The effect of impact velocity and neck stiffness on dynamic head response from body-first, head-to-glass impacts in ice hockey

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ABSTRACT

Ice hockey is a high-velocity, collision-heavy sport, increasing players' susceptibility to head impacts and traumatic brain injuries (TBIs). Recent studies emphasize the risk posed by sub-concussive and concussive impacts, particularly in professional leagues, where the intensity of play leads to repeated head traumas. Body-first impacts, such as shoulder-to-glass collisions, contribute to the incidence of TBIs in ice hockey, yet the biomechanical response involving neck stiffness and head kinematics during these impacts remains underexplored.

This study investigates the influence of neck muscle forces on head injuries during body-first impacts, specifically shoulder-to-glass collisions in professional ice hockey. By examining the effects of varying neck stiffness levels and impact velocities, this research aims to enhance biomechanical understanding and inform the development of preventive strategies to mitigate head injury risk.

Twenty professional hockey games from the 2016-2017 season were analyzed using video footage to identify 47 shoulder-first head impacts. Kinovea software quantified impact velocities and categorized them into low, medium, and high ranges for reconstruction. A Hybrid III headform, equipped with an accelerometer array and positioned on the University of Ottawa Neck Spring Apparatus (uONSA), simulated head impacts across three neck stiffness levels and three impact velocities. Maximum voluntary contraction (MVC) values guided the setup of neck stiffness conditions to represent the upper trapezius, splenius capitis, and sternocleidomastoid muscle groups. Each trial involved a high-speed impactor simulating shoulder-first impacts followed by head-to-glass contacts, with three trials per condition.

Twenty-seven shoulder impacts were collected using a fully crossed design, with three impacts at each combination of low, medium, and high neck stiffness and impact velocities of 3, 5, and 7 m/s. High-speed video and kinematic analysis examined dynamic responses during these impacts. Six impacts at 3 m/s with high and medium neck stiffness did not result in a secondary head-to-glass impact, instead displaying a whiplash-like motion. Mean peak linear and rotational accelerations and rotational velocity were calculated for each combination of velocity and neck stiffness, and dynamic response curves were generated. Results indicated that higher neck stiffness reduced linear and rotational acceleration at lower velocities, particularly 3 m/s, suggesting a potential protective effect. However, peak accelerations rose across all neck stiffness levels as impact velocity increased. For example, at 3 m/s, mean linear accelerations were 21.4 g, 10.98 g, and 10.03 g for low, medium, and high stiffness, respectively, while at 7 m/s, they reached 76.13 g, 94.40 g, and 64.30 g, respectively. Rotational accelerations followed a similar trend, with low neck stiffness producing higher values at lower velocities but converging as velocity increased.

A two-way ANOVA revealed significant main effects for neck stiffness and impact velocity on peak linear acceleration ($F=25.7$, $p<0.001$), rotational acceleration ($F=30.729$, $p<0.001$), and rotational velocity ($F=152.20$, $p<0.001$). Post-hoc analyses showed that each level of neck stiffness and impact velocity independently influenced peak accelerations, with significant differences across stiffness levels at various velocities. The study also identified five instances where the highest peak occurred during the head-to-glass impact rather than the shoulder contact. Impact events at lower velocities (3 m/s) with high and medium neck stiffness showed longer durations without secondary impacts, emphasizing that increased neck stiffness can limit

head movement in low-velocity impacts. Results aligned with prior research indicating that, while higher neck stiffness can reduce head motion, the effectiveness diminishes as impact velocity increases. For instance, medium stiffness at 7 m/s resulted in rotational acceleration values that approached the threshold associated with an 80% risk of brain injury (7900 rad/s^2).

This study's findings underscore the complex interplay between neck stiffness and impact velocity in influencing head kinematics. Although higher neck stiffness may reduce peak accelerations under low-velocity impacts, it may not be sufficient at higher velocities, as shown by instances where rotational accelerations at medium and high stiffness levels approached brain injury thresholds. Limitations include the discrepancy between laboratory simulations and real-world conditions, as the study used a 50th percentile male headform and simplified neck model that may not fully capture human neuromuscular responses. These results contribute to understanding the biomechanical factors in head injury dynamics, supporting the development of targeted interventions to enhance player safety in sports.

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CHAPTER 1: INTRODUCTION

1.1 Overview

Ice hockey is recognized as a skill-intensive sport, blending physicality and technical prowess. The sport is distinguished by its rapid pace, with players skating at high speeds while skillfully maneuvering a puck with their sticks. This dynamic environment requires athletes to exhibit high endurance, balance, and agility levels. The high-velocity nature of the sport frequently leads to collisions and physical confrontations, underscoring the need for robust physical conditioning. The National Hockey League (NHL) encompasses a gruelling 82-game season that significantly impacts player health, as evidenced by many players missing games due to injuries (Kuhn & Solomon, 2016).

There is an increasing awareness of the long-term consequences associated with physical sports, particularly concerning head impacts. The nature of injury and severity of traumatic brain injuries (TBI) are directly proportional to the magnitude of the trauma that occurred during impact (Oeur et al., 2015; Post et al., 2015; Kleiven, 2007). Athletes who suffer multiple concussions often endure prolonged recovery periods and heightened vulnerability to future injuries (Guskiewicz et al., 2003). Even in the absence of symptomatic concussions, players in contact sports often experience numerous sub-concussive impacts throughout their careers (McKee et al., 2009). Current research emphasizes the adverse effects of repetitive head trauma, underlining its association with neurodegenerative diseases, cognitive impairments, mental health issues, and chronic traumatic encephalopathy (CTE), a degenerative brain condition marked by memory loss, confusion, aggression, and dementia (McKee et al., 2009). These findings support the need for continuous monitoring and preventive strategies to address the long-term neurological consequences of head trauma in sports (Broglio et al., 2012).

The inherent physicality, high speeds, and hard playing surface of ice hockey make players susceptible to head trauma. Head injuries in ice hockey can result from direct blows to the head due to body checks, collisions with the boards or glass, or falls onto the ice (Biasca et al., 2002). Shoulder collisions have been identified as the most common cause of brain trauma in professional ice hockey (Hutchison et al., 2015b; Post et al., 2019). While helmets protect against skull fractures, they do not fully mitigate the rotational accelerations that can cause brain injuries (Post et al., 2014). Concussions occur at rates ranging from 5.8 to 6.1 per 100 games in professional leagues, accounting for approximately 14-30% of all head injuries sustained during play (Andrews et al., 2022). The complexity of concussions, with their diverse signs and symptoms, presents a significant challenge in fully describing the physical trauma inflicted on the brain (Karton et al., 2021). Velocity, impact location, direction, and compliance influence the severity of head impacts. In response to growing concerns about player safety, governing bodies like the NHL have implemented regulatory changes, such as Rule 48, which penalizes hits to the head. Despite these measures, the incidence of head injuries remains high, necessitating a deeper understanding of head impact severity and its implications for TBI (Andrews et al., 2022).

Research in head impact reconstructions is crucial for understanding the biomechanics of TBI and developing preventive measures. Head impact reconstructions utilize methodologies such as physical reconstructions and finite element modelling to simulate and analyze head trauma. Kleiven (2007) demonstrated the ability to simulate brain responses by examining brain tissue stress and strain. Laboratory reconstructions often employ comprehensive tools like accelerometers to identify critical factors influencing injury severity, such as impact velocity and location (Hoshizaki et al., 2014). This research aids in designing protective gear, such as helmets, and informs safety regulations to reduce head trauma prevalence in sports (Rowson & Duma, 2013).

There are possible mechanisms to protect the head from injuries, including ice hockey, one being the development of cervical musculature. The neck's complex anatomical structure and critical role in supporting and protecting the head make it an essential factor in biomechanical injury research. Neck mechanics are primarily studied through computational models that simulate neck responses under various conditions. Research to gain insight into automotive crashes has utilized neck simulations to predict cervical spine injuries (Nightingale et al., 1997), while test dummies have been used to understand the effects of impact accelerations, head kinematics, and muscle activation on neck injuries (Yoganandan et al., 2009). These studies underscore the importance of considering the neck's role in head impact dynamics. Research by Cournoyer et al. (2021) examining neck stiffness and head kinematics in youth football demonstrated mixed results in reducing head dynamic response. The authors reported that impact conditions play a major role in the ability of the neck to minimize brain injury.

Body-first head impact reconstructions help understand the role of the body and pre-impact conditions on brain injury outcomes. A body-first approach is important to improve the accuracy of prediction models and contribute to developing safety guidelines. Body checking, a standard play style in professional ice hockey, significantly increases the risk of head impacts due to collisions with body segments (Robidoux et al., 2020; Karton et al., 2021). Empirical investigations by Hutchison et al. (2015b) and Aguiar et al. (2020) have highlighted the prevalence and patterns of head impacts in ice hockey, particularly noting the significant proportion of impacts occurring along the rink's perimeter and involving contact with boards or glass. As ice hockey continues to employ a style of play that incorporates the glass and boards, players are exposed to impacts involving various body parts, leading to head impacts. When players are checked or pressed against the glass, elbows or shoulders are used to mitigate head impacts, as the activation of core and neck muscles helps stabilize the head (Choi et al., 2017). This research is vital for understanding the implications of such impacts along the perimeter using a body-first approach. Longitudinal studies, such as those by Andrews et al. (2022), have explored the potential career ramifications for players with concussions, reporting correlations between such injuries and diminished player longevity and performance.

The mechanics of body-first impacts are challenging to investigate due to the complexity of the human neck and the intricacy of biomechanical reconstructions. The human neck comprises several ligaments, muscles, and intervertebral discs (Siegmund et al., 2007). Findings indicate that pre-impact loading can significantly influence head response and the resultant brain injury (Smith et al., 2015). Research on body-first impacts has shown variable effects on head impact velocity, with some studies reporting a decrease and others an increase, particularly in rotational velocity (Hajiaghamemar et al., 2015; Smith et al., 2015; Choi et al., 2017). Research has shown inconsistent results regarding neck strength's impact on concussive injuries, with some studies indicating decreased risk (Collins et al., 2014) and others reporting no relationship (Schmidt et al., 2014). Similarly, increased neck muscle activity has decreased kinematic response to direct head impacts in some cases (Viano et al., 2007) but not in others (Mihalik et al., 2011). Rousseau and colleagues (2010) concluded that events where muscles are braced in anticipation might not effectively reduce the risk of TBI when looking at three different levels of neckform compliance during a helmeted impact.

Biomechanical studies have significantly advanced the understanding of head injuries in ice hockey, focusing on metrics such as peak linear and rotational acceleration. The role of cervical musculature in modulating head acceleration dynamics during body-first impacts has also received attention. Increased neck stiffness has been associated with decreased linear and rotational head acceleration (Schmidt et al., 2014), highlighting the potential role of neck strength in mitigating head injury risk. Understanding the mechanics of body-first impacts remains complex due to the neck's intricacies and the variability of impact scenarios. However, insights from previous research suggest that pre-impact loading can significantly influence head response and resultant brain injury, with implications for injury prevention strategies (Smith et al., 2015).

This study investigates the effect of neck muscle forces during body-first impacts, specifically shoulder-to-glass impacts, on head injuries in ice hockey. By exploring the biomechanical responses associated with different levels of neck stiffness and impact velocities, this research seeks to inform the development of effective injury prevention strategies to enhance the well-being of athletes participating in ice hockey. This study aims to contribute to the broader understanding of head injury mechanisms and inform targeted interventions to improve player safety in the sport by describing biomechanical factors underlying head injuries in ice hockey.

1.2 Purpose

The proposed research is designed to provide insight into the specific mechanism and understanding of head-to-glass head impact events in professional ice hockey. It will establish how shoulder-first head impacts to glass are affected by neck stiffness. Additionally, it will provide insight into the head dynamic response properties of body-first head contact events.

1.3 Research Question

What is the influence of inbound velocity and neck stiffness on peak linear and rotational head acceleration and rotational velocity during head-to-glass events where the shoulder is the first point of contact in professional ice hockey?

1.4 Objectives

1) Understand the mechanism of body-first head impacts in ice hockey
2) To compare the influence of varying neck stiffness when a body first head impact occurs

3) To compare the influence of velocity on the dynamic response of the head when a body-first impact occurs
4) To compare the overall dynamic head response of body-first impacts

1.5 Independent Variables

This study has two independent variables, each with three levels. Neck stiffness, including low neck stiffness, is represented by 63.4 N for lateral flexion, 54.7 N for extension, and 67.63 N for flexion. Medium neck stiffness is represented by 179.5 N for lateral flexion, 168.5 N for extension and 182.9 N for flexion. High neck stiffness is represented by 243.9 N for lateral flexion, 203 N for extension and 254.79 N for flexion. The second independent variable was impact velocity measured at three levels: 3m/s, 5m/s and 7m/s.

1.6 Dependent Variables

The three dependent variables for this study are the peak resultant linear acceleration of the head at impact, the peak resultant rotational acceleration of the head at impact, and the peak resultant rotational velocity of the head at impact.

1.7 Hypothesis

An increase in neck stiffness will significantly decrease the head's linear and rotational impact kinematics for shoulder-first head-to-glass impacts at three impact velocities.

1.8 Limitations

The human neck was represented by a surrogate (uONSA) with associated limitations, including the lack of identical anatomical positioning. The neck surrogate only accounted for three muscle groups, the main contributors in extension, flexion and lateral flexion. When attaching springs to the uONSA, they do not fully represent human muscle forces. Additionally, the springs move in one dimension as they are extension springs and do not account for the multidimensional movements displayed by the human neck. The 50^{th} percentile Hybrid III head form was used in reconstruction events to represent the adult head. Due to the headform comprising aluminum and rubber, it is not fully biofidelic and does not represent the actual properties of human tissue; therefore, the resulting dynamic response of the headform does not fully represent real-life events.

1.9 Delimitations

Springs were attached to the neck apparatus to represent the three muscle groups' maximum voluntary contraction (MVC). Due to the mechanical properties of extension springs, the spring sets were not identical for each MVC value.

CHAPTER 2: LITERATURE REVIEW

2.1 Brain Injuries

Brain injuries present a significant issue due to their prevalence and the profound impact they can have on an individual's quality of life. The human brain is particularly vulnerable to rapid acceleration and deceleration movements, which can cause it to collide with the skull, leading to potential injuries such as contusions (Gennarelli & Thibault, 1982). Brain injuries can be categorized by focal injuries, which occur in a single area in the brain, or diffuse brain injuries, which encompass axonal trauma. Traumatic brain injuries (TBI) represent more severe injuries, such as skull fractures or hematomas, while diffuse injuries, such as diffuse axonal injury, are due to extensive neural tissue shearing (Hoshizaki & Brien, 2004). A concussion can be classified as a mild traumatic brain injury (mTBI), a less severe form of a diffuse injury. mTBI have become a growing concern in sports due to their potentially severe long-term consequences. The severity of mTBIs ranges from mild concussion symptoms causing temporary functional impairment to more severe injuries that can cause long-term cognitive deficits and disturbances. Although there is no single specific definition of concussions, the International Conference on Concussions in Sport defines a concussion to be a mTBI caused by a direct blow to the head, neck or body, resulting in an impulsive force being transmitted to the brain (Patricios et al., 2023). This definition also refers to neurotransmitter and metabolic changes and alludes to multiple timelines for the onset of symptoms. Ommaya and Gennarelli (1974) discussed the notion that the severity of injury depends on the extent of axonal damage occurring in parts of the brain. More specifically, severe injuries happen when axonal injury encompasses more of the brain tissues, whereas less severe injury would occur when only specific areas of the brain are affected, noting the importance of classifying the strain magnitudes, volume and regions of brain tissue when examining injury severity. Kleiven (2013) discusses the primary cause of concussion being the deformation or strain that occurs in the brain tissue during rapid accelerations.

Concussions are caused by a bump, blow or jolt to the head, leading to a temporary disruption in brain function, including neuronal damage, blood-brain barrier disruption and neuroinflammation (McCrory et al., 2017); Smith et al., 2003). Concussions are diagnosed on a symptomatic basis, making the injury criteria more subjective through the presentation of symptoms; however, they are not objective diagnoses of trauma occurring to the brain. Diagnosing concussions can be challenging as their wide range of physical symptoms can span from headaches, dizziness or loss of consciousness. Athletes may sustain multiple concussions over their careers, sometimes without being aware of them, and others may go undiagnosed due to ignoring their symptoms. The impact of brain injuries extends beyond the immediate symptoms.

Neurodegenerative diseases have gained significant attention, particularly among athletes in contact sports, due to their long-term effects from repetitive brain trauma. Recurrent concussions have been linked to long-term cognitive impairments, emotional disturbances and an increased risk of neurodegenerative diseases like CTE (Lehman et al., 2012). Chronic traumatic encephalopathy (CTE) is one of the most well-documented conditions linked to repeated head impacts. CTE is a progressive degenerative disease found in individuals with a history of repetitive brain trauma, including symptomatic concussions and asymptomatic subconcussive impacts (McKee et al., 2013). It is characterized by an accumulation of tau protein in the brain, leading to symptoms such as memory loss, confusion, aggression and dementia (McKee et al., 2009). Head injuries can result from both direct or indirect hits to the head, often leading to high linear and rotational acceleration. Biomechanical research describes linear and rotational accelerations contribute to concussions (Meaney & Smith, 2011), particularly rotational accelerations, as this can cause shearing of brain tissues (Gennarelli & Thibault, 1982).

2.1.1 Mechanisms of Injury in Ice Hockey

Ice hockey is a fast-paced and physically demanding sport with a high injury risk. Several characteristics of the game contribute to injury, including event type, mechanism and equipment. Understanding the mechanisms behind these injuries is crucial for developing effective prevention strategies. In a single NHL season, 51% of all players missed at least one game due to significant injury (Kuhn & Solomon, 2016). Hutchison and colleagues (2015b) conducted a video analysis study finding that throughout 4299 regular season NHL games, a total of 260 concussions occurred from 2006-2009, analyzing 197. The same study found that 68% of those concussions had initial contact with the player's head. They found that 62% of concussions were directly related to head contact either from a shoulder, elbow or glove during gameplay, noting 47% of head contact occurred to the lateral aspect of the head (Hutchison et al., 2015b). Player positioning had notable risk levels; forwards accounted for 65% of those documented concussions, 32% for defensemen, and 3% for goalies (Hutchison et al., 2015a). The research found that players who sustained concussions were seen to have fewer games played later in their careers, suggesting that while there may not always be immediate adverse effects on performance, the longevity of their careers is at risk (Andrews et al., 2022). Aguiar et al. (2023) found that players who visibly displayed signs of concussion after a hit to the head experienced 1.3 times greater peak head rotational velocities.

2.1.2 Environmental Factors

The enclosed playing surface, rigid materials, and hard ice surface are the environmental factors associated with ice hockey. The design and material composition of rink boards and glass can impact injury rates. Flexible boards and seamless glass systems have been introduced in some rinks to reduce the severity of impacts on the perimeter. The adoption of flexible boards in arenas has been associated with a reduction in concussion rates, as they absorb some impact force, reducing the risk of neurotrauma for players (Hutchison et al., 2011). Tuominen et al., 2017 demonstrated that the high stiffness of the glass can lead to significant acceleration on the head, increasing the risk of concussions. The study used linear and rotational acceleration metrics in men's international ice hockey and noted that impacts with the glass often exceed the thresholds associated with concussive injuries. They found a 29% lower risk of injury where arenas had flexible boards and glass compared to those with traditional boards and glass. Additionally, they noted a decrease in overall injury rates (shoulders, knees) when flexible glass and boards were installed.

2.1.3 Body Checking

Body checking is a fundamental component of professional ice hockey and is a significant cause of injuries, notably the most common cause of all ice hockey injuries (Cusimano et al., 2011). The biomechanics of body checking involve high-impact collisions that often lead to head and neck injuries. Black and colleagues

(Black et al., 2017) reported a higher risk of concussions and other musculoskeletal injuries in leagues where body checking is allowed compared to non-checking leagues. It was reported that shoulder-to-head impacts involved around 15% of a player's body mass, noting a much higher difference when impacting with an elbow, which resulted in 3-5% of a player's mass (Rousseau & Hoshizaki, 2015). In youth, linear acceleration was found to be higher for unanticipated collisions. In contrast, rotational acceleration did not vary depending on an anticipated or unanticipated body check, suggesting severe impacts may be dangerous regardless of whether the body check is anticipated (Mihalik et al., 2010). Mihalik et al. (2010) demonstrated that checks to the head and body result in significant linear and rotational accelerations, leading contributors to concussions. Given the high prevalence of body contact in professional ice hockey, studying a head impact where the body contact occurs before the head contacts a surface could be influential in understanding the dynamic responses of such injuries.

2.1.4 Head to Glass events

Head-to-glass impacts in ice hockey typically occur during body checks or falls near the rink's perimeter. Guskiewicz et al. (2007) conducted a comprehensive study on concussion incidence in ice hockey, identifying that impacts with the boards and glass accounted for a significant proportion of these injuries. When following junior hockey players, impacts with the glass were associated with prolonged recovery times in comparison with other mechanisms of injury (Echlin et al., 2010). A study utilizing video analysis performed by Hutchison and colleagues (2015) found that 53% of head impacts occurred along the perimeter of the ice surface, including sideboards, end boards, glass, corners and side of the net. Noting that over half of the head impacts occur along the perimeter of the playing field, of these documented instances, 35% involved players' heads coming in contact with the boards and glass. Additionally, Aguiar and colleagues (2023) found head-to-glass impacts to be 3.7 times more common than head-to-board impacts when looking at male university hockey. Post et al. (2019) found that head-to-glass impacts typically resulted in 15-20ms durations, and they noted that these shorter duration events resulted in higher magnitudes of 50% risk. With a large number involving the glass, little research has examined the dynamic response effects of impacts along the glass, especially involving body contact first.

2.2 Brain Trauma Metrics

2.2.1 Linear Acceleration

Peak linear resultant acceleration is a common biomechanical parameter used to measure the accelerations involved in TBIs. It refers to the maximum linear acceleration experienced by the head during impact, precisely the result of a force directed through the head's center of mass (Robertson et al., 2014), commonly measured in units of gravity (g). It represents the change in velocity over a specific time interval. Previous studies have utilized linear acceleration as a predictor of brain injuries, notably in American football; however, more recently, it has been shown to predict focal injuries and not diffuse injuries such as concussions (Post & Hoshizaki, 2012). Meaney and Smith (2011) developed a computational model demonstrating that linear accelerations cause brain deformation, a critical factor in developing TBIs. Rowson et al. (2013) conducted a study using helmet-mounted accelerometers and identified a threshold of approximately 100g for linear acceleration, beyond which the risk of concussion significantly increased. Walsh et al. (2011) found that linear acceleration values were highest from front and side impacts. Additionally, Zhang and colleagues (2004) determined a 25%, 50% and 80% risk of sustaining a brain injury using maximum resultant linear acceleration values of 66, 82 and 106g.

2.2.2 Rotational Acceleration

Rotational acceleration has emerged as an important metric in predicting TBIs. The human brain is particularly susceptible to rotational accelerations due to its structure and the varying densities of tissues. Peak rotational acceleration measures the rotational acceleration experienced by the head during an impact in radians per second squared (rad/s2), quantifying the rotational accelerations acting on the brain, often leading to the shearing and stretching of neural tissues. Research by Zhang et al. (2004) described rotational accelerations more strongly associated with diffuse axonal injury, a common type of TBI, due to the shearing of brain tissue (Gennarelli & Thibault, 1982). Rotational accelerations contribute to the deformation of brain tissues, disrupting neural pathways and leading to clinical symptoms of concussions and producing more significant tissue shearing and brain deformation than linear acceleration (Meaney & Smith, 2011). Zhang et al. (2004) found that rotational accelerations exceeding 5,500 rad/s2 are a predictor of 50% risk or injury. Those same researchers described maximum resultant rotational acceleration values for sustaining a 25%, 50% and 80% risk of brain injury as 4600, 5900 and 7900 rad/s^2, respectively (Zhang et al., 2004). Mihalik et al. (Mihalik et al., 2007) found that athletes experiencing high rotational impacts exhibited prolonged cognitive impairment compared to those with primarily linear impacts. Walsh et al. (2011) reported that rotational acceleration was highest amongst side and rear impacts. Rowson and Duma (2011) found rotational kinematics to better predict concussions than linear metrics alone. Linear acceleration does not reflect the brain trauma and may be insufficient without considering rotational acceleration, as real-world impacts would typically have both linear and rotational accelerations occurring (Clark et al., 2018).

2.2.3 Rotational Velocity

Rotational velocity of head movement is quantified in radians per second (rad/s). When the head experiences rotational accelerations, different parts of the brain undergo varying degrees of acceleration. Higher rotational velocities are associated with more severe axonal injuries (Gennarelli & Thibault, 1982). Zhang et al. (2004) identify that rotational velocities exceeding 30 rad/s are above the threshold for concussive injuries. Research has linked high rotational velocities to immediate and long-term neurological issues, noting significant cognitive deficits and longer recovery times (McAllister et al., 2012; McKee et al., 2013).

2.3 Laboratory Methods

2.3.1 Video Analysis

Video analysis provides detailed insights into injury mechanics. The ability to capture frame-by-frame allows the assessment of complex movements with precision. High-speed cameras and video analysis software enable accurate tracking of body segments, velocities, and distances. Video analysis is a non-invasive technique for acquiring critical data from on-field action and translating it to a laboratory setting. Video analysis provides information to categorize the impact, including velocity and location, that can later be used to create laboratory reconstructions. There are limitations to video analysis, including player positioning and motion, camera view and perspective grid (Post et al., 2018). Errors from broadcasting camera views amounted up to 0.7m/s for a 4.5m/s skater, however the error increased up to 1.3m/s for faster skaters (Post et al., 2018).

2.3.2 Lab Reconstruction

Common methods for researching brain injuries in sports include laboratory reconstructions. These involve the use of impactor machines, head forms and accelerometers. Anthropomorphic test devices (ATDs) equipped with sensors measuring impact kinematics provide valuable data for validating and applying computational models (Mertz & Irwin, 2015; Gabler et al., 2022). A hybrid III head form provides a reliable and accurate impact response while considering impact characteristics such as mass, velocity and location (Oeur et al., 2015). The hybrid III headform was built to approximate the same mass and size distributions as the adult male head, constructed with steel and a vinyl cap to replicate the skull and skin of a human.

2.4 Neck Anatomy

The neck is a complex structure comprised primarily of the cervical spine, including vertebrae connected by intervertebral discs (IVDs) and ligaments that provide support and stability for the head (Cronin et al., 2018). The structural components of the neck are hard tissue, bones, and soft tissue (Cronin et al., 2018).

2.4.1 Neck Fundamentals

The human neck allows for an extensive range of motion in several planes comprising ligaments, tendons, muscle tissue and intervertebral discs (Siegmund et al., 2007). Cronin et al. (2018) described motions within the cervical spine as accounted for by 29 muscles, often found in symmetrical pairings. The manubrium of the sternum marks the anterior limit, the first thoracic vertebra's posterior limit. In contrast, the first pair of ribs and their costal cartilage mark the inferior limit of the neck (Cronin et al., 2018). The cervical spine comprises seven cervical vertebrae, C1 to C7. The upper cervical spine includes the first and second vertebrae, while the lower cervical spine consists of vertebrae C3 to C7 (Cronin et al., 2018). The upper cervical spine connects through 10 ligaments in a complex arrangement along with articular cartilage that allows for an extensive range of motion (Cronin et al., 2018). Bogduk and colleagues ((2000) detailed the C2-C3 junction as a separate spine compartment. C3-C7 reflect a common shape composed of a vertebral body, vertebral foramen enclosing the spinal cord, and transverse, articular and spinous processes (Cronin et al., 2018). It was deemed subtle and precise differences between the kinematics of other vertebrae, the superior articular processes of C3 face upwards and backward and sitting at about 40 degrees medially (Bogduk & Mercer, 2000). This noted difference implies that the axis of the C2-C3 vertebra should act differently than lower cervical segments (Bogduk & Mercer, 2000). Vertebral bodies are stacked upon one another, separated by intervertebral discs (Bogduk & Mercer, 2000). The union between the head and the atlas exists at the atlanto-occipital joints, allowing for nodding movements where, more notably, the head and atlas function as one unit (Bogduk & Mercer, 2000). Anatomically, the only possible movements are flexion and extension through the atlanto-occipital joint, flexion when the condyles roll forwards and slide backwards across the anterior walls of their sockets, and extension achieved by such movements conversely (Bogduk & Mercer, 2000).

2.4.2 Neck Muscle Mechanics

The neck includes more than 25 muscle pairs with multi-joint insertions and complex lines of action (Fice et al., 2018). Moore and Dalley (2013) concluded that the C0-C1 junction is where the maximum flexion/extension motions occur within the cervical spine, whereas axial rotation primarily occurs within C1-C2. Individualized biomechanical functioning of neck musculature is often complex to characterize because humans cannot voluntarily activate isolated neck muscles. When the cervical musculature is relaxed, the effective mass of the head is directly influenced by the impact force, leading to rapid acceleration (Schmidt et al., 2014). Schmidt and colleagues (2014) examined cervical isometric strength amongst football players, identifying low and high categories. They reported flexion, extension and lateral flexion values, where low presented as 0.18 Nm/kg, 0.43 Nm/kg and 0.38 Nm/kg, respectively, compared to high being 0.3 Nm/kg, 0.62 Nm/kg and 0.615 Nm/kg respectively.

2.4.3 Concussion and Neck Strength

In sports, neck muscles play an important role in absorbing and distributing forces to reduce the risk of head and neck injuries (Collins et al., 2014). Strengthening these muscles increases the effective moveable mass of the head, neck, and torso, potentially mitigating the severity of impacts (Mihalik et al., 2011). Anticipatory cervical muscle activation can occur independently of neck strength (Eckner et al., 2014). Schmidt et al. (2014) found no significant difference in the odds of moderate and severe head impacts between players with stronger versus weaker neck muscles. However, players with stiffer neck musculature during anticipated force extension had reduced odds of sustaining severe impacts.

The anticipation of impact significantly affects neck muscle activation. For example, falling backwards or sideways often activates core and neck muscles to protect the head (Choi et al., 2017). In contrast, unanticipated impacts, where cervical musculature is not tensed, reduce the head's effective mass to only the head (Mihalik et al., 2011). Smaller and weaker necks will likely experience greater linear and rotational displacements and accelerations (Eckner et al., 2014). However, Mihalik et al. (2011) found no linear and rotational acceleration differences among athletes with varying cervical muscle strengths.

Most strategies to reduce head impacts in sports focus on helmet design, functional changes, the introduction of helmets, and the adoption of rule changes (Collins et al., 2014). Neck strength measurements often require expensive equipment and extensive methods, leading to limited research evaluating anthropometric measurements as potential concussion risk factors (Collins et al., 2014). While much attention has been given to helmets and protective gear, emerging research suggests that the strength and stability of neck muscles may also play an important role in preventing head injuries (Choi et al., 2017; Eckner et al., 2014; Collins et al., 2014). Neck strength is important in many sports as it contributes to injury prevention and enhances performance. The kinematics of the head during impact are influenced by neck musculature, as described by Cronin et al. (2018) as an important factor for head injury prevention.

Rousseau and colleagues (2010) demonstrated that a reduction in rotational acceleration occurred with an increase in neck form compliance (stiffness). The authors concluded that athletes involved in unexpected collisions may have a greater risk of sustaining a head injury (Rousseau et al., 2010). Cournoyer et al. (2021) reported that the effect of increased neck stiffness on the risk of head injury was dependent on impact condition, specifically noting impact location, striking mass and impact velocity. Their results did not conclude a significant change in the risk of brain injury occurring with an increase in neck stiffness (Cournoyer et al., 2021).

Schmidt et al. (2014) examined American football players and found no relationship between stronger and larger neck muscles and mitigated head impact severity. However, they determined that greater cervical stiffness and lower angular displacement after impact reduced the odds of high-magnitude head impacts. Athletes with smaller, weaker necks will likely experience elevated resultant linear and angular head displacement, velocities, and accelerations after impact (Eckner et al., 2014). Choi et al. (2017) reported that head impact velocity, both horizontal and vertical, decreased when the sternocleidomastoid (SCM) increased in activity during backward falls in youth. The study concluded that SCM activation during backward falls to avoid head impacts did not rely on full maximal activation, except for two participants who exceeded maximum isometric conditions measured during muscle testing.

Collins et al. (2014) examined neck strength and concussion incidence in high school athletes, finding that those diagnosed with concussions had smaller mean neck circumferences and overall neck strength than those who remained uninjured. They concluded that neck strength significantly predicted concussion risk for male and female lacrosse, soccer, and basketball athletes at the high school level. Previous research shows how stiffer necks reduce translational head displacement, velocity, and acceleration when using video reconstructions of concussive head impacts within professional American football players (Viano et al., 2007). Evidence suggests that anticipatory muscle activation in bracing for an impact may reduce the overall dynamic head response in a collision (Eckner et al., 2014).

In summary, neck muscles can be influential in sports for absorbing and distributing forces and reducing head and neck injuries (Collins et al., 2014); however, the effect of neck strengthening is not fully understood. Strengthening these muscles increases the effective movable mass of the head, neck and torso, potentially lessening impact severity (Mihalik et al., 2011). While anticipatory cervical muscle activation can occur independently of neck strength (Eckner et al., 2014), stiffening neck muscles during anticipated force extension reduces the odds of severe impacts (Schmidt et al., 2014). Anticipation of impact activates core and neck muscles, protecting the head, while unanticipated impacts increase the risk of greater displacements and accelerations, particularly in those with weaker necks (Choi et al., 2017; Mihalik et al., 2011). Despite strategies focusing on helmet design and rule changes, emerging research highlights the role of neck strength and stability in preventing head injuries (Collins et al., 2014; Cronin et al., 2018). Studies indicate that greater cervical stiffness and muscle activation can reduce the severity of head impacts, underscoring the importance of neck strength in injury prevention (Schmidt et al., 2014; Eckner et al., 2014; Collins et al., 2014).

2.5 Body First Impacts

Most research on biomechanical injury mechanisms focuses on direct head impacts (Smith et al., 2015). However, there is growing interest in the indirect, body-first head impact events that represent a large portion of head impacts. Body-first impacts, where the initial force of an impact is absorbed primarily by the body, have significant implications for head injury mechanisms, as research shows that pre-impact motion can contribute to injury risk due to the transfer of energy and momentum (Fleisig et al., 1996). The biomechanics of body-first impacts involve transferring kinetic energy from the point of contact through various body tissues. The severity of the injury from body-first head impacts depends on factors such as the direction and magnitude of the impact, as well as the specific event type. The body's energy absorption capacity during impact plays a crucial role in injury outcomes (Crandall, 2002).

2.6 Biomechanical Predictions

Impact velocity, mass, location, compliance and direction of impact contribute to the duration, peak and shape of response curves and influence the severity of a head impact (Karton & Hoshizaki, 2018). Understanding head injuries through a trauma profile gives a more in-depth approach to the injury sustained by the brain and helps guide innovation strategies, intervention and risk mitigation.

2.6.1 Magnitude

High-velocity impacts are typically characterized by high peak accelerations. Significant brain tissue deformation can result in severe injuries. These impacts usually have high-intensity accelerations over a brief period, resulting in high peak accelerations. Studies indicate that high-magnitude impacts are more likely to cause concussions and other serious brain injuries due to the rapid head motion. Post et al. (2017) reported that an increase in strain is often related to the magnitude of these impacts. While low-magnitude impacts result in lower peak accelerations, they can still cause significant brain deformation and injury, especially with repeated exposures. Studies by Zhang et al. (2004) using finite element models have shown how different impact magnitudes affect brain tissue deformation patterns. Maximum Principal Strain (MPS) is a measure of deformation within the brain tissue during impact. There is thought to be a 50% risk of sustaining a concussion when MPS values reach between 19-30% in the grey and white matter cerebrum (Zhang et al., 2004; Rousseau, 2014). It is an important parameter obtained using finite element models to calculate the risk of brain injuries, as they simulate the brain's response to impacts, providing insights into the injury mechanisms (Zhang et al., 2004). Concussions are linked to damaging strain levels, most notably in the corpus callosum during sporting events (Kleiven, 2007). As the duration of the impact increases, the magnitude of acceleration required to reach injurious levels of strain will decrease (Post et al., 2017). Growing evidence supports a significant risk of neurological injury sustained from repetitive head injuries often seen in contact sports, regardless of the high or low strain (Karton & Hoshizaki, 2018). Higher magnitude impacts, characterized by greater linear and rotational accelerations, are more likely to result in concussions (Rowson et al., 2012).

2.6.2 Duration

Short-duration impacts are typically characterized by high peak acceleration, while longer-duration results have lower peak acceleration, which can cause significant injury due to sustained accelerations. Short-duration impacts have high-intensity accelerations over a brief period, resulting in high peak accelerations, but short durations can also display low magnitudes. Studies report that short, intense impacts are more likely to cause concussions due to the rapid head motion they induce (Gennarelli & Thibault, 1982). However, concussive impacts in ice hockey involving the shoulder are typically long duration (Rousseau, 2014). Post et al. (2017) highlighted that the magnitude of brain strain is often related to the magnitude of these brief events. Events longer in duration may provide an environment where accelerations of lower magnitude could cause strain to the brain, due to prolonged application of force and ultimately result in injury (Hardy et al., 2001; Post et al., 2017). Research suggests impacts in ice hockey are mostly longer durations, typically longer than 15ms, noting these longer duration events, coupled with lower magnitudes, can create the risk for concussion (Post et al., 2019). Figure 1 (Post et al., 2019) displays acceleration outcomes compared to duration in head injury events commonly seen in ice hockey. Head-to-glass events tend to fall in the 15-20ms duration range with lower magnitude accelerations, which could potentially lead to an increased likelihood of concussion. Recognizing that impact magnitude and duration are important factors when considering head impact severity, Gurdjian et al. (1966) observed that higher accelerations could be sustained as long as the duration of the accelerations is short. In contrast, lower magnitudes of acceleration can be sustained for very short durations, noting that the head can tolerate linear acceleration values of 42g for up to several milliseconds without causing severe brain injury. Brain tissues have viscoelastic properties; therefore, different variations of acceleration and duration can present the same risk for injury. Research also reports peak resultant values may not be the only contributors to brain deformation (Post et al., 2012).

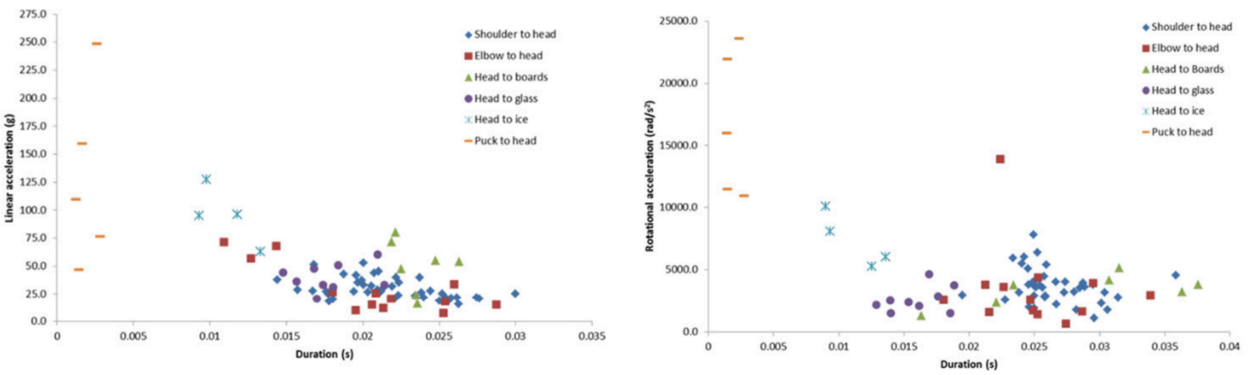

Figure 1: *Linear acceleration and rotational acceleration vs duration for multiple event types in ice hockey (Post et al., 2019)*

CHAPTER 3: METHODS

3.1 Overview

This study employed a combination of methods to examine brain injury risk in sports, including video analysis and event reconstructions. These techniques were important in investigating the effect of body-first impacts on head-impact kinematics. A Hybrid III headform with an unbiased neck with springs (uONSA) was used to represent three levels of neck stiffness with three body impact velocities using a high-speed impactor. The reconstructions were designed to reconstruct body-first head-to-glass impact events in ice hockey. A shoulder bumper at the end of the high-speed impactor was used as the initial impact, followed by glass fastened to the outside rails of the high-speed impactor to recreate the head-to-glass impacts.

3.2 Video Analysis

Video analysis is a widely used technique within the field of biomechanics, particularly for investigating head injuries in sports, as it enables the reconstruction of head impacts by providing data on impact velocity, mass, location and direction, with previous studies (Karton et al., 2021; Robidoux et al., 2020; Rousseau & Hoshizaki, 2015) demonstrating its effectiveness in improving our understanding of the dynamics of head trauma. Video analysis was employed in this study to establish reconstruction parameters for shoulder-first impacts.

Twenty open-sourced elite professional hockey games from the 2016-2017 regular professional ice hockey season were analyzed. Observed head impacts were clipped from the game video beginning five frames from impact until the impact moment, with the impact event and head contact visible. Two independent initial reviewers assessed each game, followed by an experienced, trained reviewer to ensure all documented head impacts were confirmed events by having both event type and contact with the head clearly visible. If the head contact was not clearly visible, it was classified as a suspected head impact and not included in the analysis as it could not be qualified (Chen et al., 2023). Forty-seven head impacts from game footage involving head-to-glass events where the shoulder constituted the initial point of contact were identified for further analysis. These forty-seven shoulder-first head contact videos were used to calculate the average distance between the head and the glass once the shoulder first makes contact with the glass, measured beginning at the initial point of contact of the shoulder with the glass.

The second part of the video analysis involved using Kinovea software to calculate impact velocity for each confirmed head impact incident. Employing frame-by-frame player positioning and landmarks on the playing surface, velocity was calculated by the distance between the head and impact event divided by time (Fig, 2). A perspective grid using known dimensions of landmarks on or around the ice surface, including boards, ice and nets, created a calibration and scaled the video footage; however, several factors, including player speed and camera angles, can have an effect on the accuracy of velocity calculations (Post et al., 2018). The impacts were analyzed four to five frames before impact, utilized to calculate time as footage was recorded at 25 frames per second.

$$Velocity = \frac{Distance\ travelled}{time\ to\ impact}$$

Figure 2: *Velocity calculation equation*

Velocity was then recorded for each head impact and placed into the following categories for observational purposes: very low ranging from 0-1.99 m/s, low 2-3.99 m/s, medium 4-5.99 m/s, high 6-7.99m/s and very high >8m/s (Fig, 3). To encompass the range of velocities that commonly occur during head-to-glass body first impacts in professional ice hockey, the median range of each low, medium and high category was chosen to represent low, medium and high reconstruction velocities.

Very Low	Low	Medium	High	Very High
0-2m/s	2-4m/s	4-6m/s	6-8m/s	>8m/s

Figure 3: *Velocity Categories*

3.3 Test Apparatus

3.3.1 Headform

A 50th percentile male Hybrid III headform (4.54kg) made up of a steel anterior and vinyl skin representing the average 50th adult male head was used (Fig, 4). It was equipped with Diversified Technical Systems (DTS) 6DX PRO and a SLICE NANO data acquisition system (Fig 5). Linear and rotational acceleration and rotational velocity data from each head impact were captured using a Slice Nano accelerometer array. This system consists of a triaxial accelerometer to measure linear acceleration and three angular rate sensors to measure angular velocity. The accelerometer data was collected at a sample rate of 20,000Hz and filtered using a CFC300 filter.

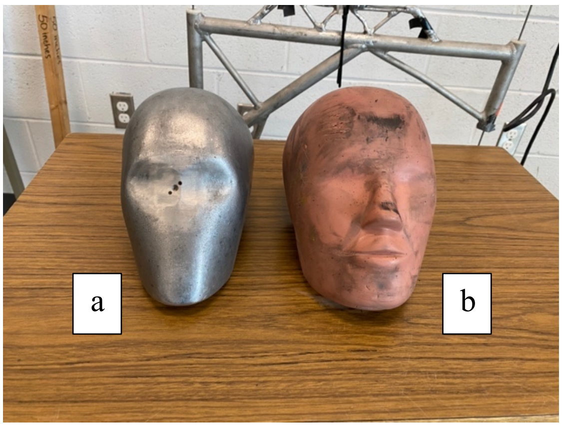

Figure 4: *Hybrid III Headform (a) and skin (b)*

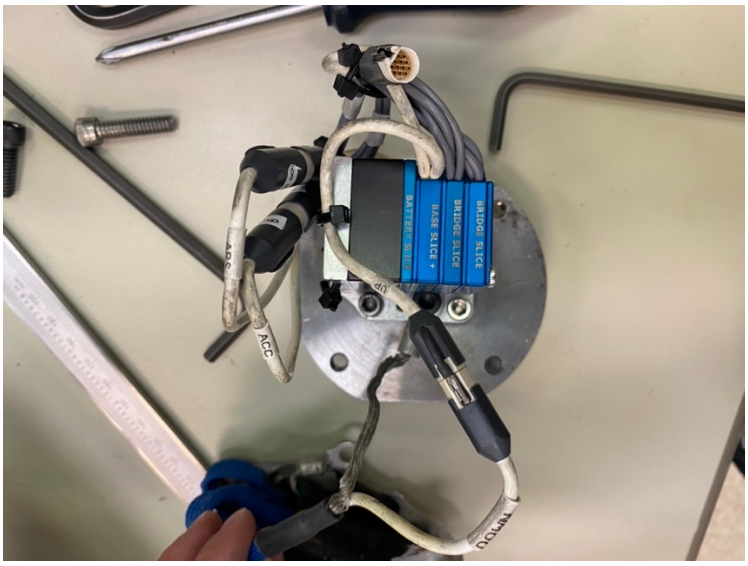

Figure 5: *Slice NANO Accelerometer array*

3.3.2 Neck form

The headform was attached to an unbiased neck form (1.54 kg) and the University of Ottawa Neck Spring Apparatus (uONSA) (Fig, 6) to approximate neck stiffness associated with the upper trapezius, splenius capitis sternocleidomastoid muscle groups. Maximum voluntary contraction (MVC) data from previous research (Hoshizaki, 2023) was employed to represent the neck muscle forces for six adult males, specifically neck flexion, extension, side flexion and rotation. MVC of neck extension represents splenius capitis, neck flexion represents the sternocleidomastoid, and lateral neck flexion represents the trapezius muscles (Cournoyer et al., 2021). MVCs in

Newtons (N) were binned into three categories, low, medium and high, to represent a spectrum of neck stiffness. The highest recorded MVC was taken as 100% voluntary contraction to distinguish three different neck stiffness groups. Low stiffness was representative of 25% MVC, medium stiffness was 75% MVC, and high stiffness was 100% MVC. Due to the purpose of this study representing professional athletes, medium stiffness was chosen to represent the higher end, assuming professional athletes tend to be trained and have greater muscle mass than untrained subjects. MVC values, specifically low (60, 54.5 and 59.75N), medium (180, 163.5 and 179.25N), and high (240, 218, 239N), represented flexion, extension and lateral flexion, respectively (Hoshizaki, 2023). Corresponding spring forces were then calculated for each muscle group, using spring constant (k) and displacement of each spring to replicate the resulting neck strength in Newtons. Each muscle stiffness had three pairs of springs attached from the head form to the neck plate. The headform was equipped with a CCM hockey helmet Vector 8, a certified helmet worn by professional ice hockey players. A new helmet was used for each neck stiffness condition to reduce variance in head impact results based on helmet performance (Post et al., 2014). The Hybrid III headform and uONSA neck plate weighed a total of 36.0 lbs.

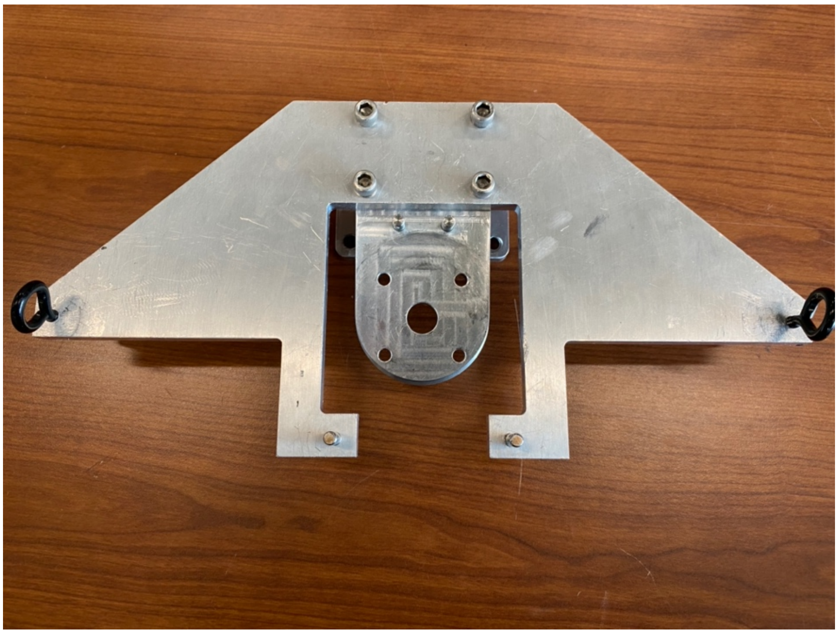

Figure 6: *uONSA Neck Plate*

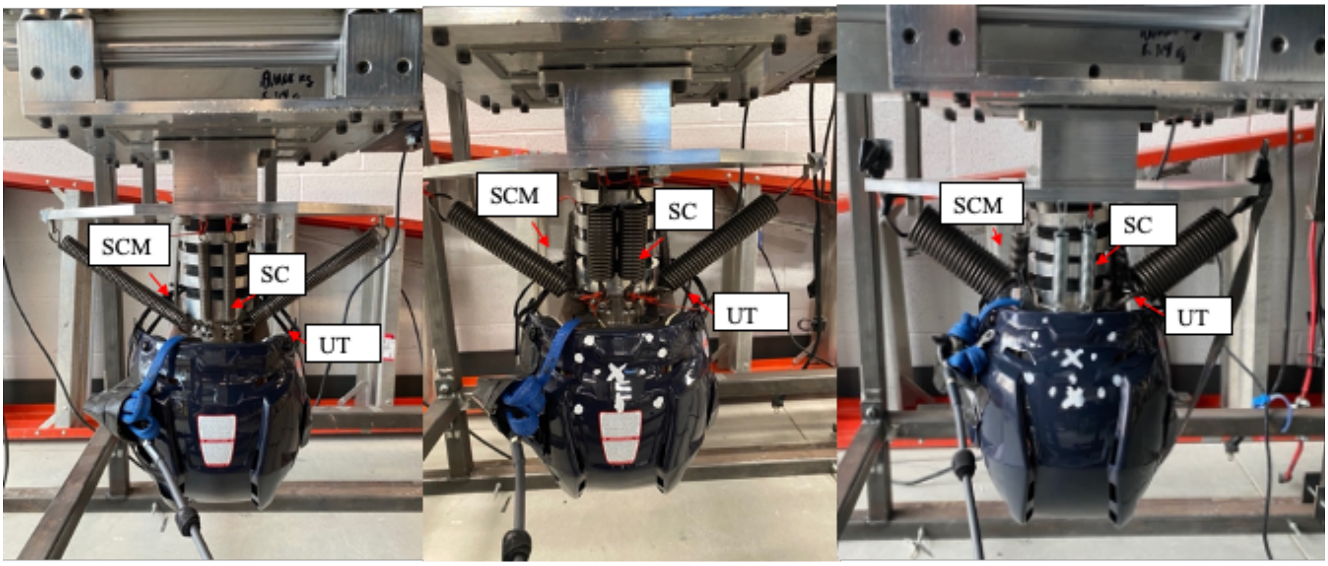

Figure 7: *Springs attached to the uONSA neck plate and base of the neckform representing three muscle groups of Sternocleidomastoid (SCM), Splenius Capitis (SC), and Upper Trapezius (UT). Different size springs were used to produce three neck tensions representing Low (left), medium (center), and high (right) neck stiffness.*

3.3.3 High-Speed Impactor

A high-speed impactor was used to replicate body-to-glass impacts at three velocities (3m/s, 5m/s and 7m/s). The head and uONSA neck apparatus and the neck plate were affixed to the impactor and positioned in an inverted posture (Fig, 7). A shoulder bumper, representing a human shoulder, composed of medium-density (97kg/m^3) vinyl foam 602 6cm thick base and a 4cm thick shoulder cap, was fastened to the end of the rails, acting as a stopper for the head, to simulate a body-first impact when a shoulder made first contact before the head impact. A 1.5-inch-thick polycarbonate material, representing rink glass, was fastened to the outside of the impactor's rail, situated 0.15 meters from the shoulder bumper to represent the distance between the head and glass after the shoulder first makes contact with the glass, as established from Kinovea video analysis.

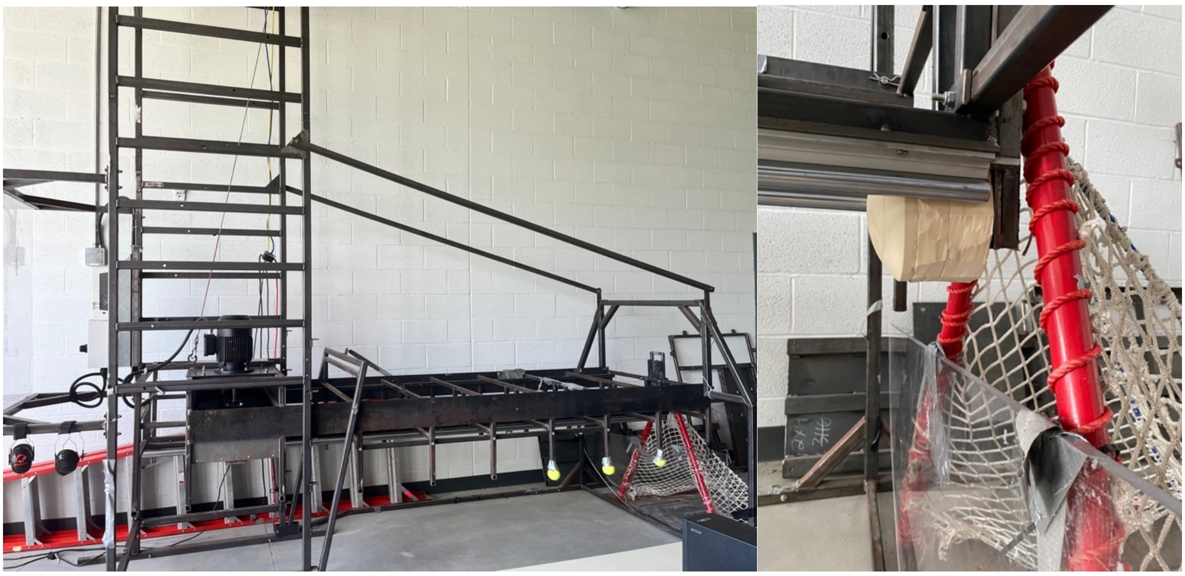

Figure 8: *Impact equipment showing (left) rail guided launcher, and (right) vinyl nitrile shoulder bumper fastened to the end of rails and sheet of polycarbonate material fastened outside the rails.*

3.4 Impact Procedure

Upon acceleration by the impactor by engagement of the wheels and motor, the head and neck assembly were released from the motor, travelling down the rails where it contacted the shoulder bumper, simulating the initial shoulder-to-glass impact. Subsequently, the headform experienced a head-to-glass impact, capturing the entire event from initial shoulder impact to head impact against the glass. A High-speed camera capturing at a rate of 1000 fps documented each impact for subsequent frame-by-frame analysis. Three trials of each impact were conducted to ensure consistency and no outliers for statistical analysis. To replicate neck positioning for each trial, the head angle and displacement were measured at the beginning of each trial to ensure the correct positioning and prevent extension or flexion of the neck (Cournoyer et al., 2021).

Figure 9: *Frame-by-frame analysis of an impact at 5m/s with medium neck stiffness*

3.4.1 Data Collection

Trials were conducted across three shoulder impact velocities (low: 3m/s, medium: 5m/s, high: 7m/s) and three neck muscle stiffness conditions low (60, 54.5 and 59.75N), medium (180, 163.5 and 179.25N), and high (240, 218, 239N) groups. Rotational acceleration was obtained from the time derivative of the measured rotational velocity time histories.

3.5 Statistical Analysis

Statistical analysis was conducted using JASP 0.17, with a significance level set at $p < 0.05$. To compare the difference between neck stiffness and velocity, three 3x3 ANOVA tests were conducted for peak linear acceleration, peak rotational acceleration and peak rotational velocity. A Tukey posthoc test was then run for each ANOVA run to determine whether significant differences existed between 3m/s, 5m/s and 7m/s velocity and low, medium and high neck stiffness. For statistical analysis, the highest of the two impact peaks was used across all conditions.

CHAPTER 4: RESULTS

4.1 Head Impacts

A total of 27 head impacts were collected: three at a low-neck stiffness using 3m/s, 5m/s and 7m/s velocity, three at a medium-neck stiffness using 3m/s, 5m/s and 7m/s impact velocity, and three at a high-neck stiffness using 3m/s, 5m/s and 7m/s impact velocity. Of the 27 shoulder impacts, six impacts with high neck stiffness and medium neck stiffness at 3m/s did not result in a head impact as the head did not impact the glass after the shoulder impact. All 27 impacts were included in the statistical analysis.

The mean peak linear and rotational acceleration values and rotational velocity were calculated for each impact velocity (3m/s, 5m/s, and 7m/s) and across levels of neck stiffness (low, medium, and high) is summarized in Table 1. Not all peak responses were taken from the same impact condition; for the majority of events, the shoulder contact resulted in the highest peak. However, as noted by an Asterix (*) in Table 1, there were five conditions where the head impact on the glass showed the highest peak and was recorded as the kinematic response.

***Table 1**. Averages for all dynamic responses for reconstructed events (brackets indicating standard deviation)*

		Peak Resultant Acceleration		
Velocity (m/s)	Neck Stiffness	Linear (g)	Rotational (rad/s2)	Rotational Velocity (rad/s)
3				
	Low	21.40 (3.21)*	1858.67 (205.87)	22.17 (2.65)
	Medium	10.98 (1.25)	1542.00 (145.36)	18.95 (1.56)
	High	10.03 (0.32)	1336.63 (16.84)	19.27 (1.36)
5				
	Low	55.17 (1.721)	4842.50 (253.88)*	43.12 (0.32)
	Medium	55.37 (1.01)	4446.80 (295.72)	34.27 (2.42)
	High	40.20 (6.61)	2613.93 (168.41)*	29.87 (0.91)
7				
	Low	76.13 (2.35)*	4131.07 (279.17)*	23.0 (1.15)
	Medium	94.40 (2.87)	7121.27 (822.28)	50.80 (0.66)
	High	64.30 (0.95)	5514.13 (353.71)	37.10 (0.30)

4.2 Peak Linear Acceleration

Peak linear accelerations are presented in Figure 10. Impacts for the 3m/s velocity resulted in mean linear accelerations of 21.4 g, 10.98 g, and 10.03 g for low, medium, and high neck stiffness, respectively. At 5m/s, the mean linear acceleration was 55.12 g, 55.37 g, and 40.20 g for low, medium, and high neck stiffness, respectively. For the highest velocity tested (7m/s), the mean linear accelerations were 76.13 g, 94.40 g, and 64.30 g for low, medium, and high neck stiffness, respectively.

4.3 Peak Rotational Acceleration

Figure 11 presents peak rotational acceleration results. The mean rotational acceleration (rad/s^2) at 3 m/s velocity

was 1858 rad/s^2, 1542 rad/s^2 and 1336 rad/s^2 for low, medium, and high neck stiffness, respectively.
The mean rotational acceleration at 5 m/s was 4842 rad/s^2, 4446. rad/s^2, and 2613 rad/s^2 for low, medium, and high neck stiffness, respectively. At 7 m/s velocity, the mean rotational accelerations were 4131 rad/s^2, 7121 rad/s^2, and 5514 rad/s^2 for low, medium, and high neck stiffness, respectively.

4.4 Rotational Velocity

Figure 12 presents rotational velocity results. For low, medium, and high neck stiffness, rotational velocity at 3m/s was 22.1 rad/s, 18.95 rad/s, and 19.27 rad/s. At 5m/s, rotational velocity was 43.1 rad/s, 34.27 rad/s, and 29.87 rad/s. At the highest velocity of 7m/s, rotational velocity for low, medium, and high neck stiffness was 23 rad/s, 50.80 rad/s, and 37.10 rad/s.

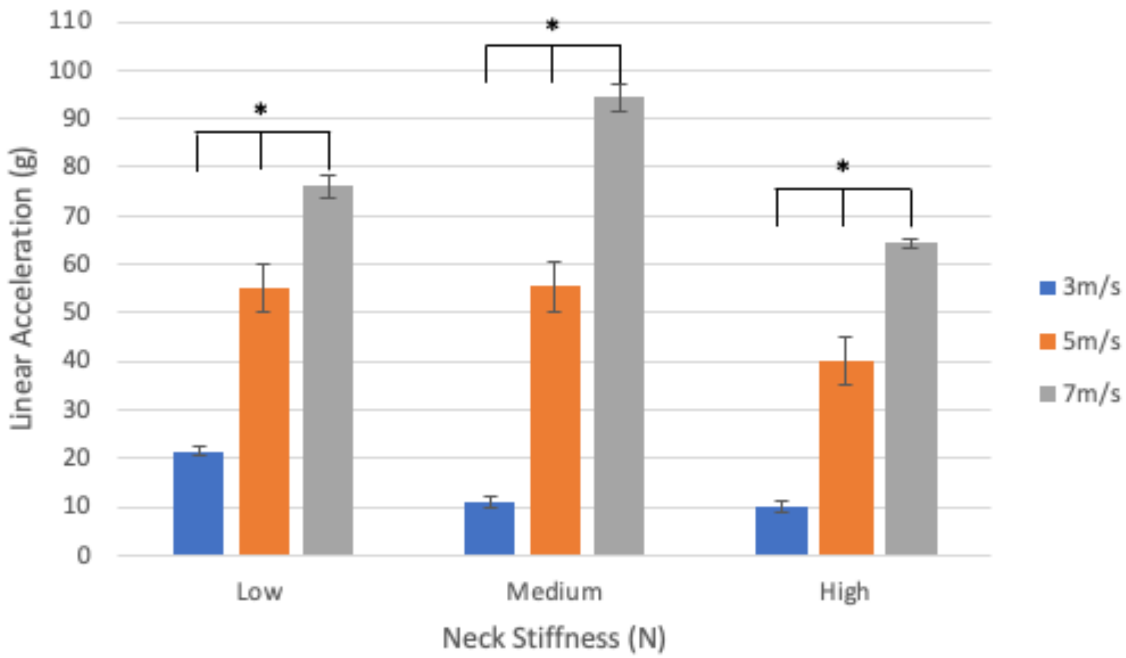

Figure 10: *Effect of neck stiffness on head linear accelerations from body-first impacts*
** denotes significance when comparing the effect of neck stiffness within each velocity level*

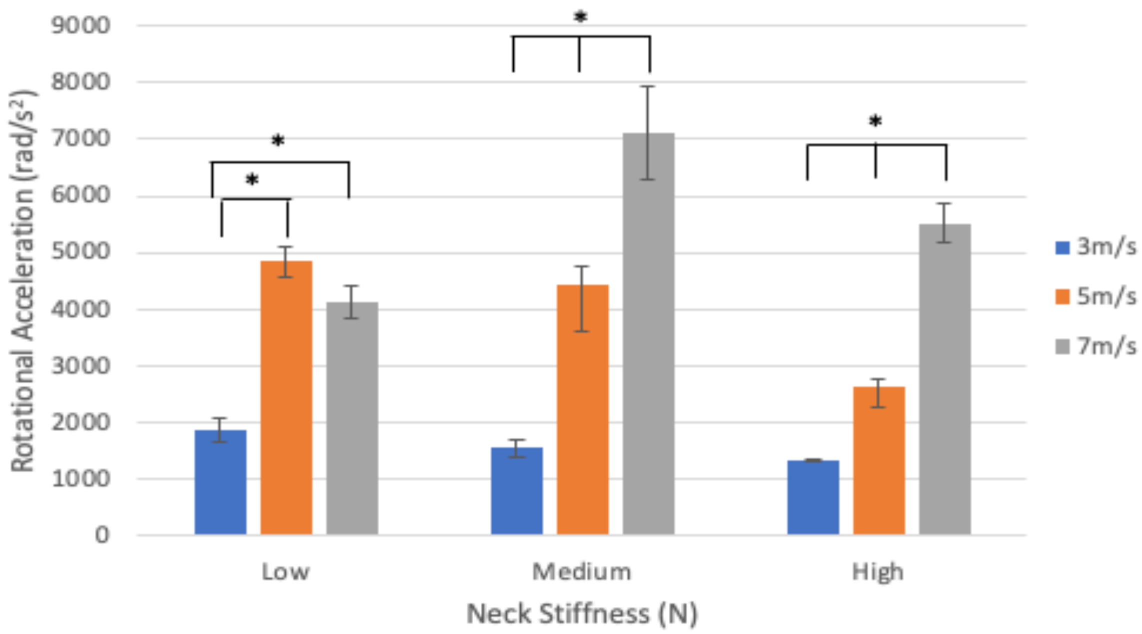

Figure 11: *Effect of neck stiffness on head rotational accelerations from body-first impacts*
** denotes significance when comparing the effect of neck stiffness within each velocity level*

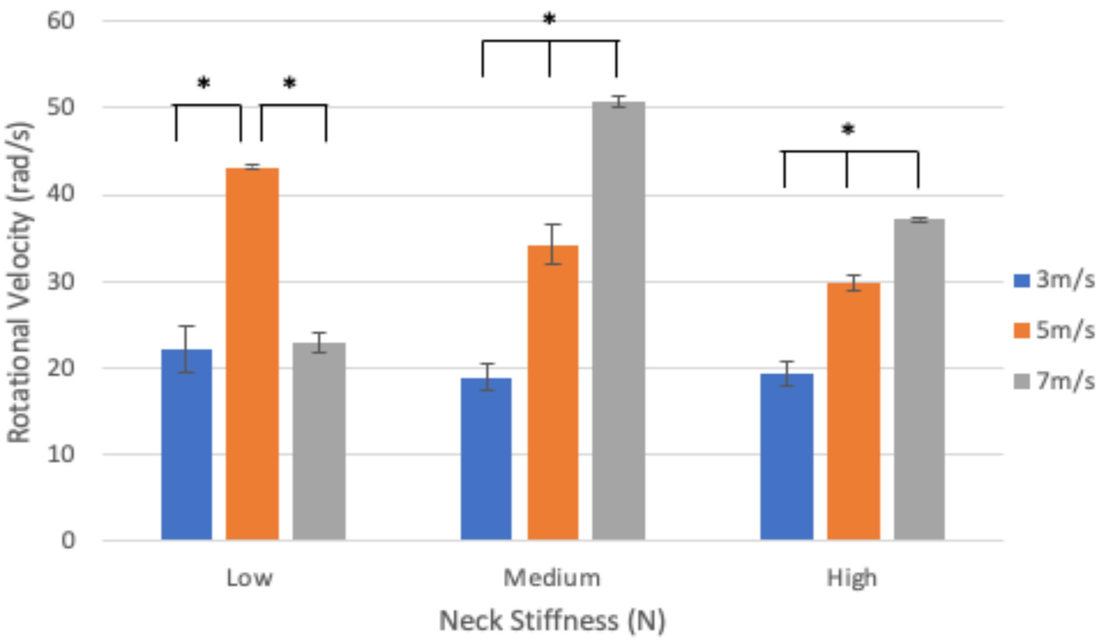

Figure 12: *Effect of neck stiffness on head rotational velocity from body-first impacts*
** denotes significance when comparing the effect of neck stiffness within each velocity level*

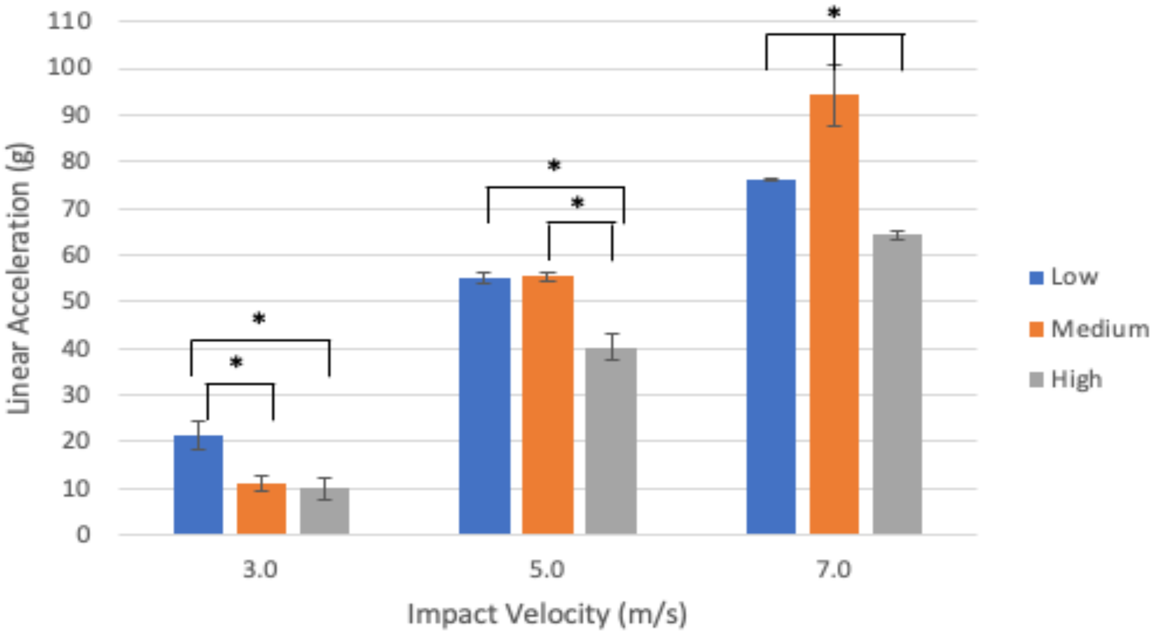

Figure 13: *Effect of impact velocity on head linear accelerations from body-first impacts*
** denotes significance when comparing the effect of velocity within each level of neck stiffness*

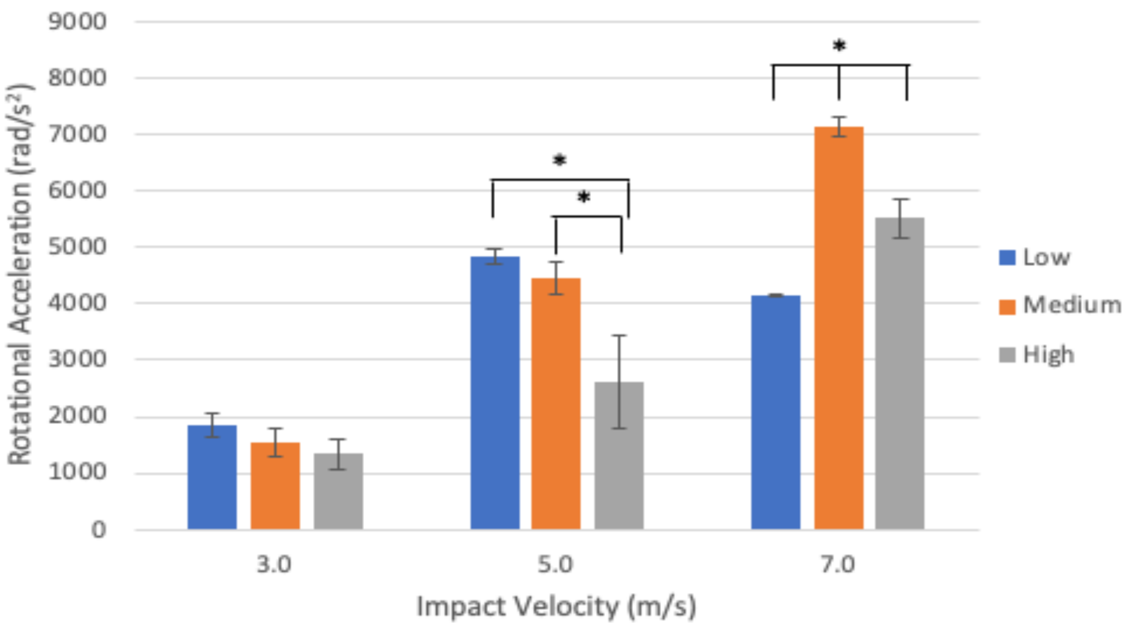

Figure 14: *Effect of impact velocity on head rotational accelerations from body-first impacts*
** denotes significance when comparing the effect of velocity within each level of neck stiffness*

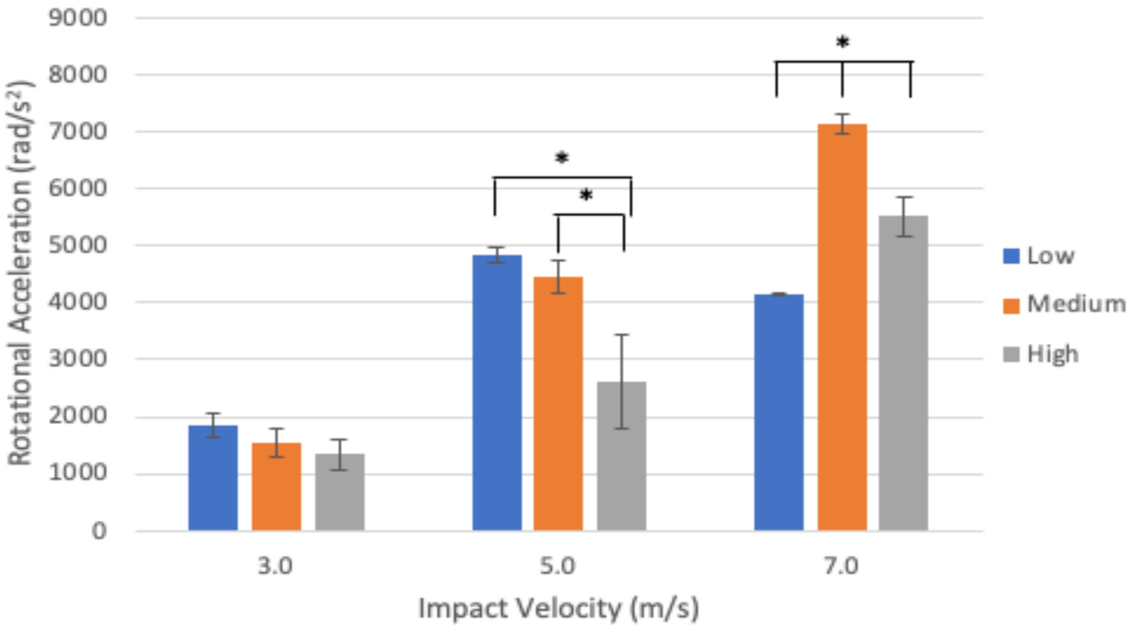

Figure 15: *Effect of impact velocity on head rotational velocity from body-first impacts*
** denotes significance when comparing the effect of velocity within each level of neck stiffness*

4.5 Two Impact Comparison

The reconstruction events created two dynamic responses for the head form, the first when the apparatus (shoulder) contacted the shoulder bumper and the second when the head contacted the glass (Fig, 28). High-speed video was used to break down each event frame by frame to correspond with the time of impact shown and the kinematic data to document each separate condition (shoulder or head impact). The highest of the two peaks was analyzed in this study, which typically resulted during the shoulder impact. Figures 13 -21 demonstrate the two event dynamic response curves for events occurring at all velocities (3,5,7 m/s) and neck stiffnesses (low, med, high). Events of the medium neck stiffness and high neck stiffness at the low velocity of 3m/s did not result in a second dynamic response of the head. These events were characterized by a dynamic response of the head form when the shoulder impacted the bumper but did result in the secondary head-to-glass impact. This occurred because the associated neck stiffness was too rigid to result in a secondary impact at low velocities, indicating an association between high neck stiffness and decreased head motion when body impacts occur at low velocities. Body-first head impacts often include two impacts; the body impact followed by the head impact. Separating the impact conditions by shoulder and head provides a better understanding of where the peak occurs during the event (Fig, 16-24). Dynamic response graphs for velocities of 3m/s were 200ms.

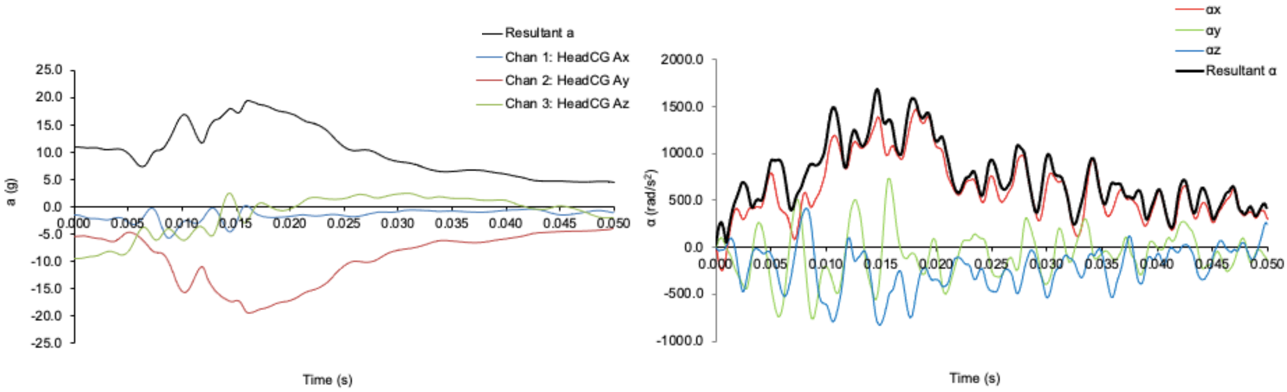

Figure 16: *Linear and rotational acceleration dynamic response curves for an impact occurring at 3m/s with low neck stiffness.*

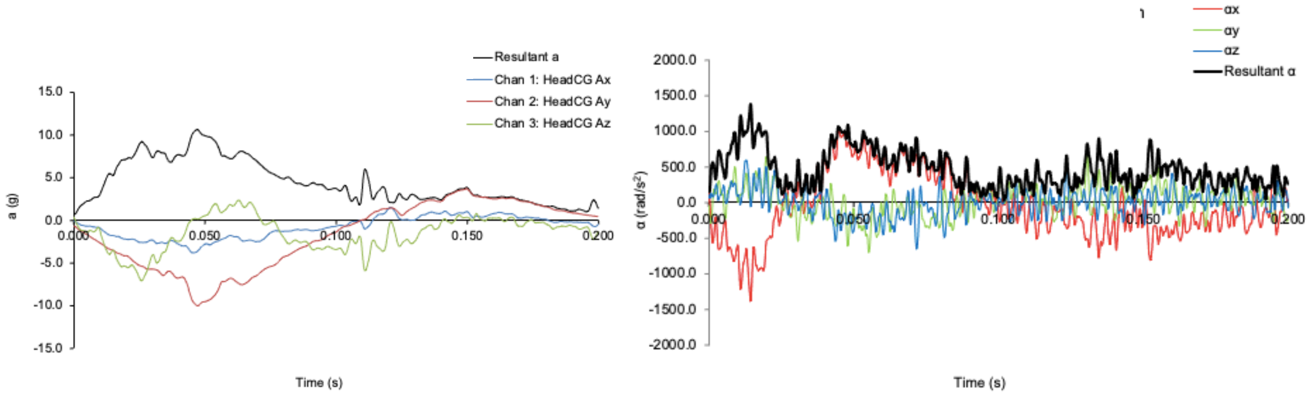

Figure 17*: Linear and rotational acceleration dynamic response curves for an impact occurring at 3m/s with medium neck stiffness*

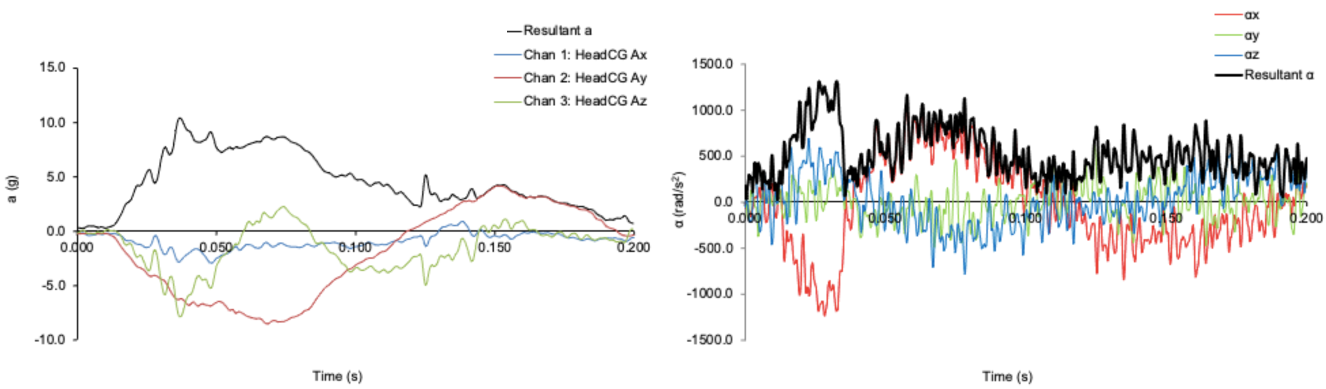

Figure 18: *Linear and rotational acceleration dynamic response cures for an impact occurring at 3m/s with high neck stiffness*

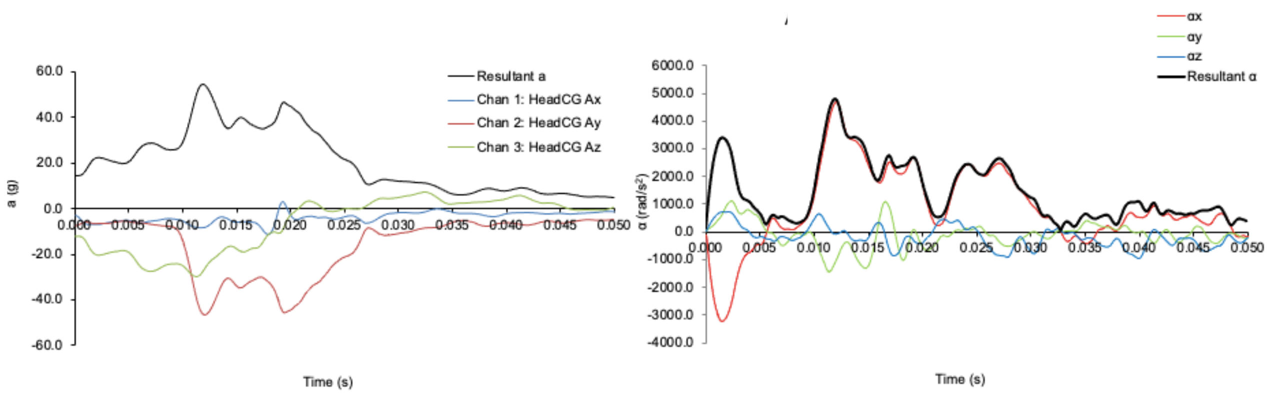

***Figure 19**: Linear and rotational acceleration dynamic response curves for an impact occurring at 5m/s with low neck stiffness*

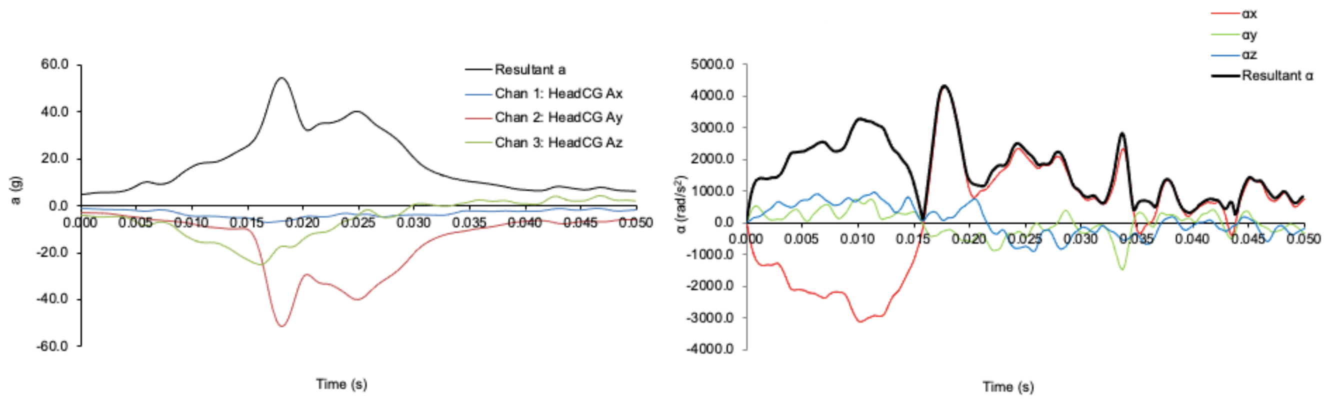

***Figure 20**: Linear and rotational acceleration dynamic response curves for an impact occurring at 5m/s with medium neck stiffness*

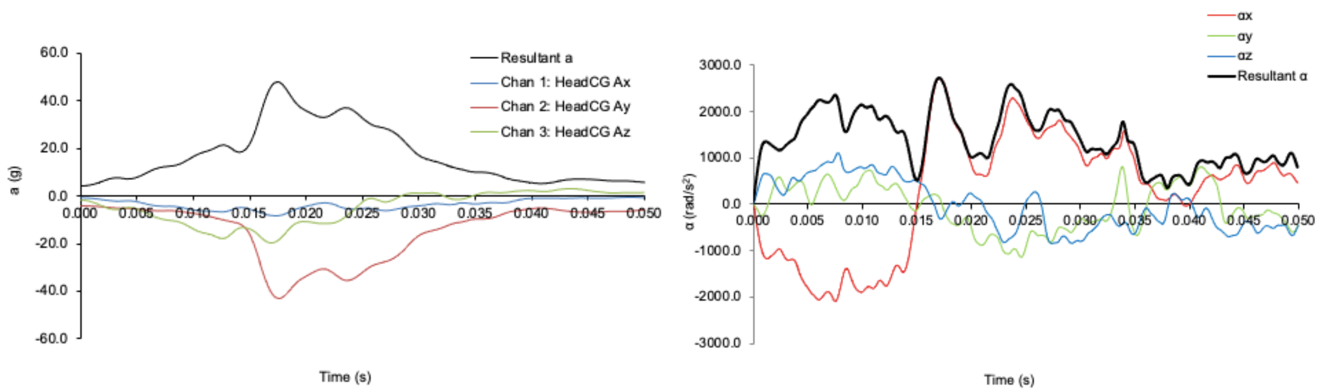

***Figure 21:** Linear and rotational acceleration dynamic response curves for an impact occurring at 5ms with high neck stiffness*

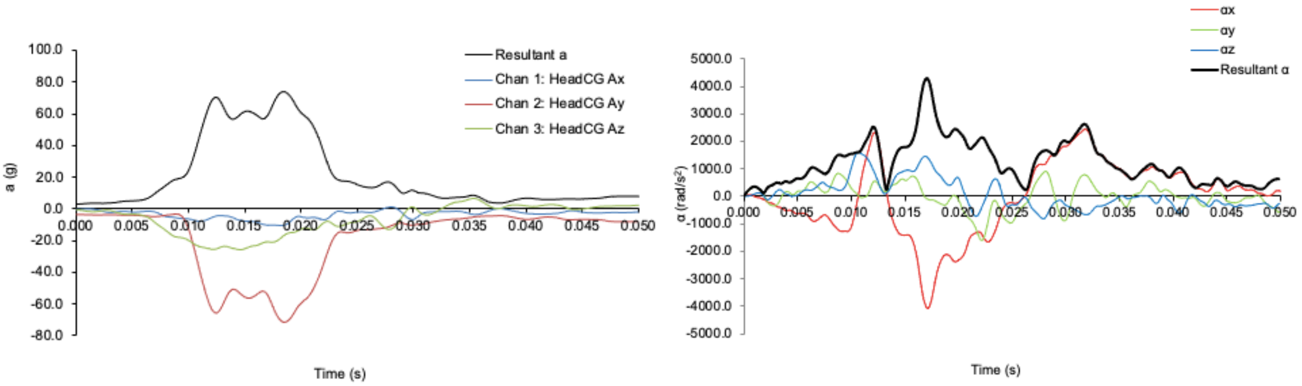

Figure 22: *Linear and rotational acceleration dynamic response curves for an impact occurring at 7m/s with low neck stiffness*

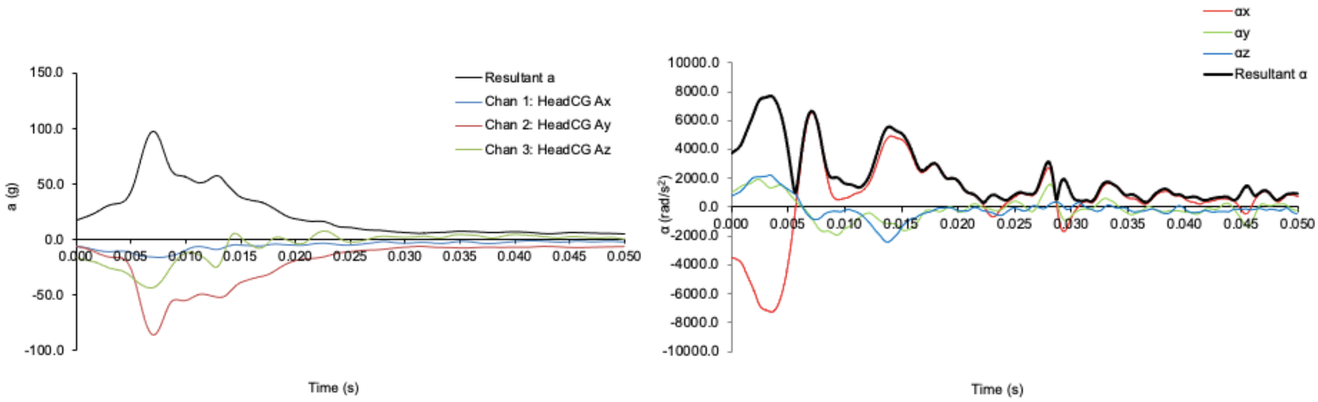

Figure 23*: Linear and rotational acceleration dynamic response curves for an impact occurring at 7m/s with medium neck stiffness*

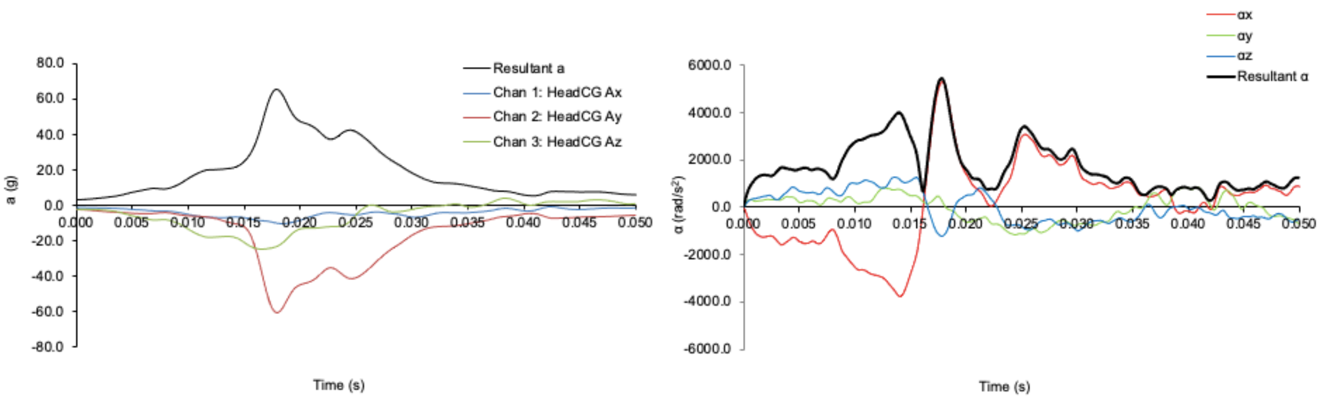

Figure 24: *Linear and rotational acceleration dynamic response curves for an impact occurring at 7m/s with high neck stiffness*

4.6 Duration

Impacts that occurred at a lower velocity (3m/s) resulted in long duration events greater than 50ms. Subsequently, the events that did not have a secondary impact, instead resulting in more of a whiplash-like motion, were greater than 80ms in duration. The dynamic response curve graphs were modified to display a 200ms impact curve for lower velocity events to incorporate the entire impact event.

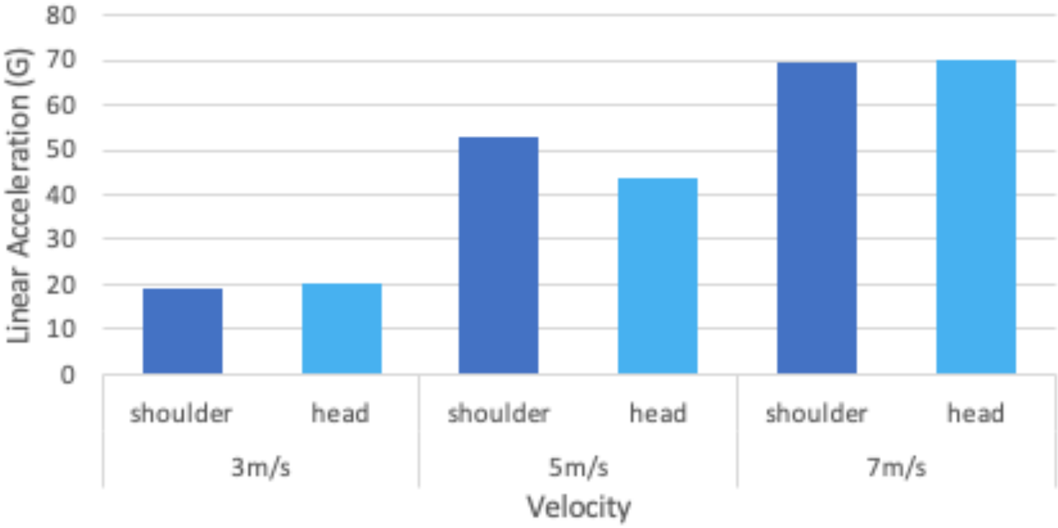

Figure 25: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for linear acceleration across three velocities (3, 5, 7m/s) for low neck stiffness*

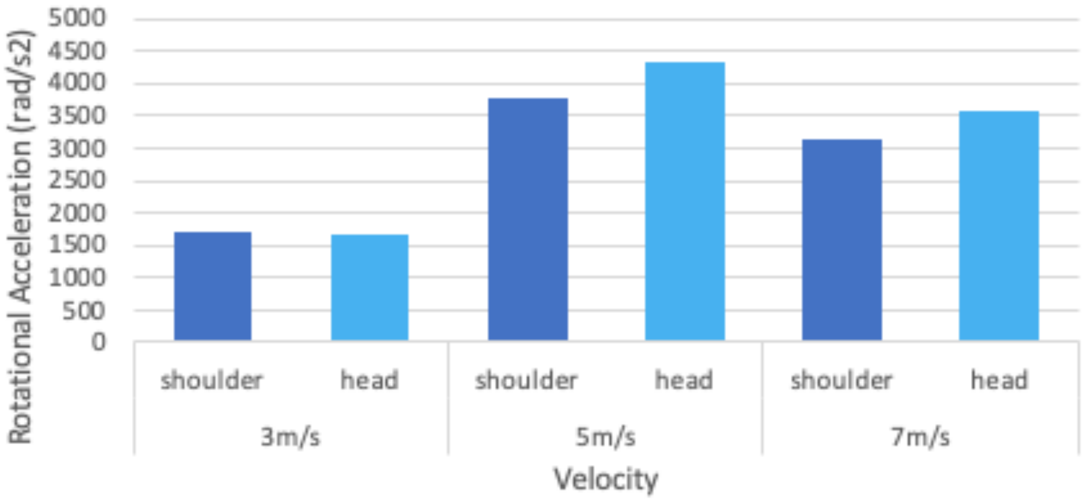

Figure 26: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for rotational acceleration across three velocities (3, 5, 7m/s) for low neck stiffness*

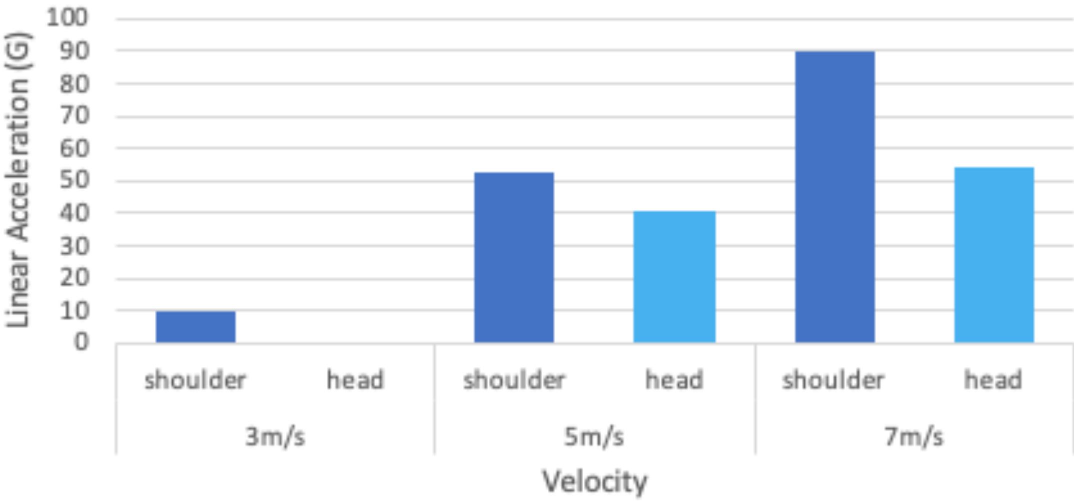

Figure 27: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for linear acceleration across three velocities (3, 5, 7m/s) for medium neck stiffness*

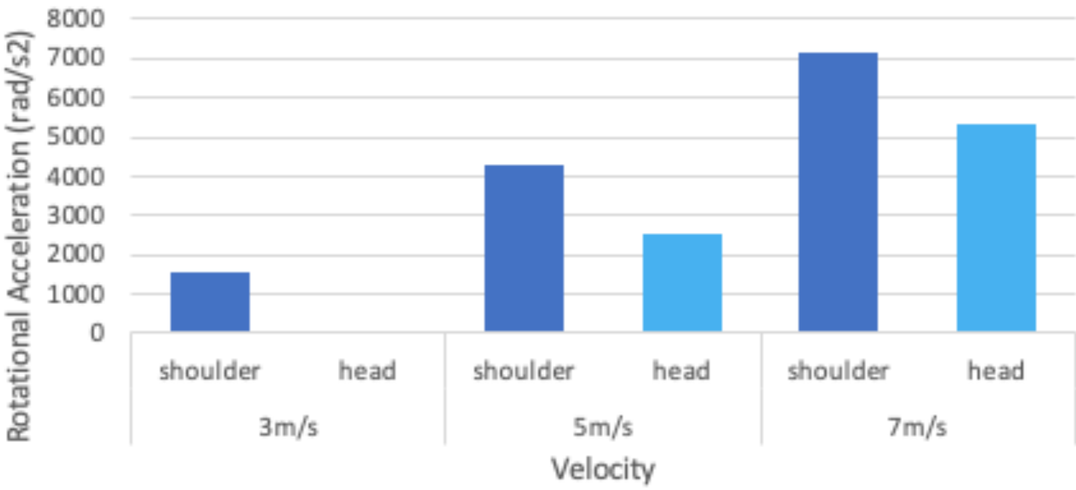

Figure 28: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for rotational acceleration across three velocities (3, 5, 7m/s) for medium neck stiffness*

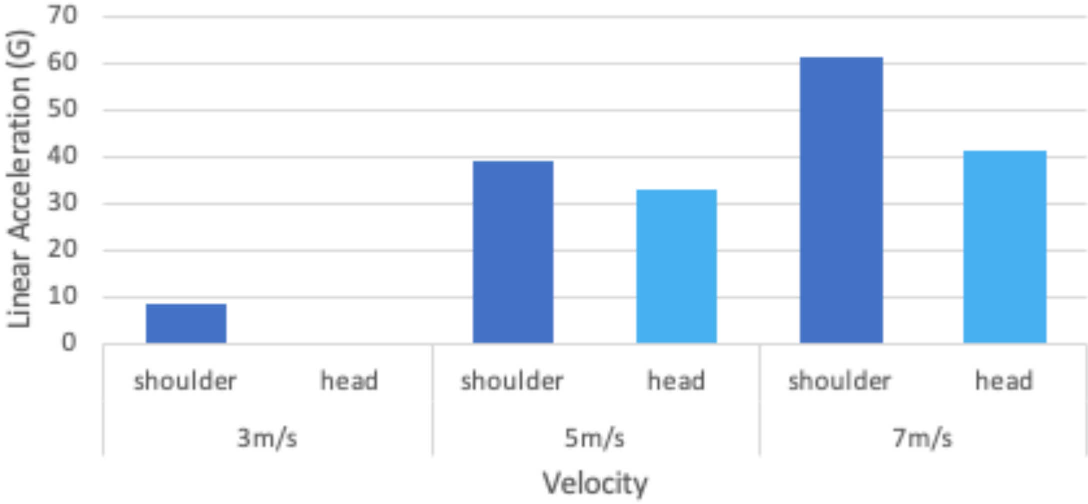

Figure 29: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for linear acceleration across three velocities (3, 5, 7m/s) for high neck stiffness*

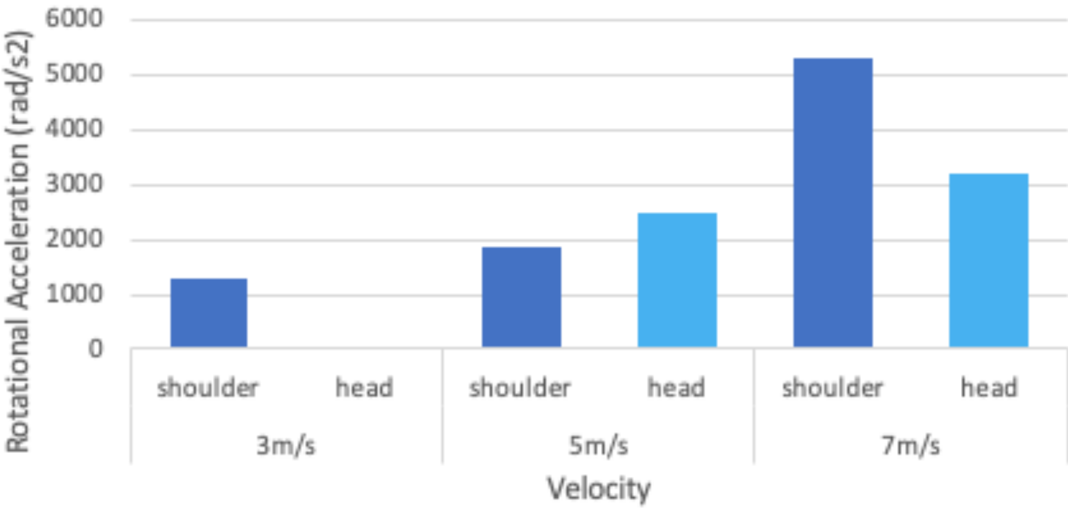

Figure 30: *Comparison of higher head kinematic impact responses depending on the event being a head or shoulder contact for rotational acceleration across three velocities (3, 5, 7m/s) for high neck stiffness*

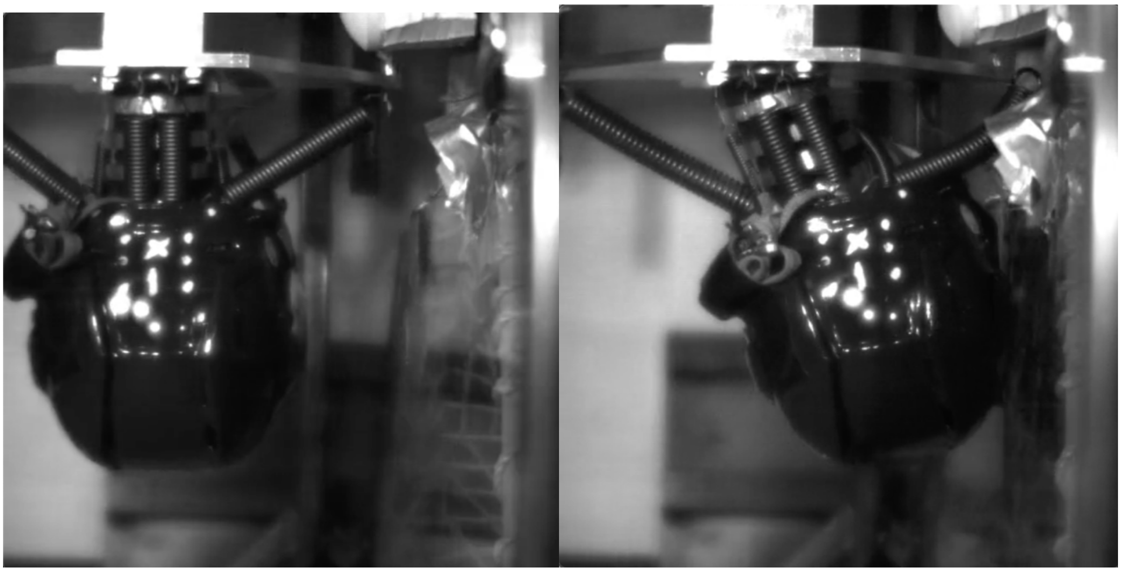

Figure 31: *Images from high-speed camera showing (left) the initial contact with shoulder bumper, followed by (right) head to glass contact. Sample image is from an impact at 5m/s using medium neck stiffness condition.*

4.7 Statistical Analysis

A two-way ANOVA was conducted to analyze the effects of neck stiffness and impact velocity on peak linear and rotational accelerations and rotational velocity during body-first impacts. The analysis revealed significant main effects for both neck stiffness and impact velocity across all dependent variables, indicating that these factors independently influenced peak accelerations ($p<0.001$). Specifically, the results showed that each 3x3 ANOVA test for the dependent variables yielded significant F values for linear acceleration ($F=25.7$, $p<0.001$), rotational acceleration ($F=30.729$, $p<0.001$) and rotational velocity ($F=152.20$, $p<0.001$). A one-way ANOVA was performed for each factor separately on each dependent variable to further examine the independent effects of impact velocity and neck stiffness. For impact velocity, results were significant across linear acceleration ($F=11227.06$, $p<0.001$), rotational acceleration ($F=290.737$, $p<0.001$) and rotational velocity ($F=378.187$, $p<0.001$). For neck stiffness, each dependent variable showed significant results as well: linear acceleration ($F= 74.085$, $p<0.001$), rotational acceleration ($F=28.002$, $p<0.001$), and rotational velocity ($F=50.271$, $p<0.001$).

A means comparison test further identified significant differences in levels of impact velocity for each dependent variable, with the exception of rotational velocity between 5m/s and 7m/s ($p=0.193$). In terms of neck stiffness, all levels were significantly different for each dependent variable, with the exception of linear acceleration between low and medium stiffness ($p=0.140$) and rotational velocity between high and low neck stiffness ($p=0.582$). Post-hoc analysis provided additional insights into the relationships between neck stiffness, impact velocity and peak accelerations. For linear acceleration, significant differences were observed across neck stiffness levels at various impact velocities. At 3m/s, low neck stiffness produced higher accelerations than both medium ($p=0.008$) and high stiffness ($p=0.003$); at 5m/s, high neck stiffness was not significantly different from medium ($p=1.000$), and no difference was observed with low stiffness; at 7m/s significant differences were noted between all neck stiffness levels ($p<0.001$). In terms of rotational acceleration, significant differences were found across neck stiffness levels at each impact velocity, with the exception of low-neck stiffness at 5m/s and 7m/s ($p=0.309$). At 3m/s, no differences were observed in rotational acceleration across neck stiffness levels, while at 5m/s, high neck stiffness produced lower rotational accelerations compared to medium and low stiffness; at 7m/s, each neck stiffness level yielded significantly different results. For rotational velocity, significant differences were seen between low neck stiffness at 3m/s and 5m/s, though not at 7m/s ($p=0.998$), with results at 5m/s being significantly higher than at 7m/s. Differences within medium and high neck stiffness across each velocity level were also significant ($p<0.001$), with no differences reported at 3m/s across neck stiffness levels. In summary, post-hoc tests revealed that both neck stiffness and impact velocity independently influenced peak accelerations.

CHAPTER 5: DISCUSSION

The findings in this study describe the biomechanical responses associated with levels of neck stiffness across different impact velocities, specifically body-first shoulder-to-glass impacts. The results represent significant main effects for both linear and rotational accelerations ($p<0.001$) for neck stiffness and impact velocity. These results demonstrate that both independent variables contributed to the magnitude of peak linear and rotational accelerations experienced during body-first head impacts. The findings are consistent with previous research describing the role of these biomechanical factors in head injury dynamics (Schmidt et al., 2014; Smith et al., 2015). Schmidt and colleagues (2014) concluded that those with stronger neck muscles did not demonstrate decreased head injury severity risks. This is consistent with the current findings demonstrating that higher neck strength may not necessarily mitigate risk when experiencing impacts at high velocities. High neck stiffness reveals a decrease in rotational acceleration, suggesting a potential trade-off between linear and rotational acceleration. Smith and colleagues (2015) noted that in events where a combination loading occurs, mimicking the body-first event in this study, there are two specific time periods where the brain undergoes a high period of strain; the first period occurs when the head makes direct contact and the second when the head is rebounding in the opposite direction. This phenomenon applies to the results from this study where impacts at lower velocities but higher neck stiffness experienced an initial peak in both linear and rotational acceleration upon the shoulder impact, in this case, not a direct head impact, but also a peak when the head is rebounding in the opposite direction as the head does not impact the glass.

The effect of neck stiffness and impact velocity on head acceleration during body first impacts played a significant role in determining the severity of head impacts. Higher neck stiffness was associated with lower peak linear and rotational acceleration values at 3m/s impact velocity, suggesting a potential protective effect. However, as impact velocity increased to 5m/s and 7m/s, a corresponding increase in acceleration values was observed across all neck stiffness levels, albeit with variations in magnitude.

Differences in peak linear acceleration were found for neck stiffness across velocities. The results revealed that as impact velocity increased, a higher linear acceleration resulted for each of the three neck stiffnesses tested. With increased neck stiffness, a decrease in linear acceleration was observed across velocities, indicating that greater neck tension could affect linear acceleration. One exception was found where higher linear acceleration was reported with a medium stiff neck compared to a high stiff neck at 7m/s. These results suggest that higher neck stiffness generally mitigates linear acceleration, corroborating previous findings (Schmidt et al., 2014), potentially reducing the risk of severe brain injury such as hematoma or skull fracture. These results also demonstrated that at the highest velocity, a medium stiff neck may not be strong enough to mitigate a substantial amount of linear acceleration, therefore not necessarily reducing the risk of injury. High levels of linear acceleration cause pressure gradients to change within the brain, leading to more focal injuries such as subdural hematomas (Gurdjian et al., 1966). At greater impact velocities (7m/s), all impacts presented linear acceleration values either close to meeting or greater than those associated with a 25% risk of brain tissue injury (Zhang et al., 2004). Medium neck stiffness resulted in 94.4g of linear acceleration, exceeding the threshold for a 50% risk of brain injury and approximating the proposed thresholds, reflecting an 80% risk of injury. Similarly, decreases were found in rotational acceleration with an increase in neck stiffness, with the exception of medium neck stiffness at the highest velocity of 7m/s, again falling within the 50% risk category of sustaining injury, however considerably closer to meeting the threshold of 80% risk (7900 rad/s^2) (Zhang et al., 2004) with a reported value of 7121.27 rad/s2.

Rotational components of acceleration and velocity are closely associated with concussions (Yoganandan et al., 2008; Post et al., 2017; Post et al., 2015; Post & Hoshizaki, 2012), and our results indicate that neck stiffness can also decrease these accelerations. However, the relationship is complex, as higher stiffness at lower velocities did not consistently result in lower rotational head responses. These findings revealed increased neck stiffness was consistent with reduced peak linear and rotational acceleration (Schmidt et al., 2014). However, the relationship between neck stiffness and impact velocity may be more complex. For example, at higher velocities, even with high neck stiffness, rotational accelerations were higher than the low-neck stiffness condition, highlighting the limitations of neck stiffness alone in mitigating injury risk. Several other factors in injury can be considered to understand the influence of neck stiffness, including impact angle, head mass and system mass. These findings are supported by those of Cournoyer and colleagues (2021), who discuss the ability of neck musculature to decrease brain injury and observed that this relationship is complex and depends on impact conditions. They noted a statistically significant increase in peak linear acceleration at side impact locations at 6m/s and peak rotational acceleration at 4 and 6m/s compared to no neck musculature attachment at 3.5kg striking mass. Additionally, with a 5kg striking mass, the neck with spring attachments showed significantly greater rotational acceleration at 6m/s (Cournoyer et al., 2021). This emphasizes that the relationship between head injury and neck musculature is more complex at different impact conditions and higher velocities.

Rousseau and colleagues (2010) conducted research using three different necks, a soft neckform, a normal neckform and a stiff neckform. All impacts were collected with a 5m/s impact velocity. The results demonstrated that as neck stiffness increased, linear acceleration also increased, however only slightly. On the contrary, rotational acceleration decreased as neck stiffness increased. The results from the current study demonstrate similar trends for rotational acceleration at 5m/s, with a decrease in rotational acceleration as neck stiffness increased; however, for linear acceleration, there was a decrease between the medium and high neck stiffness. The differences shown between soft and regular neckform stiffness for linear acceleration for Rousseau et al. (2010) were marginal (78.4 g for soft neckform, 79.1g for normal neckform), which reflects no difference in linear acceleration between low (55.17g) and medium (55.37g) neck stiffness for this study.

The increase in head impact kinematics was proportional to the rise in impact velocity, resulting in unique load demands corresponding to varying levels of neck stiffness. The peak linear acceleration displayed an extended response to changes in head impact kinematics at different impact velocities. In terms of rotational acceleration and velocity, comparable outcomes were also described, except for the low-neck stiffness condition, where higher impact velocity led to reduced rotational peak resultant values compared to other levels of neck stiffness. An interaction between neck stiffness and impact velocity may have influenced impact mechanics and warrants further investigation. Because of the functionality of the unbiased neck having rubber discs in between aluminum discs, the movements of each individual disc occurring with each reconstruction variable could provide valuable data to describe the functionality of the neck under different impact conditions. High-speed video footage from this data describes the movements of the unbiased neck with each impact. These results demonstrated that neck stiffness and impact velocity had individual effects on head kinematics during body first impacts, emphasizing the importance of considering neck stiffness as a biomechanical variable in measuring kinematic responses.

Impact events at 3m/s with high and medium neck stiffness did not result in two impacts but only the shoulder impact. After the initial shoulder impact, these events displayed a whiplash-like motion; the head quickly travelled in the other direction. These findings are consistent with those reported by Smith et al. (2015), who found that the brain underwent two periods of strain during indirect impacts. They described the first period occurring when the head made contact with the anvil and the second when the helmeted headform rebounded in the opposite direction. Although the impact in this study did not take into account strain, the kinematics displayed an initial peak when the headform and neck apparatus made contact with the shoulder and due to the lack of the head impact along the glass, the head and neck apparatus began moving in the opposite direction.

5.1 Limitations

While this study contributes insights concerning the role of neck musculature in mitigating impact severity during body-first head impacts, limitations should be acknowledged. Potential differences between laboratory simulations and real-world scenarios, where in real life, the head and body should be moving simultaneously, while the laboratory reconstructions only accounted for the head and neck movements and may have presented displacement of the head compared to a real-time head-to-glass impact. Although standardized, the use of the 50th percentile male headform is made of metal and rubber, not identical to a human biofidelic neck which could create some discrepancy in the dynamic response values of a real-life event. The neck apparatus mimicked the muscle lines of action of three neck muscle groups, only partially replicating the complex neck neuromuscular responses. Additionally, the springs attached to the neck apparatus may not have represented the complete biomechanical movements of a human neck. Due to the limited function of extension springs, once attached to the uONSA apparatus, the actual spring forces throughout an impact may not have been identical to the reported stiffness values.

5.2 Future Research

The findings have significant implications for injury prevention strategies in ice hockey. Understanding the biomechanical factors influencing head injury risk, such as neck stiffness and impact velocity, can inform targeted interventions to enhance player safety. Further research is needed to contribute to the design and function of a neck apparatus to have a more biofidelic apparatus that provides increased validity for laboratory reconstructions. Individual variability in neck muscle activation patterns could also influence the results, suggesting the need for further research to validate and extend these findings. Training and athletic staff can utilize this data to ensure athletes exposed to higher velocity impacts have neck strengthening built into training programs to decrease injury risk. Neck strength is an important factor for enhancing athletic performance and injury prevention (Gysland et al., 2012), therefore developing a more comprehensive model of the neck that describes the more complex characteristics of the 20-plus pairs of cervical muscles (Eckner et al., 2014) would contribute to further research.

This study provided insight for future research designed to understand better the effect of neck stiffness on mitigating the risk of head injury in body-first impacts.

CHAPTER 6: CONCLUSION

This research investigated the role of cervical musculature in modulating head impact severity during body-first impacts. The results revealed the effect of increased neck stiffness on peak linear and rotational head acceleration. Impacts for the 3m/s velocity significantly decreased peak linear accelerations across low, medium, and high neck stiffness. Peak rotational acceleration (rad/s^2) at 3 m/s impact velocity decreased significantly across low, medium, and high neck stiffness. At 5m/s, the mean linear acceleration (g) also decreased across low, medium, and high neck stiffness.

Similarly, at this velocity, mean rotational acceleration (rad/s^2) decreased significantly across medium and high neck stiffness. For the highest velocity tested (7m/s), the peak linear accelerations (g) results were inconsistent across low, medium, and high neck stiffness. The peak rotational accelerations (rad/s^2) were also inconsistent across low, medium, and high neck stiffness. This research involved head-to-glass impact events, a head injury mechanism in ice hockey, highlighting the need for targeted interventions and injury prevention strategies to address specific injury mechanisms. This research provides insights into the biomechanical responses associated with varying levels of neck stiffness for body-first head impacts involving the shoulder at different velocities and neck stiffnesses, contributing to a better understanding of head and neck injury mechanisms in ice hockey injuries. Future research should consider the continued development of a biofidelic neck to support additional research investigating the relationship between neck muscles and head impacts for body-first impacts. Overall, this study contributes to understanding head injury mechanisms in ice hockey and provides insights for developing effective injury prevention strategies. Further research in this area is warranted to elucidate the complex interplay between biomechanical factors and head injury risk, ultimately improving the safety and well-being of athletes participating in ice hockey.

REFERENCES:

Aguiar, O. M. G., Potvin, B. M., Yang, Y., Hua, K. N., Bruschetta, M. L., Virani, S., & Robinovitch, S. N. (2020). American Society of Biomechanics Journal of Biomechanics Award 2019: Circumstances of head impacts in Men's University Ice hockey. *Journal of Biomechanics, 108*.

Aguiar, O.M.G., Chow, T.R., Chong, H., Vakili, O., Robinovitch, S.N. (2023). Associations between the circumstances and severity of head impacts in men's university ice hockey. *Scientific Reports, 13*. 17402.

Andrews, E., Jildeh, T. R., Abbas, M. J., Lindsay-Rivera, K., Berguson, J., & Okoroha, K. R. (2022). Concussions in the National Hockey League Analysis of Incidence, Return to Play, and Performance. Orthopaedic *Journal of Sports Medicine, 10*(1).

Biasca, N., Wirth, S., Tegner, Y. (2002). The avoidability of head and neck injuries in ice hockey: A historical review. *British Journal of Sports Medicine, 36*(6), 410-427

Black, A. M., Hagel, B. E., Palacios-Derflingher, L., Schneider, K. J., & Emery, C. A. (2017). The risk of injury associated with body checking among Pee Wee ice hockey players: an evaluation of Hockey Canada's national body checking policy change. *British Journal of Sports Medicine, 51*(24), 1767–1772.

Bogduk, N., & Mercer, S. (2000). Biomechanics of the cervical spine. I: Normal kinematics. *Clinical Biomechanics*, 633–648.

Broglio, S. P., Eckner, J. T., Paulson, H. L., & Kutcher, J. S. (2012). Cognitive Decline and Aging. *Exercise and Sport Sciences Reviews, 40*(3), 138–144.

Choi, W. J., Robinovitch, S. N., Ross, S. A., Phan, J., & Cipriani, D. (2017). Effect of neck flexor muscle activation on impact velocity of the head during backward falls in young adults. *Clinical Biomechanics, 49*, 28–33.

Clark, J.M., Taylor, K., Post, A., Hoshizaki, T.B., Gilchrist, M.D. (2018). Comparison of ice hockey goaltender helmets for concussion type impacts. Annals of Biomedical Engineering, 46(7), 986–1000.

Collins, C. L., Fletcher, E. N., Fields, S. K., Kluchurosky, L., Rohrkemper, M. K., Comstock, R. D., & Cantu, R. C. (2014). Neck Strength: A Protective Factor Reducing Risk for Concussion in High School Sports. *The Journal of Primary Prevention, 35*(5), 309–319.

Cournoyer, J., Koncan, D., Gilchrist, M. D., & Blaine Hoshizaki, T. (2021). The influence of neck stiffness on head kinematics and maximum principal strain associated with youth American football collisions. *Journal of Applied Biomechanics, 37*(3), 288–295.

Crandall, J. R. (2002). Designing road vehicles for pedestrian protection. *BMJ, 324*(7346), 1145–1148.

Cronin, D. S., Singh, D., Gierczycka, D., Barker, J., & Shen, D. (2018). Modelling the Neck for Impact Scenarios. In Basic Finite Element Method as Applied to Injury Biomechanics, 503–538. Elsevier.

Cusimano, M. D., Taback, N. A., McFaull, S. R., Hodgins, R., Bekele, T. M., Elfeki, N., & Canadian Research Team in Traumatic Brain Injury and Violence. (2011). Effect of bodychecking on rate of injuries among minor hockey players. *Open Medicine : A Peer-Reviewed, Independent, Open-Access Journal, 5*(1), e57-64.

Echlin, P. S., Tator, C. H., Cusimano, M. D., Cantu, R. C., Taunton, J. E., Upshur, R. E. G., Hall, C. R., Johnson, A. M., Forwell, L. A., & Skopelja, E. N. (2010). A prospective study of physician-observed concussions during junior ice hockey: implications for incidence rates. Neurosurgical Focus, 29(5), E4.

Eckner, J. T., Oh, Y. K., Joshi, M. S., Richardson, J. K., & Ashton-Miller, J. A. (2014). Effect of neck muscle strength and anticipatory cervical muscle activation on the kinematic response of the head to impulsive loads. *American Journal of Sports Medicine, 42*(3), 566–576.

Fice, J. B., Gunter, X., Siegmund, P., & Blouin, J.-S. (2018). Neck muscle biomechanics and neural control. *J Neurophysiol, 120*, 361–371.

Fleisig, G. S., Barrentine, S.W., Escamilla, R.F., Andrews, J.R. (1996). Biomechanics of overhand throwing with implications for injuries. *Sports Medicine, 21*(6), 421- 437.

Gabler, L., Patton, D., Begonia, M., Daniel, R., Rezaei, A., Huber, C., Siegmund, G., Rooks, T., & Wu, L. (2022). Consensus Head Acceleration Measurement Practices (CHAMP): Laboratory Validation of Wearable Head Kinematic Devices. *Annals of Biomedical Engineering, 50*(11), 1356–1371.

Gennarelli, T. A., & Thibault, L. E. (1982). Biomechanics of Acute Subdural Hematoma. *The Journal of Trauma: Injury, Infection, and Critical Care, 22*(8), 680–686.

Gurdjian, E. S., Roberts, V. L., & Thomas, L. M. (1966). Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. *The Journal of Trauma: Injury, Infection, and Critical Care, 6*(5), 600–604.

Guskiewicza, K. M., Marshall, S. W., Bailes, J., McCrea, M., Harding, H. P., Matthews, A., MihalikI, J. R., & Cantu, R. C. (2007). Recurrent Concussion and Risk of Depression in Retired Professional Football Players. *Medicine & Science in Sports & Exercise, 39*(6), 903–909.

Guskiewicz, K. M., McCrea, M., Marshall, S. W., Cantu, R. C., Randolph, C., Barr, W., Onate, J. A., & Kelly, J. P. (2003). Cumulative Effects Associated With Recurrent Concussion in Collegiate Football Players The NCAA Concussion Study. *JAMA, 290*(19), 2549–2555.

Gysland, S. M., Mihalik, J. P., Register-Mihalik, J. K., Trulock, S. C., Shields, E. W., & Guskiewicz, K. M. (2012). The Relationship Between Subconcussive Impacts and Concussion History on Clinical Measures of Neurologic Function in Collegiate Football Players. *Annals of Biomedical Engineering, 40*(1), 14–22.

Hajiaghamemar, M., Seidi, M., Ferguson, J. R., & Caccese, V. (2015). Measurement of Head Impact Due to Standing Fall in Adults Using Anthropomorphic Test Dummies. *Annals of Biomedical Engineering, 43*(9), 2143–2152.

Hardy, W. N., Foster, C. D., Mason, M. J., Yang, K. H., King, A. I., & Tashman, S. (2001, November 1). Investigation of Head Injury Mechanisms Using Neutral Density Technology and High-Speed Biplanar X-ray.

Hoshizaki, T. (2023). University of Waterloo. [Unpublished Data]

Hoshizaki, T.B., Brien, S.E. (2004). The science and design of head protection in sport. *Neurosurgery, 55*(4), 956–966.

Hoshizaki, T. B., Post, A., Kendall, M., Cournoyer, J., Rousseau, P., Gilchrist, M. D., Brien, S., Cusimano, M., & Marshall, S. (2017). The development of a threshold curve for the understanding of concussion in sport. *Trauma, 19*(3), 196–206.

Hoshizaki, T. B., Post, A., Oeur, R. A., & Brien, S. E. (2014). Current and Future Concepts in Helmet and Sports Injury Prevention. *Neurosurgery, 75*(Supplement 4), S136–S148.

Hutchison, M. G., Comper, P., Meeuwisse, W. H., & Echemendia, R. J. (2015a). A systematic video analysis of National Hockey League (NHL) concussions, part I: who, when, where and what? *British Journal of Sports Medicine, 49*(8), 547–551.

Hutchison, M. G., Comper, P., Meeuwisse, W. H., & Echemendia, R. J. (2015b). A systematic video analysis of National Hockey League (NHL) concussions, part II: how concussions occur in the NHL. *British Journal of Sports Medicine, 49*(8), 552–555.

Karton, C., & Blaine Hoshizaki, T. (2018). Concussive and subconcussive brain trauma: the complexity of impact biomechanics and injury risk in contact sport. *Handbook of Clinical Neurology, 158*, 39–49.

Karton, C., Post, A., Laflamme, Y., Kendall, M., Cournoyer, J., Robidoux, M. A., Gilchrist, M. D., & Hoshizaki, T. B. (2021). Exposure to brain trauma in six age divisions of minor ice hockey. *Journal of Biomechanics, 116*, 110203.

Kleiven, S. (2013). Why most traumatic brain injuries are not caused by linear acceleration but skull fractures are. *Frontiers in Bioengineering and Biotechnology, 1*, 1–15.

Kleiven, S. (2007). Predictors for traumatic brain injuries evaluated through accident reconstructions. *Stapp Car Crash Journal, 51*, 81–114.

Kuhn, A. W., & Solomon, G. S. (2016). Concussion in the National Hockey League: a systematic review of the literature. *Concussion, 1*(1).

Lehman, E. J., Hein, M. J., Baron, S. L., & Gersic, C. M. (2012). Neurodegenerative causes of death among retired National Football League players. *Neurology, 79*(19), 1970–1974.

McAllister, T. W., Flashman, L. A., Maerlender, A., Greenwald, R. M., Beckwith, J. G., Tosteson, T. D., Crisco, J. J., Brolinson, P. G., Duma, S. M., Duhaime, A.-C., Grove, M. R., & Turco, J. H. (2012). Cognitive effects of one season of head impacts in a cohort of collegiate contact sport athletes. *Neurology, 78*(22), 1777–1784.

McCrory, P., Meeuwisse, W., Dvorak, J., Aubry, M., Bailes, J., Broglio, S., Cantu, R. C., Cassidy, D., Echemendia, R. J., Castellani, R. J., Davis, G. A., Ellenbogen, R., Emery, C., Engebretsen, L., Feddermann-Demont, N., Giza, C. C., Guskiewicz, K. M., Herring, S., Iverson, G. L., Vos, P. E. (2017). Consensus statement on concussion in sport—the 5th international conference on concussion in sport held in Berlin, October 2016. *British Journal of Sports Medicine, 51*(11), 838–847.

McKee, A. C., Cantu, R. C., Nowinski, C. J., Hedley-Whyte, E. T., Gavett, B. E., Budson, A. E., Santini, V. E., Lee, H.-S., Kubilus, C. A., & Stern, R. A. (2009). Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy After Repetitive Head Injury. *Journal of Neuropathology & Experimental Neurology, 68*(7), 709–735.

McKee, A. C., Stein, T. D., Nowinski, C. J., Stern, R. A., Daneshvar, D. H., Alvarez, V. E., Lee, H.-S., Hall, G., Wojtowicz, S. M., Baugh, C. M., Riley, D. O., Kubilus, C. A., Cormier, K. A., Jacobs, M. A., Martin, B. R., Abraham, C. R., Ikezu, T., Reichard, R. R., Wolozin, B. L., Cantu, R. C. (2013). The spectrum of disease in chronic traumatic encephalopathy. *Brain, 136*(1), 43–64.

Meaney, D. F., & Smith, D. H. (2011). Biomechanics of Concussion. *Clinics in Sports Medicine, 30*(1), 19–31.

Mertz, H. J., & Irwin, A. L. (2015). Anthropomorphic Test Devices and Injury Risk Assessments. In Accidental Injury (pp. 83–112). Springer, New York.

Mihalik, J. P., Bell, D. R., Marshall, S. W., & Guskiewicz, K. M. (2007). Measurement of head impacts in collegiate football players. *Neurosurgery, 61*(6), 1229–1235.

Mihalik, J. P., Blackburn, J. T., Greenwald, R. M., Cantu, R. C., Marshall, S. W., & Guskiewicz, K. M. (2010). Collision Type and Player Anticipation Affect Head Impact Severity Among Youth Ice Hockey Players. *Pediatrics, 125*(6), e1394–e1401.

Mihalik, J. P., Guskiewicz, K. M., Marshall, S. W., Greenwald, R. M., Blackburn, ¶ J Troy, & Cantu, R. C. (2011). Does Cervical Muscle Strength in Youth Ice Hockey Players Affect Head Impact Biomechanics? In *Clin J Sport Med* (Vol. 21).

Moore, K. L., Dalley, A. F., & Agur, A. M. R. (2013). Clinically Oriented Anatomy (Illustrated).

Nightingale, R. W., McElhaney, J. H., Camacho, D. L., Kleinberger, M., Winkelstein, B. A., & Myers, B. S. (1997, November 12). The Dynamic Responses of the Cervical Spine: Buckling, End Conditions, and Tolerance in Compressive Impacts.

Oeur, R. A., Karton, C., Post, A., Rousseau, P., Hoshizaki, T. B., Marshall, S., Brien, S. E., Smith, A., Cusimano, M. D., & Gilchrist, M. D. (2015). A comparison of head dynamic response and brain tissue stress and strain using accident reconstructions for concussion, concussion with persistent postconcussive symptoms, and subdural hematoma. *Journal of Neurosurgery, 123*(2), 415–422.

Ommaya, A. K., & Gennarelli, T. A. (1974). Cerebral concussion and traumatic. *Brain, 97*(1), 633–654.

Patricios, J. S., Schneider, K. J., Dvorak, J., Ahmed, O. H., Blauwet, C., Cantu, R. C., Davis, G. A., Echemendia, R. J., Makdissi, M., McNamee, M., Broglio, S., Emery, C. A., Feddermann-Demont, N., Fuller, G. W., Giza, C. C., Guskiewicz, K. M., Hainline, B., Iverson, G. L., Kutcher, J. S., Meeuwisse, W. (2023). Consensus statement on concussion in sport: the 6th International Conference on Concussion in Sport–Amsterdam, October 2022. *British Journal of Sports Medicine, 57*(11), 695–711.

Post, A., Hoshizaki, T.B., Gilchrist, M. D., & Cusimano, M. D. (2017). Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. *Journal of Biomechanics, 61*, 183–192.

Post, A., Hoshizaki, B., & Gilchrist, M. D. (2012). Finite element analysis of the effect of loading curve shape on brain injury predictors. *Journal of Biomechanics, 45*(4), 679–683.

Post, A., & Hoshizaki, T. B. (2012). Mechanisms of brain impact injuries and their prediction: A review. *Trauma, 14*(4), 327–349.

Post, A., Hoshizaki, T. B., Gilchrist, M. D., Brien, S., Cusimano, M., & Marshall, S. (2015). The dynamic response characteristics of traumatic brain injury. *Accident Analysis and Prevention, 79*, 33–40.

Post, A., Hoshizaki, T. B., Karton, C., Clark, J. M., Dawson, L., Cournoyer, J., Taylor, K., Oeur, R. A., Gilchrist, M. D., & Cusimano, M. D. (2019). The biomechanics of concussion for ice hockey head impact events. *Computer Methods in Biomechanics and Biomedical Engineering, 22*(6), 631–643.

Post, A., Koncan, D., Kendall, M., Cournoyer, J., Michio Clark, J., Kosziwka, G., Chen, W., de Grau Amezcua, S., Hoshizaki, T.B. (2018). Analysis of speed accuracy using video analysis software. *Sports Engineering*, 21(3), 235-241.

Post, A., Oeur, A., Hoshizaki, T.B., Gilchrist, M.D. (2013). Examination of the relationship between peak linear and angular accelerations to brain deformation metrics in hockey helmet impacts. *Computer Methods in Biomechanics and Biomedical Engineering, 16*(5), 511-519.

Robertson, D.G.E., Caldwell, G.E., Hamill, J., Kamen, G., Whittlesey, S.N. (2014). Research methods in biomechanics. *Research Methods in Biomechanics*.

Robidoux, M. A., Kendall, M., Laflamme, Y., Post, A., Karton, C., & Hoshizaki, T. B. (2020). Comparing concussion rates as reported by hockey Canada with head contact events as observed across minor ice-hockey age categories. *Journal of Concussion, 4*, 205970022091128.

Rousseau, P. (2014). Analysis of Concussion Metrics of Real-World Concussive and Non-Injurious Elbow and Shoulder to Head Collisions in Ice Hockey. Doctorate Thesis, University of Ottawa.

Rousseau, P., & Hoshizaki, T. B. (2015). Defining the effective impact mass of elbow and shoulder strikes in ice hockey. *Sports Biomechanics, 14*(1), 57–67.

Rowson, S., & Duma, S. M. (2011). Development of the STAR Evaluation System for Football Helmets: Integrating Player Head Impact Exposure and Risk of Concussion. *Annals of Biomedical Engineering, 39*(8), 2130–2140.

Rowson, S., & Duma, S. M. (2013). Brain Injury Prediction: Assessing the Combined Probability of Concussion Using Linear and Rotational Head Acceleration. *Annals of Biomedical Engineering, 41*(5), 873–882.

Rowson, S., Duma, S. M., Beckwith, J. G., Chu, J. J., Greenwald, R. M., Crisco, J. J., Brolinson, P. G., Duhaime, A.-C., McAllister, T. W., & Maerlender, A. C. (2012). Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion. *Annals of Biomedical Engineering, 40*(1), 1–13.

Schmidt, J. D., Guskiewicz, K. M., Blackburn, J. T., Mihalik, J. P., Siegmund, G. P., & Marshall, S. W. (2014). The Influence of Cervical Muscle Characteristics on Head Impact Biomechanics in Football. *The American Journal of Sports Medicine, 42*(9), 2056–2066.

Siegmund, G. P., Blouin, J. S., Brault, J. R., Hedenstierna, S., & Inglis, J. T. (2007). Electromyography of superficial and deep neck muscles during isometric, voluntary, and reflex contractions. *Journal of Biomechanical Engineering, 129*(1), 66–77.

Smith, D. H., Meaney, D. F., & Shull, W. H. (2003). Diffuse Axonal Injury in Head Trauma. *Journal of Head Trauma Rehabilitation, 18*(4), 307–316.

Smith, T. A., Halstead, P. D., McCalley, E., Kebschull, S. A., Halstead, S., & Killeffer, J. (2015). Angular head motion with and without head contact: implications for brain injury. *Sports Engineering, 18*(3), 165–175.

Tuominen, M., Stuart, M.J., Aubry, M., Kannus, P., Parkkari, J. Injuries in men's international ice hockey: a 7-year study of the International Ice Hockey Federation adult world championship tournaments and Olympic winter games. *British Journal of Sports Medicine, 49*(1), 30-36.

Viano, D. C., Casson, I. R., & Pellman, E. J. (2007). Concussion in professional football. *Neurosurgery, 61*(2), 313–328.

Walsh, E. S., Rousseau, P., & Hoshizaki, T. B. (2011). The influence of impact location and angle on the dynamic impact response of a Hybrid III headform. *Sports Engineering, 13(*3), 135–143.

Yoganandan, N., Li, J., Zhang, J., Pintar, F. A., & Gennarelli, T. A. (2008). Influence of angular acceleration-deceleration pulse shapes on regional brain strains. *Journal of Biomechanics, 41*(10), 2253–2262.

Yoganandan, N., Pintar, F. A., Zhang, J., & Baisden, J. L. (2009). Physical properties of the human head: Mass, center of gravity and moment of inertia. *Journal of Biomechanics, 42*(9), 1177–1192.

Zhang, L., Yang, K. H., & King, A. I. (2004). A Proposed Injury Threshold for Mild Traumatic Brain Injury. *Journal of Biomechanical Engineering, 126(*2), 226–236.