppose external forces during impacts. The results of this study provide a foundation up which further research can expand to explore potential risk factors associated with concussions in female hockey players.

Limitations

There were a few recognized limitations in Part I of this study involving the measurement of maximal isometric cervical muscle strength in a sample of female hockey players. First, the sample of female hockey players were considered to be at a "competitive" level, playing either Collegiate hockey or having a history of playing Rep hockey. The sample was, therefore, homogeneous in nature. Since athletes of all playing calibre are at risk for sustaining concussions, it is unclear whether significant differences in cervical muscle strength would have been observed in a sample including a combination of experienced and less-experienced players. Furthermore, a methodological limitation was identified during the cervical muscle strength testing with the equipment design and positioning of the participants. Participants' torsos or shoulders were not restrained during the cervical muscle strength testing, which may have allowed for accessory movements of the upper body to contribute to the force output achieved during the maximal isometric efforts. The design of the Nautilus neck strength machine did not allow for the restraint of participants' upper body, although the cervical muscle strength measures are assumed to represent only the force generated by the cervical musculature. Finally, this study has a relatively low sample size of 25 participants, so the normative data created is based off a fairly small sample of the population.

Despite the effort to ensure that the simulation conditions were as characteristic of real-life head impact scenarios for female hockey players, there were limitations to Part II of this study as well. It has been recognized that the results of this study are specific to the testing conditions, meaning that results may vary for different neckform torque, impact locations, or impact mechanisms. There are also some design limitations in the equipment used during the simulations. Although the mechanical neckform was designed to emulate the dynamic loading and unloading response of a human neck during impact, the mechanical neckform did not exhibit behaviour that would be identical to a human neck, limiting the direct generalizability of results. Similarly, the torque of the neckform could not be adjusted to a high level of accuracy due to the limited sensitivity of the load cell used to torque the neckform. Since load cell only measured to the near whole Newton, more accurate measures of neckform torque could not be achieved. The torque of the neckform was also adjusted regularly throughout testing, it is also possible that the torque was altered slightly after several impacts, making the neckform more or less compliant.

Another limitation in equipment design was the inability to measure angular acceleration of the head due to the configuration of accelerometers within the headform. Although shear force, as measured in this study, may provide invaluable information to the understanding of concussion injuries in sport, the additional measure of angular acceleration would compliment other outcome measures. Due to the strong link between angular acceleration and concussion (Kleiven, 2013; Rowson et al., 2012) and the relationship between angular acceleration and shear force (Hall, 2006), using a headform with the ability to measure angular acceleration would further strengthen the data gathered in the simulation testing.

Finally, the transformation of data from strength to torque measures may have been limited by the scale variance between the human neck and the mechanical neckform. Since the force values obtained during the calibration of the neckform were much greater than the force values measured during the maximal isometric neck strength testing, the human data was scaled up to be modelled on the mechanical neckform. Although the scaling and data transformation was performed appropriately, it would have been ideal for the data to be directly transferrable to ensure that the neckform torques used were the best representation of the human cervical muscle strength measures. This limitation in the mechanical neckform reinforces the need for female-specific surrogate devices that accurately represent human characteristics. Nonetheless, the scaling procedure allowed for the transformation of data from the human to the mechanical neckform, which was a key component to this research.

Future Directions

Future research should continue to incorporate human data in simulation-based concussion research to gain a better understanding of the behaviour of the head and neck during impacts as it applies to specific target populations. Since there is empirical evidence to suggest the potential link between cervical muscle strength and concussions (Collins et al., 2014; Eckner et al., 2014), future research should use a larger sample size to develop a more robust set of normative data for the maximal isometric cervical muscle strength of female hockey players. Having this data available will expand the ability to examine the relationship between cervical muscle strength and concussion risk in female hockey players. Furthermore, with the availability of 3D finite element modelling and advanced 3D printing software, researchers should move forward in designing a surrogate neckform with anthropometry that is representative of a female human neck. This model would be useful during simulation-based concussion research to provide a more accurate representation of the response of the human head and neck during an impact. In addition to peak linear acceleration, peak angular acceleration should also be measured during testing, since a combination of both measures is more valuable than either one on its own (Rowson & Duma, 2013). Moreover, future research should continue to explore the various impact mechanisms that are common in women's hockey to develop a better understanding of how to replicate the head impacts through simulation testing. Finally, researchers may also consider conducting similar research using a horizontal impactor to represent horizontal impacts experienced in hockey, such as player-on-player collisions, which is the most common source of injury in the sport (Agel et al., 2007; Schick & Meeuwisse, 2003).

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Appendix A:

Informed Consent Form

Informed Consent Form

I ________________________________ agree to participate in the study titled "Measuring the Isometric Cervical Strength of Female Ice Hockey Players", which will measure the neck strength of female hockey players in flexion, extension, and side flexion.

I have read and understand the terms and conditions of this research study as outlined in the information letter. I willingly agree to participate in this study.

I understand the potential risks and benefits of the study. I also understand that I have certain rights as a participant in this study. I understand that as a volunteer, I may withdraw at any time and may refuse to answer any questions or perform any activities.

I understand that personal information used in the study will remain anonymous and confidential, as it will only be used by the researchers conducting the study. I understand that I will be protected and remain anonymous in any presentation of the research findings. I am also aware that the data recorded in this study will be securely stored at Lakehead University for five years with Dr. Derek Kivi. I have been informed that the results from this study will be made available to me via email once the study has been completed. I also understand that the data may be published on journals or presented publicly; although no individual results will be made available.

I understand that the neck strength data from this study will be used in a second part of the study using simulation technology to examine factors affecting concussion risk in women's hockey. I agree to allow my data to be used for the second part of the study and I understand that I will not be needed for any further data collection.

Please indicate if you would like a copy of the results via email at the completion of the study.

☐ Yes Email: ______________________________________

☐ No

____________________________________ ____________________

Signature of Participant Date

____________________________________ ____________________

Signature of Student Researcher Date

____________________________________ ____________________

Signature of Research Supervisor Date

Appendix B:

ParQ Form

Physical Activity Readiness
Questionnaire - PAR-Q
(revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ______________________________

SIGNATURE ______________________________ DATE ______________________________

SIGNATURE OF PARENT ______________________________ WITNESS ______________________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

CSEP | SCPE

Appendix C:

Pre-Screening Questionnaire

Pre-Screening Questionnaire

Participant Name:________________________________

Age:________

Years of Playing Experience:_______

History of Neck Disorders

Question	Y/N? If Yes, Explain
Do you currently suffer from persistent neck pain?	
Are you currently taking any medications for neck pain?	
Have you ever had neck surgery?	
Have you ever had a neck or spinal injury that prevented you from playing your sport?	
If you answered yes to the above question, how long ago? Did you receive medical clearance to return to your sport?	

History of Concussions

Question	Y/N? If Yes, Explain
Have you sustained a diagnosed concussion in the past? How many?	
If so, when did your most recent concussion occur?	
How long did it take you to recover from your most recent concussion?	
Describe the mechanism of injury of your most recent concussion (e.g. direct hit to the head, whiplash, collision with player, etc).	
Did you receive medical clearance to return to play following your most recent concussion?	

Neck Range of Motion

Movement Direction	Normal Range of Motion	Range of Motion (degrees)		Pain Present? (Y/N)
Flexion	80-90			
Extension	70			
Side Flexion	20-45			
Rotation	90			

Head and Neck Anthropometrics

Height:____________

Body Mass:________

Head Circumference:____________

Neck Circumference:____________

Neck Length:_______

Maximal Isometric Strength Measures

	Trial 1	Trial 2	Trial 3	Average
Flexion				
Extension				
Side Flexion				

Appendix D:

Warm-Up and Cool-Down

Standardized Dynamic Warmup

Participants will warm up on stationary bike for 5 minutes at a moderate pace. The intensity should be set high enough to elevate heart rate and increase blood flow to muscles, but the participant should still be able to carry on a conversation.

Dynamic Neck Exercises

The participants will perform 12 repetitions of each exercise using a **slow, continuous motion.**

1. Neck Flexion and Extension

Perform a “nodding” motion, flexing the neck to bring the chin to the chest and then extending the neck to look at the ceiling in a slow and controlled manner.

2. Right/Left Head Rotation

Rotate your head to one side as if looking over your shoulder, then rotate in the opposite direction to look over the other shoulder in a slow and controlled manner.

3. Right/Left Lateral Neck Flexion

Keep your head facing forward and move your ear down toward your shoulder until you feel a stretch along the opposite side of your neck. Do not shrug the shoulder to try to have the ear and the shoulder touch. Return to neutral and repeat on the opposite side in a slow and controlled manner.

Static Cool Down Stretches

Perform each stretch 2-3 times. **Hold each stretch for 15 to 20 seconds.**

1. Neck Flexion and Extension

Slowly tuck your chin and allow your head to drop down towards your chest. Apply slight pressure to the back of the head with either hand to increase the stretch. You should feel a stretching sensation in the neck and back. Slowly tilt your head backwards as if looking up towards the ceiling. Apply slight pressure to your forehead with either hand to increase the stretch.

2. Right/Left Neck Rotation

Slowly rotate your head to the side to look over your shoulder. Apply slight pressure to the side of the head to increase the stretch, as if trying to look further over your shoulder. You should feel a stretching sensation along the side of neck.

3. Right/Left Lateral Neck Flexion

Slowly laterally flex your head by bringing your ear to the shoulder. Apply slight pressure to the side of the head with the same side hand to increase the stretch. You should feel a stretching sensation in the opposite side of the neck.

All images have been adapted and modified from “Neck training 101”, 2011, by B. Contreras. https://bretcontreras.com/neck-training-101/

Appendix E:

Body Segment Parameters for Estimated Head Mass

Table E1

Body segment parameter data from Zatsiorsky et al. (1990), as modified by deLeva (1996)

Segment	Endpoint		Mass (%mass)		CM (%length)		Sagittal k (%length)		Transverse k (%length)		Longitudinal k (%length)	
	proximal	distal	female	male	female	male	female	male	female	male	female	male
Head	VERT	MIDG	6.68	6.94	58.94	59.76	33.0	36.2	35.9	37.6	31.8	31.2
	VERT	CERV	6.68	6.94	58.94	59.76	27.1	30.3	29.5	31.5	26.1	26.1
Trunk	SUPR	MIDH	42.57	43.46	41.51	44.86	35.7	37.2	33.9	34.7	17.1	19.1
	CERV	MIDH	42.57	43.46	41.51	44.86	30.7	32.8	29.2	30.6	14.7	16.9
	MIDS	MIDH	42.57	43.46	41.51	44.86	37.9	38.4	36.1	35.8	18.2	19.7
Upper Trunk	SUPR	XYPH	15.45	15.96	20.77	29.99	74.6	71.6	50.2	45.4	71.8	65.9
	CERV	XYPH	15.45	15.96	20.77	29.99	46.6	50.5	31.4	32.0	44.9	46.5
Mid Trunk	XYPH	OMPH	14.65	16.33	45.12	45.02	43.3	48.2	35.4	38.3	41.5	46.8
Lower Trunk	OMPH	MIDH	12.47	11.17	49.20	61.15	43.3	61.5	40.2	55.1	44.4	58.7
Upper Arm	SJC	EJC	2.55	2.71	57.54	57.72	27.8	28.5	26.0	26.9	14.8	15.8
Forearm	EJC	WJC	1.38	1.62	45.59	45.74	26.1	27.6	25.7	26.5	9.4	12.1
	EJC	STYL	1.38	1.62	45.59	45.74	26.3	27.8	25.9	26.7	9.5	12.2
Hand	WJC	MET3	0.56	0.61	74.74	79.00	53.1	62.8	45.4	51.3	33.5	40.1
	WJC	DAC3	0.56	0.61	74.74	79.00	24.4	28.8	20.8	23.5	15.4	18.4
	STYL	DAC3	0.56	0.61	74.74	79.00	24.1	28.5	20.6	23.3	15.2	18.2
	STYL	MET3	0.56	0.61	74.74	79.00	51.9	61.4	44.3	50.2	32.7	39.2
Thigh	HJC	KJC	14.78	14.16	36.12	40.95	36.9	32.9	36.4	32.9	16.2	14.9
Shank	KJC	LMAL	4.81	4.33	44.16	44.59	27.1	25.5	26.7	24.9	9.3	10.3
	KJC	AJC	4.81	4.33	44.16	44.59	26.7	25.1	26.3	24.6	9.2	10.2
	KJC	SPHY	4.81	4.33	44.16	44.59	27.5	25.8	27.1	25.3	9.4	10.5
Foot	HEEL	TTIP	1.29	1.37	40.14	44.15	29.9	25.7	27.9	24.5	13.9	12.4

Adapted from "Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters", P. de Leva, 1996, *Journal of Biomechanics, 29*(9), p. 1223-1230.

Appendix F:

Neckform Calibration and Data Transformation Procedures

Neckform Calibration

The first step in transforming the human cervical muscle strength values to torque measures was to calibrate the neckform. The goal of the calibration was to establish a relationship between human force and neckform torque to convert cervical muscle strength measures to torque measures. The surrogate head and neckform were mounted to a fixed base during the calibration. A lightweight chain was attached to the mechanical neckform at the base of the headform and was pulled taut to attach to a load cell at the opposite end (Figure F1). The chain was pulled tighter using an adjustable clamp, which caused the neckform to flex, extend, or laterally flex to 10 degrees. The angle was limited to 10 degrees because this angle has been used in previous neck strength research and is easily achieved by a healthy sample of human participants (Garcés et al., 2002). The headform was equipped with the CROM device for all trials to obtain a consistent angle measure of 10 degrees.

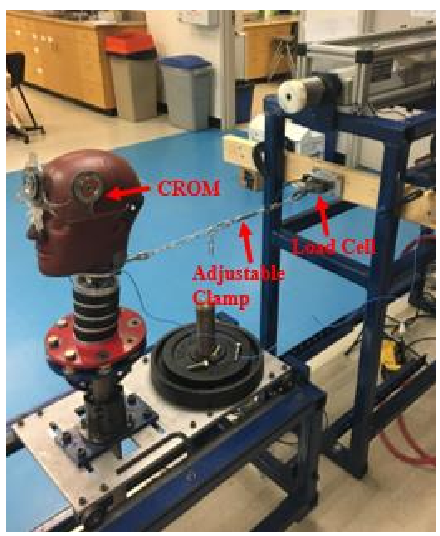

Figure F1. Neckform calibration setup. The chain attaching the neckform to the load cell was tightened using an adjustable clamp until the CROM indicated a 10-degree angle of flexion. Force was measured by the load cell.

The stiffness of the neckform was adjusted to 11 different torques ranging from 0.84Nm to 5.04Nm using the same method used in the head impact simulation testing, described in

Chapter 4 (Figure 12). The torques were selected based upon the force applied to the Chatillon gauge, which increased by 1N increments, due to the sensitivity of the instrument. For each neckfrom torque, the amount of force required to flex, extend, and laterally flex the neckform to 10 degrees was measured on a load cell to the nearest tenth of a Newton. Three trials were conducted for each direction for all neckform torque measures to ensure reliability, and an average of the three trials for each direction was calculated. Since Part II of the research was concerned with overall cervical muscle strength, the mean of the three directions was calculated to obtain an overall average force measure for each neckform torque (Table F1). Figure F2 illustrates the relationship between neckform torque and overall force, generating a linear relationship defined by Equation F1.

Table F1

Average force measures generated for varying neckform torques.

Neckform Torque (Nm)	Force to Adjust Torque (N)	Neck Flexion (N)	Neck Extension (N)	Neck Side Flexion (N)	Overall Strength (N)
0.84	2	317	176	289	261
1.36	3	318	187	295	267
1.68	4	327	208	320	285
2.10	5	329	211	337	292
2.52	6	340	227	357	308
2.94	7	359	237	368	321
3.36	8	411	250	401	354
3.78	9	421	314	452	396
4.20	10	434	319	456	403
4.62	11	475	323	470	423
5.04	12	471	353	473	432

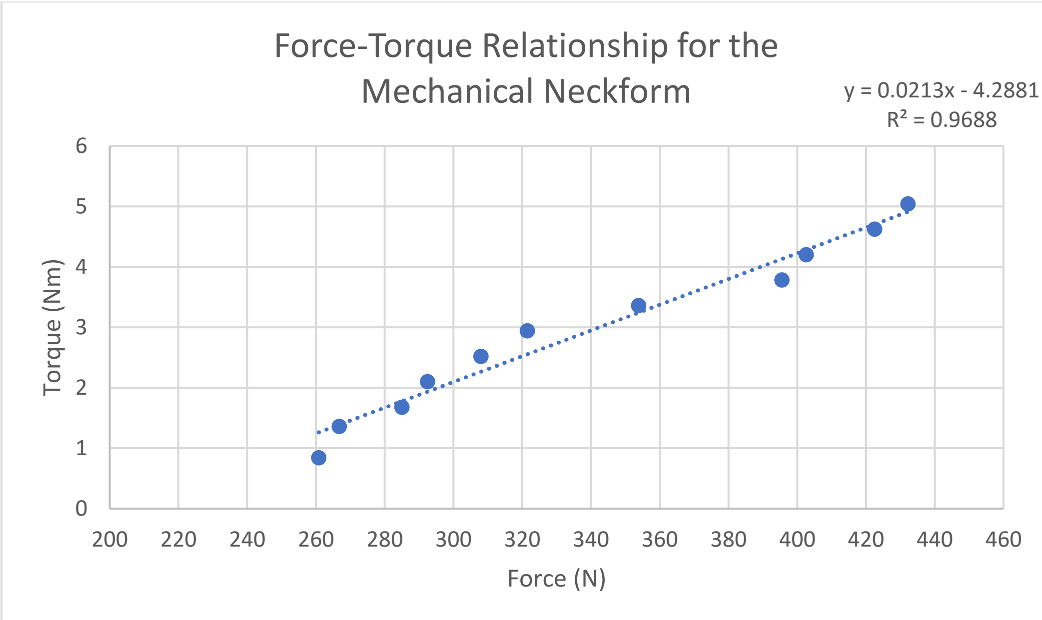

Figure F2. The relationship between torque and force for the mechanical neckform. The linear relationship is described by Equation F1.

$$y = 0.0213x - 4.2881 \quad \text{(F1)}$$

Where:

y= torque (Nm)
x= force (N)

Force-Torque Conversion

The calibration equation generated was used to convert the selected cervical muscle strength measures from Part I to the torque measures used to adjust the stiffness of the neckform in Part II. Specifically, the 10^{th}, 50^{th}, and 90^{th} percentile strength values were first scaled using the z-score technique and then converted to torque values. The original strength values were transformed to z-scores using Equation F2, converted to a scaled strength measure using Equation F3, and finally transformed to a corresponding torque measure using the established calibration equation (Equation F1). A scaling process was necessary due to the difference in magnitude of the human strength measures and the force measures obtained during the neckform calibration process (Table F1). This force-torque conversion process is summarized in Table F2

Due to the sensitivity of the load cell used during the torque adjustment process, the calculated torque measures had to be rounded to the nearest obtainable value, outlined in Table F1.

$$Z = \frac{x - \bar{X}}{SD} \quad \text{(F2)}$$

Where:

x = overall cervical muscle strength value (N)
$\bar{X}$= mean overall cervical muscle strength (N)
SD= standard deviation of cervical muscle strength measures (N)

$$Scaled\ Strength = \bar{X}_{neckform} + (Z * SD_{neckform}) \quad \text{(F3)}$$

Where:

$\bar{X}_{neckform}$= mean overall force of neckform during calibration
Z = corresponding z-score
$SD_{neckform}$= standard deviation of overall force of neckform during calibration

Table F2

Conversion and Scaling Process

Percentile	**10th**	**50th**	**90th**
Overall Cervical Muscle Strength (N)	58.64	82.08	108.00
Z-Score	-1.34	0	1.48
Scaled Value (N)	254.64	340.12	434.64
Calculated Neckform Torque (Nm)	1.14	2.96	4.97
Neckform Torque Used (N)	**1.36**	**2.94**	**4.62**
Neck Strength Represented	Weak	Average	Strong

Appendix G:

Descriptive Summary Tables for Impact Simulations

Table G1

Summary of Descriptive Statistics, mean (SD), for head impact simulations.

Mechanism of Injury	**Impact Location**	**Neckform Torque**	**Peak LA (g)**	**Peak SF (N)**	**Severity Index**
Direct Impact	Front	Weak	98.83 (39.47)	1597.57 (438.75)	315.66 (220.09)
		Average	105.29 (40.28)	1754.42 (671.47)	343.11 (239.05)
		Strong	104.10 (40.26)	1650.97 (588.04)	334.82 (238.18)
		Total	102.74 (39.24)	1667.65 (565.70)	331.20 (227.90)
	Rear	Weak	113.32 (27.93)	2279.04 (892.37)	388.49 (176.50)
		Average	119.96 (28.20)	2066.12 (537.78)	438.16 (201.39)
		Strong	123.92 (30.03)	2065.47 (567.50)	464.74 (217.11)
		Total	119.07 (28.46)	2136.88 (677.90)	430.46 (197.36)
	Side	Weak	111.71 (31.41)	2177.64 (695.47)	397.13 (188.32)
		Average	126.83 (28.81)	2465.57 (826.29)	444.68 (189.97)
		Strong	139.84 (37.54)	2489.24 (759.27)	526.88 (242.21)
		Total	126.13 (34.12)	2377.48 (759.43)	456.23 (210.93)
	Total	Weak	107.95 (33.22)	2018.08 (749.66)	367.09 (195.15)
		Average	117.36 (33.45)	2095.37 (735.26)	408.65 (211.90)
		Strong	122.62 (38.38)	2068.56 (718.85)	442.15 (241.69)
		Total	115.98 (35.38)	2060.67 (730.25)	405.96 (217.77)
Whiplash+Impact	Front	Weak	147.29 (27.07)	1503.46 (474.08)	609.01 (342.80)
		Average	159.88 (28.77)	1412.23 (292.80)	677.75 (339.11)
		Strong	155.47 (16.97)	1315.31 (298.93)	705.52 (364.72)
		Total	154.21 (24.85)	1410.33 (365.56)	664.09 (344.01)
	Rear	Weak	137.35 (40.45)	1364.81 (560.98)	761.31 (424.63)
		Average	139.52 (39.74)	1527.47 (469.34)	764.94 (414.17)
		Strong	141.64 (28.08)	1557.25 (534.92)	789.56 (423.22)
		Total	139.50 (35.79)	1483.18 (519.00)	771.94 (411.84)
	Side	Weak	138.26 (32.43)	1595.82 (508.70)	645.72 (332.43)
		Average	134.93 (19.32)	1692.26 (641.93)	676.60 (316.94)
		Strong	140.41 (18.48)	1522.04 (639.42)	623.82 (261.41)
		Total	137.87 (23.85)	1603.37 (591.23)	648.71 (299.36)
	Total	Weak	140.96 (33.35)	1488.03 (513.77)	672.02 (366.91)
		Average	144.78 (31.74)	1543.99 (492.61)	706.43 (353.91)
		Strong	145.84 (22.36)	1464.86 (511.82)	706.30 (355.11)
		Total	143.86 (29.42)	1498.96 (503.71)	694.92 (356.54)
Total	Front	Weak	123.06 (41.40)	1550.52 (451.86)	462.34 (320.17)
		Average	132.58 (44.21)	1583.32 (538.39)	510.43 (334.95)
		Strong	129.78 (40.06)	1483.14 (489.52)	520.17 (356.76)
		Total	128.48 (41.68)	1538.99 (491.09)	497.65 (335.02)

	Rear	Weak	125.33 (36.31)	1821.92 (867.92)	574.90 (371.74)
		Average	129.74 (35.32)	1796.80 (566.92)	601.55 (360.81)
		Strong	132.78 (29.98)	1811.36 (600.79)	627.15 (369.74)
		Total	129.29 (33.76)	1810.03 (684.53)	601.20 (364.20)
	Side	Weak	124.99 (34.18)	1886.73 (668.29)	521.42 (294.25)
		Average	130.88 (24.48)	2078.91 (827.09)	560.64 (282.75)
		Strong	140.12 (29.11)	2005.64 (847.46)	575.35 (252.74)
		Total	132.00 (29.87)	1990.43 (780.81)	552.47 (275.15)
	Total	Weak	124.46 (37.04)	1753.06 (692.53)	519.55 (330.05)
		Average	131.07 (35.24)	1819.68 (681.41)	557.54 (326.47)
		Strong	134.23 (33.35)	1766.71 (690.89)	574.22 (330.02)
		Total	129.92 (35.35)	1779.82 (686.50)	550.44 (328.50)