Evaluating the Combined Effect of Thermoplastic Polyurethane Material in Ice Hockey Goaltender Helmets and Cervical Muscle Strength in Mitigating Concussion Risk During Simulated Horizontal Head Collisions

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Abstract

Concussions, or mild traumatic brain injuries (mTBI), occur due to an impact on the head and are the most common type of brain injury for goaltenders in the sport of ice hockey. The two main techniques used to mitigate concussion risk for ice hockey goaltenders include improving the impact absorption capabilities of the goaltender helmets and increasing the athlete's cervical muscle strength. Based on these two strategies, this study examined the effect of thermoplastic polyurethane (TPU) as a goaltender helmet liner material to mitigate concussion risk for individuals with different neck strength levels during simulated horizontal head collisions.

To address the purpose of this study, static testing was conducted to examine the material properties of the goaltender helmet liners, which was then used to identify the best TPU liner design that improved the performance of the goaltender helmet technology. One particular TPU liner design was found to weigh 2.7 times more on average than the standard liner; however, it was capable of statically absorbing 10.8 times more energy per kilogram than the standard liner. The researcher selected this TPU design and used it in repeated impact testing to further gauge the performance of the TPU and standard liner materials compared to a bare head without liner protection. Repeated dynamic impact trials revealed that the TPU liner mitigated impacts similarly to the standard liner, providing evidence of the TPU effectiveness as a possible helmet liner.

The researcher then conducted dynamic testing to simulate goaltenders' head collisions with different neck strength levels to assess the protective capabilities of the goaltender helmet technology during horizontal head collisions. A pneumatic impactor instrumented with a surrogate headform and mechanical neckform was used to simulate the horizontal head collisions. The dynamic data collection included 18 impact velocities across three helmet

locations (front, side, and back), four neck strength levels (30th percentile female, 50th percentile female, 50th percentile male, and 80th percentile male), and two helmet conditions (helmet fitted with TPU and standard helmet) for measures of liner acceleration and risk of injury.

The results of the dynamically simulated goaltenders' head collisions indicated that the TPU liner resulted in lower peak resultant linear acceleration (PRLA) values than the standard liner for the 80th percentile male neck strength level. Similarly, the TPU liner resulted in lower risk of head injury (HIC) scores for the 80th percentile male neck strength level. The results, however, revealed no statistically significant effect on neck strength levels for measures of PRLA and HIC for any of the helmet impact locations tested. Although the TPU liner performed better than the standard liner during static and repeated impact testing, the simulated goaltenders' head collision suggested further research is required into the design of the TPU liner to improve its capability to mitigate the risk of head injury across different neck strength levels and helmet locations.

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List of Abbreviations

COM	Center of mass
CSA	Canadian Standards Association
DAI	Diffuse Axonal Injury
HEEC	Hockey Equipment Certification Council
HIC	Head Injury Criterion
ISO	International Organization of Standardization
MPS	Maximal Principal Strain
mTBI	Mild Traumatic Brain Injury
NCAA	National Collegiate Athletic Association
NHL	National Hockey League
NOCSAE	National Operating Committee on Standards for Athletic Equipment
ROM	Range of motion
TPU	Thermoplastic polyurethane
VN	Vinyl nitrite

Chapter 1 – Introduction

Ice hockey goaltenders experience an incident of 1.7 concussions or mild traumatic brain injury (mTBI) per year, according to the National Collegiate Athletic Association (NCAA; Laprade et al., 2009). This outcome makes concussion injuries the second most common injury for ice hockey goaltenders in North America (Clark et al., 2018; Laprade et al., 2009). Concussions are a form of diffuse axonal injury (DAI) that can cause transient disturbances in mental status and trauma-induced alterations in cognitive function (Biasca et al., 2002). These injuries occur when the human head experiences intolerable energy levels in the brain tissue produced by rapid impact accelerations. The structures and material composition of the brain become vulnerable to the rapid application of load due to the transfer of impact energy to blood vessels and brain tissue. The brain tissue has a high bulk modulus and low shear modulus, causing the brain to behave in a manner that shows low resistance to shear forces produced due to oblique impacts on the head (Clark et al., 2018).

In the case of goaltenders in ice hockey, the risk of sustaining a concussion is due to head collisions encountered while playing ice hockey. Factors that influence the risk of concussion during head collisions include the speed of the puck striking a goaltender's head, speed and size of other players during head impacts, unpredictable rapid changes in the direction of play, and head collisions with hard ice surfaces. Since the goaltender's primary objective is to prevent pucks from entering the goal, the goaltenders are more susceptible to impacts from pucks and other players (Clark et al., 2018). Recent improvements in equipment technology have led to increased skating velocity, faster shots with the ice hockey pucks, and, consequently, more collisions while playing the sport (Clark et al., 2020; Meaney & Smith, 2011). This technological advancement in ice hockey has increased the occurrence of concussions and the need to improve the protective capacity of goaltenders' helmet equipment to mitigate concussion risk.

There are two possible strategies to mitigate the occurrence of concussions in the sport of ice hockey for goaltenders. The first strategy is to increase the goaltender's cervical muscle strength. Cervical muscle strength is an adaptable and trainable risk factor that can significantly aid in reducing sports concussions. Increased neck strength can decrease the risk of concussion from a given impact (Mihalik et al., 2011). Raising awareness and identifying individuals at a higher risk of injury is imperative to lower overall concussion rates. Increased identification of individuals at risk of concussion can be achieved by improving the quality and the frequency of neck strength testing. There is currently no normative data for ice hockey goaltender's neck strength.

The second strategy is to increase the protective capabilities of goaltenders' hockey helmets. Combining the shell, liner, cage, or strapping system of existing goaltender helmets with other elastic materials may increase their protective capacity. The use of thermoplastic polyurethane (TPU) material can potentially reduce the linear and rotational impact accelerations to the goaltender's head more effectively than the standard vinyl nitrite (VN) helmet liners used in current goaltender helmets (Gimbel & Hoshizaki, 2008). Although TPU has not been tested in goaltender helmets, it showed potential for decreasing concussion risk in ice hockey player helmets and boxing headguards (McGillivray, 2020; Rybak, 2020).

Combining these two strategies may provide an avenue to design better protocols to train goaltenders' necks in combination with improved helmet technology. Goaltender's helmet must be strong enough to protect the athletes against various concussions caused by different mechanisms of injury, such as collisions between athletes, falls on the ice, puck impacts to the head, and impacts with surface material in the ice rink. The helmet, however, must also be light enough not to fatigue the neck, which increases the risk of concussion.

A strategy that considers cervical muscle strength and helmet technology is required due to the complexity of the injury mechanisms involved for ice hockey goaltenders. Based on this need, the purpose of this study was to examine the combined effect of neck strength level and helmet liner type in mitigating the magnitude of the impact accelerations and the risk of concussion for goaltenders during horizontal head collisions at different head locations.

From the theoretical perspective, the results of the current study built on the research work of Clark et al. (2018), which suggested that current goaltender helmet testing protocol need to be modified to accommodate for horizontal impacts. The results of the current study also built on studies conducted by Pennock et al. (2021) and Carlson (2016) because the current study examined a wider range of neck strength levels to include male and female populations.

From the practical perspective, the results of the current study seemed to highlight the need to examine other TPU liner structure as an avenue to improve current helmet liner technologies. That is, further research on the shape, structure, and density of the TPU material may be necessary to improve the protective capabilities of the goaltender helmet technology during horizontal head collisions.

The results of this study have implications for researchers to further modify and improve the absorption capacity of goaltender helmet liners. The data from the current study also provided insight into the need to create standardized measures of neck strength levels for male and female goaltenders in ice hockey when examining the protective capabilities of goaltender helmets.

Chapter 2 - Review of the Literature

Head Injury Classifications

Head injuries are one of the most common injuries in ice hockey players and, more specifically, goaltenders (Clark et al., 2018; Decloe et al., 2014). These high rates of head injuries occur due to the fast-paced and contact nature of the sport, combined with the high-speed shooting of the hockey puck and low friction ice surface (Biasca et al., 2002). Ice hockey goaltenders are at significant risk for brain injury due to environmental factors related to ice hockey. The environmental factors include the speed and size of the players, speed of the puck, unpredictable changes in the direction of play, and collisions with the hard ice surface. Brain injuries are classified as either focal or diffuse depending on the type of impact to the head (Andriessen et al., 2010).

Direct impacts produce focal injuries to the head, which compress the brain's tissues (Andriessen et al., 2010). The compression can either occur directly underneath the location of the impact or opposite to the location of the impact (Andriessen et al., 2010). Depending on the magnitude of force and impact location, focal injuries are more likely to cause traumatic brain injuries (TBI; Andriessen et al., 2010). Traumatic brain injuries include skull fractures, cerebral contusions, and epidural hematomas (Andriessen et al., 2010; Kleiven, 2013). Initially, goaltender helmets were designed to protect against focal injuries prior to the understanding of other types of brain injury (Clark et al., 2020). Due to the vast improvement in helmet technology, these injuries have been reduced for ice hockey goaltenders (Clark et al., 2020). Despite focal or TBI being the most severe brain injury, DAIs are equally as concerning and occur far more frequently for ice hockey goaltenders.

A DAI is a form of traumatic brain injury caused by the rapid acceleration of the skull and brain (Mesfin et al., 2017). The term “diffuse” refers to the large volumes of distribution stress and strains, which cause brain damage and clinical injury (Clark, 2015). As the brain rapidly rotates in the skull, a shearing force damages the axons in the brain’s white matter. The damaged neurons can lead to a vast range of neurological dysfunction. These DAIs are the leading cause of death and disability in children and young adults in the United States of America (Mesfin et al., 2017). There are three grades of DAIs according to the Adams’ DAI Classification (Mesfin et al., 2017). Grade 1 is a mild DAI with microscopic white matter changes to the cerebral cortex, corpus callosum, and brain stem (Mesfin et al., 2017). Grade 2 is a moderate DAI with gross focal lesions in the corpus callosum. Grade 3 is a severe DAI with the same findings as a grade 2 plus additional focal lesions in the brainstem (Mesfin et al., 2017). Concussions are the mildest form of DAIs and typically categorized as a grade 1 injury. As the frequency of focal injuries decreased for goaltenders, DAIs causing concussions have remained prevalent. Due to DAIs not being visible without the help of medical instrumentation, they can be challenging to detect and diagnose; therefore, understanding the pathophysiology of concussion is essential when addressing this type of injury.

Concussions

As stated previously, concussions are the second most common injury for ice hockey goaltenders caused by violently shaking the brain due to an impact to the head, resulting in the impairment of neurological brain functions (Anderson et al., 2020; Clark et. al., 2018; Laprade et al., 2009). The violent shaking inside the skull strains the brain tissue causing a transient local deformation, which damages and tears the nerve fibers (Strich, 1961). When this occurs, there is a production of reactive oxygen and an increase in glucose utilization, combined with a decrease in blood flow, leading to an energy mismatch (Harmon et al., 2019). The energy mismatch causes insufficient levels of glucose and other nutrients, which increases vulnerability for the well-being of the individual (Harmon et al., 2019). If a second injury occurs during this time of increased vulnerability, it can lead to more swelling of the brain and potentially death. This pattern is called second impact syndrome (Wetjen et al., 2010). Based on these concerns, it is essential to understand the signs and symptoms of a concussion to minimize the risk of brain injury.

Symptoms of Concussion

The signs and symptoms of a concussion are highly individualized and specific to each type of head injury. Several factors can change the outcome of a particular head impact, including the applied force, location, and personal history of concussion (Clark et al., 2018). According to the American Medical Society for Sports Medicine, there are six categories of concussion symptoms including headache/migraine, anxiety/mood, fatigue, vestibular, ocular, and cognitive symptoms (Pujalte et al., 2020). Headache/migraine symptoms consist of nausea, sensitivity to light, neck pain, or sensitivity to noise (Pujalte et al., 2020). Emotional symptoms may encompass irritability, mood lability, depression, and frustration (Pujalte et al., 2020). Fatigue-related symptoms of concussion involve decreased and increased sleep, as well as, having irregular sleeping patterns (Pujalte et al., 2020). Vestibular symptoms are associated with alterations in balance. Ocular symptoms of concussion entail double vision, blurred vision, and difficulty reading (Pujalte et al., 2020). Finally, cognitive symptoms of concussion may include decreasing mental clarity, fatigue, mental fogginess, memory issues, confusion, and difficulty concentrating (Pujalte et al., 2020).

It is essential to understand that there is overlap among these categories, which can lead to difficulty in understanding the concussive injury. The combination of these symptoms can lead to alterations in the activities of daily living; however, certain types of symptoms can be more problematic for ice hockey goaltenders.

Vestibular, ocular, and cognitive concussion symptoms, for example, are detrimental to a goaltender’s performance on the ice. The goaltender needs to move rapidly in front of the net using explosive lower-body power, balance, visual acuity, and the ability to remain singularly focused (Marcotte-L’heureux et al., 2021). If the goaltender’s balance or vestibular function is inhibited in any way, it affects the ability to prevent the puck from entering the net while it is travelling up to 160 km/h in an unpredictable manner. Additionally, goaltenders must have exceptional visual acuity and be able to accurately track the hockey puck around the ice and intercept the puck if it is directed towards the goal (Panchuk et al., 2017). This task is cognitively demanding in the sport of ice hockey. Consequently, any type of cognitive impairment due to a concussion compromises the goaltender’s performance while playing ice hockey.

Hence, understanding the varying signs and symptoms of a concussion becomes imperative when attempting to increase a goaltender’s safety. Researchers also pay much attention to the biomechanics specific to concussions as an avenue to understand the nature of the head injury and develop methods to increase a goaltender’s safety.

Biomechanics of Concussion for Goaltenders

The primary cause of a concussion is the deformation or strain that occurs in the brain tissue during rapid accelerations (Kleiven, 2013). When a collision to the head occurs, it causes the head and neck to accelerate both linearly and rotationally in all three x, y, and z directions

(Namjoshi et al., 2013). The relative amount of acceleration produced during the head collision depends on the type of force, direction, location, and tissue properties of the skull and brain.

Currently, it is challenging to measure this acceleration by just assessing the deformation of the skull and brain tissue. In response to this, other measures have been explored to assess the magnitude of the impact and the risk of concussion. The two primary methods of measuring concussion risk are linear and angular acceleration. Linear and angular accelerations cause the head to move drastically during an impact, while producing shear, tensile, and compressive forces on the brain tissue (Andriessen et al., 2010). Linear acceleration is the result of a force directed through the center of mass (COM) of the head and is the rate at which linear velocity changes over time described in units of gravity (g; Robertson et al., 2014). The unit g refers to multiples of gravitational acceleration, which is 9.81 m/s^2. Rotational or angular acceleration is the change in angular velocity over time, measured in radians per second squared (rad/s^2). Rotational acceleration is caused by forces acting on an object that do not pass directly through the COM of the head, causing off-center rotation (McLean & Anderson, 1997).

Initially, linear acceleration was the primary variable assessed when addressing concussion risk. It was discovered, however, that linear acceleration was more closely related to focal injuries as compared to diffuse brain injuries (Gennarelli et al., 1972). An in vivo study showed that pure translational head impacts in animals did not result in a concussion (Gennarelli et al., 1972). It was also determined that angular acceleration produced more significant tissue shearing and brain deformation than linear acceleration (Kleiven, 2013; Meaney & Smith, 2011).

Due to the nature of real-world impacts, it is almost impossible to have only linear or angular acceleration occurring during an injury. Consequently, it is recommended to consider both linear and angular accelerations when assessing concussion injury risk (Clark et al., 2018).

The mechanism of an injury must be understood to protect goaltenders from a concussion. Clark et al. (2018) analyzed 20 concussion-causing impacts in National Hockey League (NHL) games. With the use of motion capture and analysis software, they determined the type of impact, velocity, mass, and angle that caused the concussion. The research team utilized the NHL database to determine the player's mass and confirm if the incident caused a concussion. They separated the helmet's circumference into 12 sectors of 30° degrees each and six segments transversely to determine the exact location of the impact. Once these data points were collected, laboratory reconstructions occurred. They found three categories of mechanisms of injuries for goaltenders including falls, collisions, and projectile impacts.

Falls

Falls are the most common events leading to concussions in the general population and are also a common mechanism of concussion for goaltenders (Clark et al., 2020; Pujalte et al., 2020). Falling causes impacts that are short in duration but high magnitude linear and rotational acceleration to the head and brain (Post et al., 2019).

For ice hockey players, and specifically goaltenders, helmets are certified based on drop tests to simulate falls leading to helmet design to mitigate fall-related forces and injuries (Clark, 2015). Current goaltender helmets are required to meet the standard for skull fractures and other serious TBI based on falling simulations (Clark, 2015).

This standard focuses on preventing the most severe brain injuries for goaltenders (Clark, 2015). Despite a reduction in serious brain injuries, mTBI and less severe injuries have remained. Kendall et al. (2012) showed that falling to the ice generated a maximum principal strain (MPS) of 0.424 on the brain tissue and a linear acceleration of 264.4 g on average. The maximum principal strain is the maximum amount of strain experienced on the brain during an impact and expressed as a decimal out of a total value of one (Clark et al., 2018). The researchers found that the combination of maximal linear and angular accelerations (106 g and 7900 rads/s^2) was above the 80% threshold for injury risk (Kendall et al., 2012).

Falling and hitting the head on the ice while wearing a helmet is associated with a high risk of concussion for goaltenders in ice hockey (Clark, 2015). Clark (2015), for example, found that falls produced the highest risk of injury for goaltenders in ice hockey compared to other incident events. This outcome is due to the large amount of acceleration transferred to the head due to the rigid ice surface, which results in a high magnitude of acceleration for a short duration (Hoshizaki et al., 2014; Post et al., 2013).

Clark (2015) also compared helmeted to unhelmeted fall simulations and found that goaltender helmets significantly reduced the accelerations for falls, which may explain why concussions due to falling are not the most common injury for goaltenders in ice hockey. Although falls are the most common mechanisms of injury causing concussions in the general population, horizontal head collisions are the most common and serious type of head injuries causing concussions for goaltenders (Clark, 2015).

Collisions

Collisions between two players cause the most concussions for goaltenders in ice hockey (Clark, 2015). Laprade et al. (2009) found that all seven of the goaltender concussions observed between 2000 and 2007 in the men's NCAA were caused by collisions. Collisions that cause a concussion occur when a player's body makes contact with the goaltender (Clark, 2015). These impacts can be defined as high-mass and high-velocity impacts, leading to the transfer of high amounts of energy during the collision (Clark, 2015). These collisions occur as a result of a player losing control while skating at high speed, a player being pushed into the goaltender, and from on-ice altercations where contact is made with the goaltender (Liberman & Mulder, 2007).

Researchers have conducted additional studies on head collisions for ice hockey goaltenders by reconstructing head injuries due to shoulder impacts and flagrant fouls (Clark, 2015). Kendall et al. (2012), for example, simulated horizontal shoulder impacts to the side location of the head. A linear impact system fitted with a 50^{th} percentile adult male headform and neckform was used for this study. The inbound velocity of the impactor arm was 6.5 m/s, because it was within the range of possible impact velocities during a game of ice hockey (Kendall et al., 2012). This study found that shoulder impacts to the side of the headform produced peak linear and rotational accelerations of 112.5 g and 9659 rad/s^2, respectively, which were above 50% risk for concussion (Zhang et al., 2004). Furthermore, the average MPS values were found to be 0.035, which were associated with a 50% chance of getting a concussion demonstrating the high risk of a mTBI from shoulder impacts (Kendall, Post, et al., 2012).

Research studies have also been completed with elbow impacts to the head. Reconstructing flagrant fouls in ice hockey has provided insight into the potential outcome of a skater's elbow colliding with a goaltender's head. Coulson (2011) reconstructed various major penalties in ice hockey to determine the peak linear and rotational accelerations of a Hybrid III head form. Coulson et al. (2009) recruited eight male hockey players to strike a headform moving at 0.56 ± 0.11 m/s with their elbow at a maximal force. Coulson et al. (2009) found that the players' peak linear accelerations generated were well above the 50% probability thresholds and the players' peak rotational accelerations generated were above the 80% probability threshold.

Although these studies showed significant injury risk, goaltenders wear different helmets than other ice hockey players. Differences in foam type, shell geometry, and other characteristics may influence the response to protect the head against a concussion for goaltenders (Rousseau et al., 2009). Clark (2015), for example, found that goaltender helmets were not effective at reducing the risk for collision-related concussions. Despite significant differences between helmeted and unhelmeted conditions, the differences were minimal and did not represent a decrease in the risk of concussion (Clark, 2015).

Goaltenders' helmets are also instrumented with a mask. Clark (2015) identified that ice hockey goaltender's masks are less effective at reducing the risk of concussion from collisions due to the method that they are tested. Currently, ice hockey goaltender masks standards only consider drop tests to represent falls and do not consider collisions. Falls, and collisions such as the shoulder impacts to the head, are defined by different impact characteristics, resulting in different responses, and it is unlikely that falls alone can provide an accurate description for the risk of collisions (Hoshizaki et al., 2014). Other mechanisms of injury such as projectile impacts may highlight the concussion risk differently.

Projectile Impacts

Projectile impacts primarily occur because a 0.17 kg puck constructed of vulcanized rubber strikes a goaltender's head in a projectile motion (Clark et al., 2018). The definition of projectile impact is a high-velocity impact with minimal energy due to the low mass (Clark et al., 2018). These projectile impacts occur more frequently to goaltender head than any other type of impacts (Clark et al., 2018). However, due to the design of the goaltender helmet, a low percentage of impacts actually cause a concussion (Clark et al., 2018).

The most elite professional players, for example, can shoot the puck up to 40-45 m/s, and many youth athletes can shoot up to 20-30 m/s (Bežák & Přidal, 2017). These velocity ranges, however, are large enough to potentially cause a concussion in goaltenders under the correct impact parameters.

Goaltender helmets have evolved to dissipate the impact force of pucks (Clark, 2015) with the use of a cage made of carbon, steel, or titanium, a helmet shell made of carbon and Kevlar® composite, fibreglass, or polycarbonate, and an energy-absorbing liner consisting of VN foam (Clark et al., 2018). The three main elements of an ice hockey helmet for goaltenders to protect the head against concussions include the thickness of the liner, the stiffness of the shell, and the external shell geometry (Clark et al., 2018). The thickness of the liner describes the amount of VN material and ranges between 11 mm to 16 mm in current goaltender helmets (Clark et al., 2018). The stiffness of the shell is determined by the construction materials, with Kevlar®-based helmets being the most rigid (Clark et al., 2018). The external shell geometry refers to the shape of the helmet and how it is contoured (Clark et al., 2018).

Ouckama and Pearsall (2014) found that softer liners performed better during low-speed impacts, whereas stiffer liners were optimal for reducing impact accelerations. Ouckama and Pearsall (2014) also identified that the stiffer the shell was, the lower the risk of injury. The shape of the goaltenders' helmets also affected their protective capabilities. Many new helmets are rounded or have wave-like moulding on the top of the forehead to decrease the likelihood of a perpendicular impact to the head (Spyrou et al., 2000).

The more energy the helmet can deflect, the less energy will transfer to the brain (Ouckama & Pearsall, 2014). Ouckama and Pearsall (2014) indicated that after one impact over 35 m/s, a helmet becomes less effective and requires replacement. Consequently, it becomes critical to understand the parameters of the impact conditions to reconstruct the injury and better assess the performance of the helmet technology.

Role of Impact Conditions

The parameters of the impact conditions play a critical role in understanding the protective capabilities of the ice hockey goaltenders' helmets. The combination of impact velocity, angle of impact, mass, location, and the state of the goaltender determines the magnitude of the brain injury (Gennarelli et al., 1982; Kleiven, 2003; Pellman et al., 2003).

The inbound velocity of the collision, regardless of impact type, has a relationship with the magnitude of the dynamic response. Increased velocity will lead to increased rates of injury when all other factors have been controlled for (Clark et al., 2018).

The angle of impact will influence the dynamic response. For ice hockey goaltenders, there are impacts from pucks, the playing surface, as well as from the bodies of other players. If a puck strikes the goaltender's helmet perpendicular to the head, for example, greater force is transferred directly to the head than if the impact occurs on an angle (Clark et al., 2018). When the impact occurs on an angle, the force transferred directly to the head is the result of the impact force multiplied by the sine of the impact angle (Robertson et al., 2014). The smaller the angle of impact, the less the amount of force transferred directly to the head and helmet. The force applied to a rigid body in an off-center axis, however, creates rotation about the other two in proportion to the axis' perpendicular distance, which also influences the dynamic response of the head and helmet (Hoshizaki et al., 2012).

The location of the impact also leads to variance in the outcome. A study by Clark (2015) identified that impact location significantly affected linear acceleration, rotational acceleration, and MPS. The side and back impacts produced higher linear and rotational acceleration outcomes than similar impacts to the front of the helmet. Another difficulty arises with the relationship between linear and rotational acceleration with impact location. As the impact moves farther from the COM, the magnitude of the linear acceleration decreases and rotational acceleration increases (Clark, 2015). Clark believed that there must be consideration of both linear and rotational accelerations when determining risk injury measures for concussion (Clark, 2015). The purpose of determining the linear and angular accelerations that occur during an impact is to determine the risk of that impact causing a concussion. Determining the risk of a particular injury requires the use of a severity index.

Severity Index

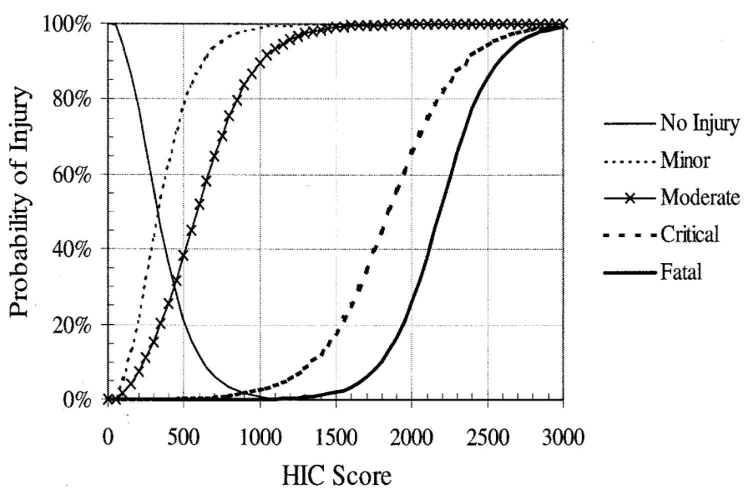
Despite difficulties in diagnosing concussions with measures of linear accelerations only, researchers have made attempts to develop severity indices to estimate the risk of TBI and mTBI due to a head collision. The first attempt was the development of the Wayne State Tolerance Curve (WSTC) by determining the relationship between intensity and duration of intracranial pressure. The researchers obtained the intracranial pressure by blowing air onto animals' exposed dura mater (Namjoshi et al., 2013). This model was the first biomechanically based quantitative human brain injury model and led to several other models, including the Gadd Severity Index (GSI). The GSI estimates injury hazard due to an anterior-posterior acceleration-time pulse waveform to the human head (Hodgson et al., 1970). The index relies on the premise that the severity of the head injury is related to overall head acceleration including the pulse duration and magnitude of the impact (Hodgson et al., 1970). The GSI effectively determines severe brain injuries and skull fractures; however, there are limitations when predicting mTBI like those experienced by ice hockey goaltenders (Hodgson et al., 1070). For linear acceleration, the risk of concussion is usually estimated using the head injury criterion (HIC) index.

Head Injury Criterion

The HIC in Equation 1, theoretically predicts skull fractures and cerebral contusion from linear acceleration (Kimpara & Iwamoto, 2012). The HIC value utilizes scalar values of the accelerations at the center of gravity in the x, y, and z directions. A 36 ms cut-off time is applied to the HIC to capture the time interval that produces the maximal HIC scores. This time cut-off also reduces the influence of long-time durations of the impact with low accelerations (Ouckama & Pearsall, 2014). A HIC threshold value of 500 represents an 80% chance of a minor head injury or a 40% chance of a moderate head injury (Hutchinson et al., 1998). Figure 1 illustrates the probability of a specific head injury level for a given HIC score.

Figure 1

Probability of Injury Based on HIC Score



Note. Probability of a Specific Head Injury Level for a Given HIC Score. From "Standard Specification for Impact Attenuation of Surfacing Materials Within the Use Zone of Playground Equipment" by Seveska, p. 17 West Conshohocken, PA; ASTM International

$$HIC = \left\{(t_2 - t_1)\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}\right\}_{max} \quad (1)$$

where:

a = resultant linear acceleration

t_2 - t_1 = time interval where peak acceleration occurs; $t_2 - t_1 \leq 36$ ms

Although researchers have developed index models to estimate the risk of brain injury and concussions to better design helmet technologies, they still believe that current helmet materials do not perform very well at reducing the acceleration of impacts such as shoulder collisions to the head that cause longer periods of acceleration, and, consequently, brain tissue damage (Clark et al., 2018). The combination of helmet technology with improvement in athletes' neck cervical strength, however, may offer an avenue to reduce concussion risk for linear impacts, as well as longer durations of the head collision (Clark et al., 2018; Schmidt et al., 2014).

Concussion and Cervical Spine

Cervical spine injuries are reported to have similar clinical symptoms to concussion injuries and often occur in conjunction with concussions (Elkin et al., 2016). Comorbid neck injuries and concussions occur between 41.9% and 68.2% of the time (King et al., 2020). The cervical spine region is vulnerable to injury when a concussion occurs due to a head collision because of the proximity to the head. These cervicogenic symptoms are present in acute and chronic stages and are associated with worsened clinical outcomes and recovery times (King et al., 2020). Since concussions occur from inertial loading due to direct and indirect impacts to the head, it is essential to understand the musculoskeletal structures of the head and neck.

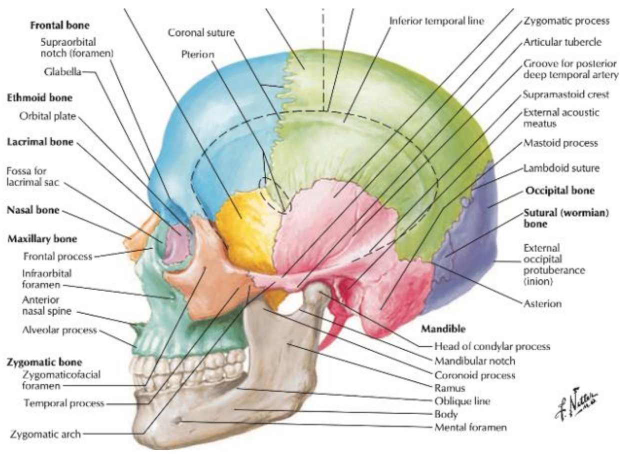
Anatomy of the Head and Neck

The structure and shape of the cervical vertebrae influence the movements controlled by the muscles of the head and neck (Bogduk & Mercer, 2000). Only the systems associated with concussions will be discussed in this section due to the complexity of the cervical region of the spine.

The skull is a boney structure responsible for protecting the brain and consists of the cranium and the facial bones (Figure 2). The cranium houses the brain, and it is formed by eight separate bones, including two parietal bones, two temporal bones, the frontal bone, the occipital bone, the ethmoid bone, and the sphenoid bone (Tortora & Nielsen, 2017). The cranial bones form the cranial cavity and are fixed structures held together by sutures to protect the brain. Within the cranial cavity, the brain has a protective layer of meninges that act to stabilize the brain (Torta & Nielsen, 2017). The brain is also surrounded by cerebrospinal fluid (CSF). The CSF is a liquid comprised primarily of water acting to protect the brain from both physical and chemical injuries. Rapid movements cause the brain to shift in the CSF, impacting the inner walls of the cranium (Andriessen et al., 2010). When the brain impacts the inner surface of the cranium, damage can occur to the intercranial tissues (Rivara & Graham, 2014). Damage to the intercranial tissues from impacts disrupts the cell function, causing a concussion (Rivara & Graham, 2014). The movement of the head is dictated by the structure of the cervical spine leading to variation in injury risk when impacts occur.

Figure 2

Bone Structures of the Human Skull

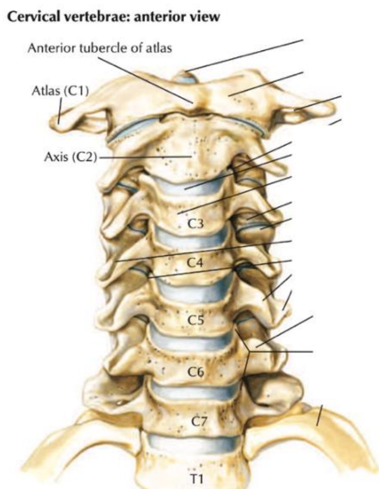


Note. Sagittal view of the bones of the human skull. From "Atlas of Human Anatomy" by F. H. Netter, 2019, p. 28 Philadelphia, PA: Elsevier Inc.

The cervical spine (Figure 3) supports the skull, acts as a shock absorber and facilitates the transfer of the weight and bending of the neck (Nordin & Frankel, 2012). The seven cervical vertebrae form a column of bones that vary considerably in function and range of motion (ROM). Because of the mobility of the cervical region, the neck is capable of flexing, extending, rotating, and side flexing through all three anatomical planes (Nordin & Frankel, 2012). The high degree of mobility is a characteristic that causes the stressing of the neck structures to their limits during inertial loading impacts (Morin et al., 2016). Some of the prime movers of the cervical spine include the sternocleidomastoid, anterior scalene, middle scalene, posterior scalene, and trapezius muscles (Figure 4; Cournoyer et al., 2021). Additional muscles that aid in the overall muscle strength in the cervical region include the semispinalis capitis, splenius capitis, and longissimus capitis muscles (Tortora & Neilson, 2017). These muscles contribute to the development of the cervical muscle strength and is responsible for handling up to 80% of the mechanical load placed on the head during an impact (Schmidt et al., 2014).

Figure 3

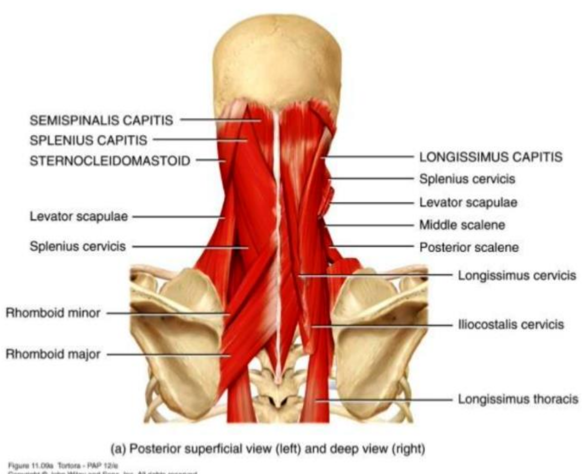
Bone Structures of the Cervical Spine



Note. Frontal view of the bone structure of the cervical spine. From "Atlas of Human Anatomy" by F. H. Netter, 2019, p. 31 Philadelphia, PA: Elsevier Inc.

Figure 4

Cervical Musculature



Note. Posterior view of the muscles of the neck that move the head. The left side of the diagram illustrates the superficial cervical muscles, while the right side identifies the deep view of the cervical musculature. Adapted from "Principles of human anatomy" by G. T. Torta and M. T. Nielson, 2012, (12th ed.), p. 361 Hoboken, NH: John Wiley & Sons, Inc.

Defining Strength, Torque, and Stiffness

Previous literature documenting the properties of the cervical musculature and the behaviour of the cervical spine during dynamic impact testing has used the terms of strength, torque, and stiffness somewhat interchangeably and without consistent definitions. Clarifying the operational definitions of strength, torque, and stiffness is essential to understanding their relationship with concussions. Strength is defined as the amount of force a particular muscle group can produce (Richards, 2008). Strength can be influenced by the muscle insertion, the muscles' angle of pull, and the speed and type of contraction (Richards, 2008). Muscle torque is the product of the force produced by the muscle and the shortest perpendicular distance from the muscle attachment to the axis of rotation (Nordin & Fankel, 2012). Muscle force and muscle torque are often used interchangeably in the literature because of their close relation to each other. Torque is dependent, however, on the length of the moment arm and it is difficult to accurately calculate for cervical muscle torque. Neck stiffness is the ability of the cervical structures to resist deformation in response to an applied load (Eckner et al., 2014). For neck strength simulation studies, however, cervical muscle strength values can be used to determine the torque needed to calculate the stiffness of a mechanical neckform to represent the cervical muscle strength of an athlete (Pennock et al., 2021).

Cervical Muscle Strength

Cervical muscle strength is a mitigating factor for linear and angular accelerations experienced by the head during impacts (Schmidt et al., 2014). Athletes with stronger cervical spine muscles have a decreased risk of getting a concussion or head injury because the neck has the capacity to mitigate the accelerations applied during the indirect and direct impact loadings to the head and neck (Mihalik et al., 2011).

A study performed by Gutierrez et al. (2014) found a positive relationship between increased cervical muscle strength and decreased head accelerations in response to external forces. If an athlete has insufficient levels of cervical muscle strength, the athlete may not be able to generate sufficient force to dampen the accelerations experienced by the head. This outcome has been confirmed by Collins et al. (2014), who identified that athletes who had been diagnosed with a concussion had significantly less cervical muscle strength and smaller neck circumferences than healthy athletes.

A similar study found that athletes who sustained a concussion in the previous 12 months had higher head accelerations during impacts caused by decreased muscle activation of the trapezius and the splenius capitis (Bussey et al., 2019). The results of this study supported the notion that increases in cervical muscle strength decreased the magnitude of the linear and rotational acceleration experienced by the brain during an impact. When considering muscle activation and other factors relating to risk of injury, sex-based differences in cervical muscle strength must be considered.

Sex-based differences in cervical muscle strength are consistent throughout the literature, including athletic populations where females demonstrated weaker cervical musculature than males (Cagnie et al., 2007). Additionally, when comparing male and female athletes' neck circumferences in centimeters, males had larger neck circumferences (M = 36.10 cm, SD = 2.53) than females (M = 32.55 cm, SD = 2.35; Collins et al., 2014).

Differences in neck circumferences between males and females seem to play a role in increasing the risk of concussion for female athletes in the sport of ice hockey. In a study by Collins et al. (2014), for example, it was shown that athletes who had been diagnosed with a concussion had smaller neck circumferences and significantly less cervical muscle strength than those who did not sustain a concussion.

Different research studies also highlight the differences in cervical muscle strength for males and females in the sport of ice hockey. Broennnle (2011), for example, conducted a study to establish normative neck strength data for male ice hockey players and found higher values for males' cervical muscle strength (M = 185.75 N, SD = 46.95) as compared to a study conducted by Pennock (2018) on female ice hockey players (M = 76.01 N, SD = 17.52) for measures of overall neck strength. Overall neck strength refers to the average of the strength scores for flexion, extension, and side flexion.

Cervical muscle strength is essential for ice hockey goaltenders due to the demands of the position, the weight of the equipment, and the posture of the goaltender. In addition to the goaltender being the only player that remains on the ice for the entire 60 minutes of the game, a goaltender helmet weighs roughly 1.2 kg (Clark et al., 2020). When goaltenders are in their ready stance, their knees are flexed to roughly 100º with their upper body leaning forward (Figure 5). The goaltender's ready stance refers to the goaltender's posture when the puck is near, and the offensive team can attack the goal. The goaltender's hands and upper body are in front of the goaltender's net, while the head is positioned forward in front of the rest of the body (Bell et al., 2008). The protruded position of the head, with the addition of the weight of the helmet, can potentially lead to fatigue of the neck musculature (Bell et al., 2008). Therefore, when designing and studying goaltender helmets, the position of the goaltender's head and fatigue of the neck musculature must be considered in the head injury prevention process (Bell et al., 2008; Schmidt et al., 2014).

Figure 5

Ice Hockey Goaltender Stance

Goaltender Helmets

The development of ice hockey goaltender helmets for head injury prevention has evolved in the last six decades. The first example of goaltenders wearing facial protection in professional ice hockey began in 1959, 40 years after the creation of the NHL. These helmets were constructed of plaster and designed to protect against facial lacerations. In late 1970s, manufacturers began designing helmets with additional features to protect the head from more than facial lacerations (Biasca et al., 2002).

Current goaltender helmet technologies, for example, are constructed of a shell, cage, and foam liner held together with glue, screws, and elastic straps to provide better protection for the goaltender’s head during collisions. The shell is the rigid exterior portion of the helmet that consists of varying blends of fibreglass, carbon fibre, and Kevlar® (Clark et al., 2020). The purpose of the shell is to deflect forces that contact the helmet, specifically projectiles (Spyrou et al., 2000). The wavy shell geometry in the forehead region is designed to deflect oblique impact forces from puck collisions to the head instead of the energy being absorbed by the helmet and head (Spyrou et al., 2000). The cage of the helmet is made of carbon steel, and it attaches to the shell with a series of screws. The cage is used to protect the face against puck impacts (Clark et al., 2020). Finally, the foam liner is a thin layered structure consisting of VN or other foam materials to absorb impact forces.

Many new commercial helmets use different liner densities to mitigate the concussion risk across different locations of the helmet. All commercial goaltender helmets must be certified to improve a player's safety.

Goaltender Helmet Certifications

Goaltenders must wear a certified helmet by the Canadian Standards Association (CSA) in Canada, the Hockey Equipment Certification Council (HECC) in the United States of America, or the International Organization of Standardization (ISO; Clark, 2015). Helmets must be able to function at a range of temperatures and different impact conditions. These standards primarily involve drop tests and measures of peak linear acceleration to establish attenuation properties because most concussions occur in the general population due to falls on ice (Clark, 2015).

Each helmet must limit peak linear acceleration to below 275 g for a vertical drop of a helmeted headform to qualify for certification (Clark, 2015). This certification standard has resulted in goaltender helmets designed to withstand falls. Clark et al. (2018) conducted a study analyzing goaltender concussions and tt was identified that goaltender helmets can withstand impacts for both projectile and falls; however, they were ineffective when collisions involved players' bodies contacting the head (Clark et al., 2018). Additionally, Clark et al. (2018) found that safety certification standards and equipment manufacturers should develop standards for ice hockey goaltender masks that are more effective at reducing the risk of head injury mechanisms causing concussions.

A potential method of improving goaltender helmet technologies beyond the current testing standard is to enhance the quality, structure, and material type of the shock-absorbing liners (Clark et al., 2018). One of the developments in the helmet industry is the use of polyurea-based plastics and foams (Qi & Boyce, 2005). One shock-absorbing material that has shown promise in early trials is the use of TPU liner material (Gimbel & Hoshizaki, 2008).

Thermoplastic Polyurethane

The TPU material (Figure 6) dissipates force more reliably than VN foams due to the structure of the material (Qi & Boyce, 2005). The TPU material contains large cells to store air within a diameter ranging between 300 and 500 µm (Ramirez & Gupta, 2018). As stress increases on the TPU material due to a force impact, the large cells collapse, forcing the air to escape through the perforations of the TPU material. Since the air escape rate cannot catch up with the loading stress applied to the material, some of the cells stiffen and deform slowly (Ramirez & Gupta, 2018).

This material property of the TPU allows energy to dissipate by the dynamic bending, twisting, and rotation of the cell walls similarly to other foam materials. In addition, the cell walls slide and stretch freely with their flexible molecular chains leading to increased energy dissipation (Ramirez & Gupta, 2018). In summary, this mechanism enables the TPU material to manage different strain rates that occur within a singular loading event.

Figure 6

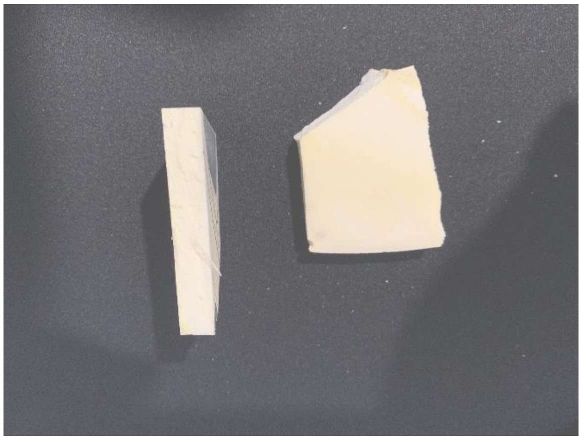
Thermoplastic Polyurethane

On the contrary, the VN material (Figure 7) contains three distinct regions when examining its stress versus strain relation, which cause it to respond differently from TPU under stress. The first is the linear elastic region, which contains the cells associated with the bending walls of the material when stress is applied. The second is the stress plateau region, which results from the continuous collapse of the foam's cell structure. This region includes the viscoelastic or recoverable portion, as well as the irrecoverable plastic-type that leads to permanent deformation of the material under stress (Nagy et al., 1974). The third is the densification region, which is where the collapsed cells begin to further compress into each other causing a rising stress to the affected region.

The linear elastic and stress plateau regions determine the foam attenuation under stress. These regions, however, are dependent on the rate of loading (Jeong et al., 2012). During a collision, for example, the material strain rate is lower and then increases as the foams density increases. As the material compresses due to bending, twisting, or buckling, it dissipates the impact energy through the viscoelastic process (Jeong et al., 2012). Finally, the stress reduces significantly once the impact slows down (Ramirez & Gupta, 2018).

Figure 7

Vinyl Nitrite Helmet Material



One possible way to optimize the VN foam performance will be to create materials with modulated densities and stiffnesses within the same loading structures of lining material. This engineering will be costly to produce in the market (Ramirez & Gupta, 2018). Another method will be changing the lining material of a helmet and using TPU materials that act entirely differently under tension (Ramirez & Gupta, 2018).

Several studies have incorporated TPU and other polyurea-based materials in sports helmets, including a study by Ramirez and Gupta (2018). The researchers replaced one football helmet with a viscoelastic polyurea foam and compared the performances with a standard helmet. The researchers dropped the helmets from a range of 0.305 m to 1.524 m. Comparison of the helmets utilized the peak linear acceleration and HIC values. On average, the TPU had a 22% reduction in peak linear acceleration and a 25% reduction in the HIC values.

Gimbel and Hoshizaki (2008) identified similar results when they compared three different VN helmets with two different TPU helmets. They dropped each of the helmets from 0.20 m, 0.40 m, and 0.60 m, adding a mass of 3.04 kg, 4.04 kg, and 5.04 kg, and measured the resulting linear accelerations. The TPU managed the impact energy significantly better through a broader range of drop heights and impact masses than the VN material.

Despite the early clinical success of TPU materials in ice hockey helmets, research has yet to address goaltender helmets. The technique to assess goaltender helmet performance entails the use of impactors, mechanical neck, and surrogate headforms to simulate either free falling or horizontal head collisions.

Mechanical Simulations of Head Collisions

Simulating impacts to investigate sports-related concussions is a popular method of examining head and neck-related injuries. Previous research using head impactors to examine head impact biomechanics has used similar procedures.

Most simulation-based research has used the Hybrid III surrogate headform and neckform (Beckwith et al., 2012; Clark et al., 2018; Oeur et al., 2014; Pellman et al., 2003; Walsh et al., 2011). Other studies have used the National Operating Committee on Standards for Athletic Equipment (NOCSAE) surrogate headform for head impacts (Carlson et al., 2016; Pennock et al., 2021; Post et al., 2019; Zerpa et al., 2017).

Both types of headforms are equipped with accelerometers to measure linear and angular accelerations (MacAlister, 2013). The NOCSAE surrogate (Figure 8) is a gender-neutral headform with the 50th percentile adult head and has increased anatomical representative features than other headforms (MacAlister, 2013). The combination of the mechanical headform and neckform is the standard equipment for measuring concussion risk for ice hockey goaltenders and testing the quality of goaltender helmets.

Nur et al. (2015) mounted ice hockey goaltender helmets on a headform and neckform to determine the protective capacity of ice hockey goaltender helmet protection from puck impacts.

Additionally, Clark et al. (2018) used the same orientation to measure the safety of ice hockey goaltender helmets for ice hockey-specific goaltender concussions. The mechanical headform and neckform are generally used in conjunction with a vertical or horizontal pneumatic impactor to test the goaltender helmets and simulate vertical or horizontal collisions with the ice, other athletes, or wall surfaces in the ice hockey rink.

Figure 8

NOCASE Headform and Mechanical Neckform

Note: The headform is designed to represent the 50th percentile of an adult head and has anatomically correct bone structure and facial features. The neckform is attached to the headform and can be adjusted to modify the tension of the neck. The headform will be equipped with a goaltender helmet.

Research Problem

Despite improvements in helmet technology to mitigate the magnitude of head collisions, concussions remain a common injury for ice hockey players. The complexity and variation of the mechanisms of injury facing ice hockey players have motivated researchers to explore the combined effect of helmet lining materials and human cervical muscle strength in reducing concussion risk for ice hockey goaltenders.

In the case of goaltenders, for example, helmets mitigate short-duration impact accelerations due to falls and projectile head collisions reasonably well; however, the helmets have difficulty reducing head injury risk (HIC) from long-duration impact acceleration caused by shoulder collisions to a goaltender's head (Clark et al., 2020). One potential reason for this outcome is the type of liner material used for constructing goaltender helmets.

Standard VN goaltender helmet liners, for instance, become denser when impacted, causing the rigidity of the liner to increase throughout the impact and dissipate energy less effectively (Jeong et al., 2012). On the contrary, TPU liners dissipate energy more effectively because these liners consist of relatively large air-filled cells that stretch and move together freely (Ramirez & Gupta, 2018). The air-filled cells slowly release air during impacts, and move freely, leading to increased energy dissipation over a longer duration (Ramirez & Gupta, 2018). The use of TPU liner material has been shown to improve the safety of cycling helmets, boxing headgear, and hockey player helmets (Rybak, 2021; McGillivray, 2020); however, it has not been tested for ice hockey goaltenders.

Researchers also believe that cervical muscle strength is a strong indicator of concussion risk, with higher neck strength ratings being associated with decreased risk of concussion (Schmidt et al., 2014). The cervical muscles act to decrease the acceleration caused by an impact (Mihalik et al., 2011). Increasing neck strength will aid in the reduction of acceleration for both short and long-duration impacts (Mihalik et al., 2011).

In the case of goaltenders, it becomes critical to improve their neck strength to reduce concussion risk because the neck needs to support the weight of the head and helmet and maintain a protruding position in front of the body for the goaltender to increase his/her visual range on the movement of the puck while standing in front of the net. The combination of the helmet's weight and the posture of the neck leads to increased stress and fatigue of the cervical spine muscles and, consequently, an increase on the risk of concussive injuries. It is also important to consider that goaltender helmets weigh approximately 1.5 kg. This weight is more than ice hockey player helmets, which weigh around 0.6 kg (Clark et al., 2018). This extra weight also contributes to neck muscle fatigue and increase in the risk of concussion.

Furthermore, studies involving simulations of head collisions and human neck strength levels to assess the risk of concussion require the use of a mechanical headform and neckform. Despite the regular use of mechanical neckforms in helmet testing literature, the neckforms are tensioned to standardized values and not calibrated to human neck strength values to include both male and female populations (Jeffries et al., 2017; Pennock et al., 2021), and to include low to high percentiles of these populations. Finally, human neck strength simulations have not been explored in combination with the use of TPU lining material for ice hockey goaltender helmets as a possible avenue to mitigate concussion risk for ice hockey goaltenders.

Purpose

Based on the gaps in the literature, the purpose of this study was to examine the combined effect of neck strength level and helmet liner type in mitigating the magnitude of the impact accelerations and the risk of concussion for goaltenders during horizontal head collisions at different head locations.

Research Questions

The following research questions guided the purpose of the study:

1. Which helmet liner material (TPU or VN) would absorb more energy in joules (J) per kilogram mass when loaded with compressive and shear forces during static testing?

2. Was there a difference between helmet liner conditions (bare head, TPU, or VN) in mitigating measures of force and acceleration during repeated head impact testing?
3. Would there be a combined effect of neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male) and type of goaltender helmet liner (TPU and VN) on measures of linear acceleration during dynamic head collisions?
4. Would there be an interaction effect between neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, or 80^{th} percentile male) and helmet liner type (TPU or VN) on measures of risk of head injury (HIC) during dynamic head collisions?

Chapter 3 – Methodology

For this study, the researcher addressed each research question using the following instrumentation, procedures, and data analysis.

Instrumentation

Bauer© 960 Goaltender Helmet

The Bauer© 960 XPM goaltender helmet (see Figure 9) was used for this study. This type of goaltender helmet consisted of a shell, backplate, cage, and foam liner held together with glue, screws, and elastic straps. The shell contained a blend of fiberglass, carbon fiber, and Kevlar® materials (Clark et al., 2020). The backplate was made of the same blend of materials as the shell and was attached to the helmet structure using five elastic straps. The straps are used to adjust the fit of the helmet and allowed the goaltender to put on and take off the helmet easily. The cage was constructed with carbon steel wires, and it is attached to the shell with six screws. The foam liner of the helmet consisted of two types of VN foam material. The type-one liner was light brown and ranged between 10 mm and 15 mm thick. This type of liner is the predominant liner type used in the helmet, and it is used in vulnerable areas of the helmet. The type-two liner was dark grey and perforated, and it was used in low-risk areas (Clark et al., 2020).

Figure 9

Bauer© 960 Goaltender Helmet

Note. Bauer© 960 ice hockey goaltender helmet was used in this study. Arrow 1 indicates the type one liner. Arrow 2 shows the type two liner.

3D Printer

A Creality CR-10S three-dimensional (3D) printer (see Figure 10) was used in the goaltender helmet. The Creality CR-10S printer can make customized objects as large as 300 mm x 300 mm x 400 mm. The CR-10S printer has a nozzle diameter of 0.4 mm to print the melted TPU filament and build or print the liner structures layer upon layer. In other words, the CR-10S follows the same process as traditional two-dimensional (2D) printing, with the exception that it stacks layers of TPU material on top of the previous layers to create a 3D structure. This approach allowed for the creation of complex geometric helmet liner shapes that were used in a time-effective manner in the current study. The CR-10S has a precision rating of ±0.1 mm, which leads to reliable prints.

Figure 10

Creality CR-10S Three-Dimensional Printer

Note: Taken from https://3dprintingcanada.com/products/creality-cr-10-smart?variant=40161623998533&ab_version=B&gclid=CjwKCAjwx7GYBhB7EiwA0d8oe_zbak3rBW39Gpdao0rQ2nr-2SA6SWh77QS-T4igSINnCByq_o6P8BoC5FQQAvD_BwE

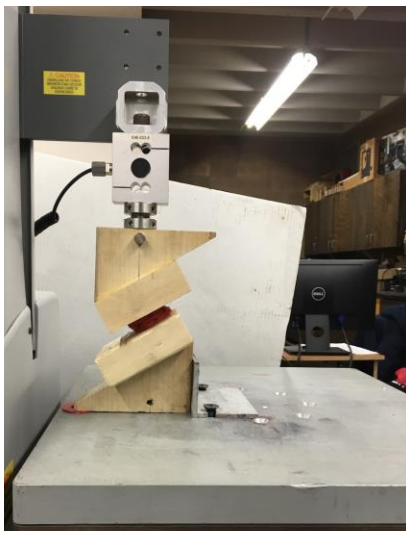
Chatillion ® TCD1100 Force Tester and American Medical Technology Incorporated Force Plate

A Chatillion® TCD1100 force tester instrumented with an American Medical Technology Incorporated (AMTI®) force plate (see Figure 11) was used to statically assess the material properties of the helmet liners (VN foam and 3D printed TPU). The Chatillion® TCD1100 force tester was adjusted with blocks placed at a 30° angle to compress the liner material to measure the shear and compressive forces required to deform the liner. The liner samples were compressed to 5 mm for 15 cycles at a rate of 20 mm/minute against the TCD series load cell and AMTI® force plate to assess the energy absorption of the material.

The AMTI® force plate located at the base of the Chatillion® TCD1100 tester allowed for the shear forces during static testing to be measured. The AMTI® force plate was fixed to a steel plate to minimize vibrations and accurately measure the forces (F_x, F_y, and F_z) in three directions throughout the testing cycle at a sampling rate of 1000 Hz. The force measures from the Chatillion® TCD1100 tester and AMTI® force plate allowed the researcher to determine the amount of compressive force, shear force, and total force, and the amount of vertical (or total) displacement. Such data were used to determine the compressive, shear and total energy absorptions, which were then converted to specific energy absorption by dividing those values by the mass of the material.

Figure 11

Chatillion® TCD1100 Force Tester



Note: The Chatillion® TCD1100 force tester was modified for static testing TPU sample at a 30^{o} angle to compute both compressive and shear forces

Pneumatic Repeated Impactor with AMTI® Force Plate

The pneumatic repeated impactor with AMTI® force plate shown in Figure 12 contained a cylinder with a pneumatic piston that connects to a cast urethane headform assembly weighing 5 kg. The pneumatic repeated impactor moved forward then backward continuously with the use of two limit switches placed under the moving frame that supported the rod and head. When the first limit switch activated the impactor, the air was released, and the head travelled towards the force plate. The frame carrying the head then contacts the second limit switch, which caused the frame to be retracted back to the starting position. Once the frame returned to the starting position, the first limit switch activated again, and the process repeated until the machine turned off.

The repeated impactor can strike a precise location consistently at a specific velocity based on the air pressure in the tank. The air pressure in the tank was measured in psi (pounds per squared inch) using an MGA-100-A digital pressure gauge with an accuracy of ±1% when the machine was activated. The air pressure was calibrated by the researcher for impact velocities ranging from 0.21 m/s to 0.84 m/s.

The velocity of the impact referred to the velocity of the head immediately before it struck the AMTI® force plate. When the mechanical headform struck the AMTI® force plate, the impact force was measured. The AMTI® force plate oriented at an angle of 16.5° from the vertical captured the compression and shear forces involved in the impacts. A piezoelectric accelerometer located on the pneumatic piston measured the acceleration experienced by the head during the impacts.

Figure 12

Pneumatic Repeated Impactor with AMTI® Force Plate

Note. The helmet liner has been taped to the AMTI force plate for repeated impact testing.

NOCSAE Mechanical Headform

A medium-sized NOCSAE headform (see Figure 8) constructed with facial features that represent the 50^{th} percentile of an adult head and weighing 4.90 kg was used to simulate the head collisions for this study. The anthropometric information for the headform can be seen in Table 1. The structure of the NOCSAE headform allows for accurate simulations of the human head's dynamic response when studying protective equipment. The headform is more accurate than metallic headforms and is more durable than cadaver heads, previously used in impact testing (Hodgson, 1975). The NOCSAE headform was used in this study as it aligns with current NOCSAE helmet testing protocols, and it has been used in hockey goaltender helmet testing studies in the past (Clark et al., 2018, 2020).

Table 1

Anthropometric Measurements of the NOCSAE Headform in inches and (mm) (NOCSAE, 2018)

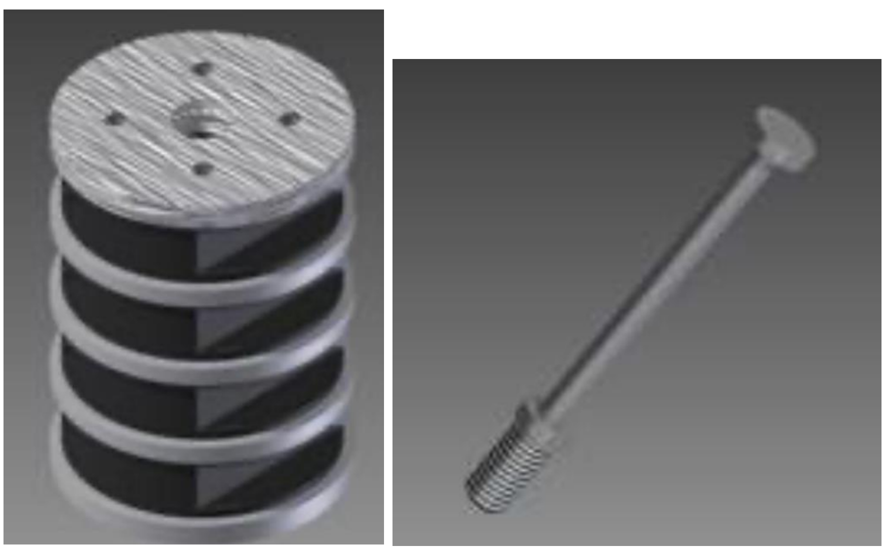
POINTS OF MEASURE	HEADFORM SIZES		
	SMALL 6 5/8”	MEDIUM 7 1/4”	LARGE 7 5/8”
Head breadth	5.63 (143)	5.98 (152)	6.46 (164)
Maximum brow width (frontal diameter)	4.65 (118)	5.20 (132)	5.52 (140)
Ear hole to ear hole (bitragion diameter)	5.24 (133)	5.51 (140)	6.06 (154)
Maximum jaw width (biagonal diameter)	4.13 (105)	4.65 (118)	5.08 (129)
Head length (glabela landmark to back of the head)	7.09 (180)	7.87 (200)	8.15 (207)
Outside eye corner (external canthus) to back of the head	6.22 (158)	6.81 (173)	7.32 (186)
Ear hole (tragion) to back of the head	3.50 (89)	3.86 (98)	4.25 (108)
Earhole to the outside corner of the eye (tranigon to ext. canthus)	2.72 (69)	2.95 (75)	3.07 (78)
Ear hole to the top of the head (tranigon to vertex)	4.72 (120)	5.24 (133)	5.67 (144)
Eye pupil to top of the head	4.13 (105)	4.53 (115)	4.96 (126)
Ear hole to jaw angle (tragion to gonion)	3.31 (84)	3.03 (77)	2.84 (72)
Bottom of the nose to the point of the chin (subnasal to menton)	2.56 (65)	2.80 (71)	3.03 (77)
Top of the nose to the point of the chin (nasion to menton)	4.45 (113)	4.88 (124)	5.39 (137)
Head circumference	21.02 (534)	22.68 (576)	24.17 (614)
Head weight, including mounting interface	9.08 lb (4.12 kg)	10.8 lb (4.90 kg)	13.08 lb (5.93 kg)

Mechanical Neckform

The mechanical neckform used in this study accurately represents the 50th percentile of a human neck in terms of strength and ROM (see Figure 13). The mechanical neckform contains steel plates that alternate with neoprene discs to simulate the intervertebral joints. A large cut-out on the posterior side, as well as a small cut on the anterior side of each neoprene disc allow for the neck to move through the normal cervical ROM. To simulate human neck strength levels during a head impact, a steel cable runs longitudinally through these discs. Loosening or tightening the steel cable running longitudinally through these discs changes the stiffness of the neckform and, consequently, alters the neck strength.

Figure 13

Computer Model of Mechanical Neckform and Mechanical Tensioning Cable



Note. This steel cable runs longitudinally through the center of the neckform and can be tightened or loosened with the use of a bolt on the threaded end.

This neckform has been used in previous helmet testing research (Carlson et al., 2016; Pennock et al., 2021; Zerpa et al., 2017). In these studies, the neckform was calibrated to a standard value of 1.35 N, as well as 30% above and below this value to analyze the influence of the neck compliance characteristic on the dynamic response based on a protocol used by

Hoshizaki (2009). Pennock et al. (2021) modified this protocol by using *z*-statistics to fit a normal distribution of female hockey players to the normal distribution of neck stiffness of this mechanical neck. This study was believed to be the first to establish a connection between human neck strength values and the mechanical neckform instead of using arbitrarily high- and low-neck stiffness ratings (Pennock et al., 2021).

Accelerometers, Sensors, Power Supply, and Software Interfaces

The mechanical headform was instrumented with piezoelectric sensors designed to measure the magnitude of the impact acceleration in three directions (a_x, a_y, and a_z; Jefferies et al., 2017). A PCB© model 482A04 integrated circuit piezoelectric sensor (ICP) amplifier and AD Instruments® PowerLab26T analog to digital converter captured the signals from the accelerometers instrumented in the mechanical headform (Jefferies et al., 2017). The acceleration was measured using a sampling frequency of 20 kHz and was converted to units of g with the use of the arithmetic function from PowerLab® software shown in Equation 2 (Jefferies et al., 2017).

$$g_{(i)} = Ch_{(i)} / 0.0104 \tag{2}$$

where:

$g_{(i)}$ = acceleration value from a respective channel in g;

$Ch_{(i)}$ = channel acquiring the acceleration information from either the X, Y, or Z axis in measures of volts; and

i represents the location of the acceleration value in the data set

The X, Y, and Z directions were combined to create a resultant acceleration channel, as shown in Equation 3, which represented the total magnitude of the impact. A 1000 Hz low-pass filter was applied to the resultant acceleration channel to eliminate the effect of high-frequency noise generated from the helmet vibrations due to impacts.

$$\text{RLA} = \sqrt{\alpha^2{}_x + \alpha^2{}_y + \alpha^2{}_z} \tag{3}$$

where:

RLA = resultant linear acceleration;

α_x = acceleration in the X- direction;

α_y = acceleration in the Y-direction; and

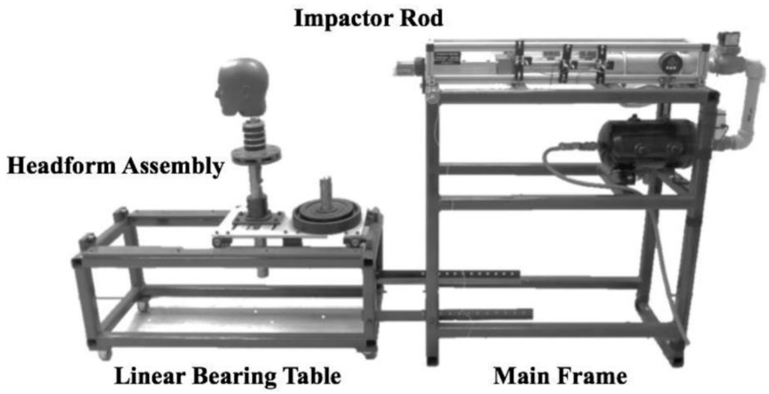
α_z = acceleration in the Z-direction.

Pneumatic Horizontal Impactor

The horizontal impactor used in this study consisted of a main frame and linear bearing table (see Figure 14). Within the main frame, the horizontal impactor contained a compressed air tank, air cylinder, air release valve, and impactor rod (Jefferies et al., 2017). The pressure in the air cylinder propelled the 13.4 kg impact rod along a 0.49 m track by discharging the pressure from the air tank with a control solenoid valve to strike the surrogate headform. Air pressure in the tank was measured with an MGA-100-A digital pressure gauge with an accuracy of ± 1% to generate the desired head impact velocities. The linear bearing table contained a shuttle plate that moved backwards during impacts. This shuttle plate is stopped by rubber bumper blocks at the end of the track. On the shuttle plate, there was an attachment to secure the surrogate neckform in the desired location. The headform attachment plate has five degrees of freedom to position the headform with respect to specific movement impact angles (Jefferies et al., 2017). The degrees of freedom included forward-backward tilt, lateral rotation, forward-backward movement, lateral movement, and up-down movement.

Figure 14

Pneumatic Horizontal Impactor



Jefferies et al. (2017) provided evidence of reliability for this horizontal impactor and conducted 100 impacts to the front, side, and rear impact locations at 40 psi air pressure which relates to an impact speed of 4.39 m/s. The impact data were analyzed using the split-half method between odd and even numbered trials. The results provided evidence of reliability with strong significant intraclass correlation coefficients (ICC) for the front (r = .86, n = 50), side (r = .79, n = 50), and rear (r = .81, n = 50). Jefferies et al. (2017) also compared measures of acceleration for horizontal impact trials to vertical drop impacts with identical parameters to show evidence of validity. The vertical drop impactor was previously validated by Carlson et al. (2016). Jefferies et al. (2017) completed 25 impacts to the front, side, and rear locations of the helmet with the vertical drop impactor at different drop velocities. The 25 impacts in each of these locations were compared over the same impact locations and velocities with the horizontal impactor. The results provided strong evidence of concurrent validity with strong significant intraclass correlations between the measures obtained from both systems to the front (r = .95, n = 25), side (r = .85, n = 25), and rear (r = .88, n = 25) locations. This impactor was used in previous research to measure the risk of brain injury and the capacity of bicycle helmets, boxing headgear, as well as ice hockey player helmets (McGillivray, 2020; Pennock et al., 2021; Rybak, 2021; Zerpa et al., 2020).

Procedures

Static Testing

The TPU and VN liners underwent static testing with the Chatillion® force tester in conjunction with an AMTI® force plate located at the base of the force tester. The researcher analyzed the total, compression, and shear static energy loadings of the helmet liner materials to help with understanding the design and potential modifications of the TPU liner structures. The Chatillion® force tester was fitted with a pair of 30° angled wooden blocks. When the force sensor, connected with the top wooden block and controlled by the Chatillion® force tester, compressed on the liner materials, the 30° wooden blocks produced the compression and shearing effects on the liners. The shearing effect generated a parallel force to the surface of the material (Pennock et al., 2021). The researcher computed both compressive and shear forces because these forces are critical in helmet design and testing to minimize the risk of concussions (Clark, 2015). The force sensor recorded the vertical loading and unloading forces on the liner material, together with the vertical displacements. The bottom wooden block, on the other hand, was connected to an aluminum bracket which was bolted onto the AMTI® force plate. This prevented the liner material and bottom block from sliding when compressed against the AMTI® force plate. The researcher fitted each piece of liner material between the wooden mounting blocks of the machine (see Figure 11).

The Chatillion® TCD1100 force tester acquired the vertical force and displacement data of the material over the 15 cycles of static testing. The researcher set the force capacity of the Chatillion® TCD1100 force tester to 5000 N to prevent damaging the machine. The researcher also set the speed of the Chatillion® TCD1100 force tester to 25 mm per minute, which was slow enough to consider the testing static. A total of 15 cycles were conducted for each sample of liner material corresponding to the front, rear, and side locations of the goaltender helmet. The researcher used the measures of forces and displacements data to compute the total, compression, and shear energy absorptions via the MATLAB® software.

The horizontal force was computed using two force measures from the AMTI® plate. Three force signals, F_X, F_Y, and F_Z, were obtained from the AMTI® force plate in the X, Y, and Z-directions. The X- and Y-force vectors were added together using Equation 4 to compute the horizontal resultant force (F_H). The vertical force vector (F_Z) was simply measured in the Z-direction and could be obtained from the Chatillion® TCD1100 force tester as well.

$$F_H = \sqrt{Fx^2 + Fy^2} \tag{4}$$

where:

F_X = force in the X direction

F_Y = force in the Y direction

Once F_H were calculated, the shear force (T) and compressive force (N) were determined using Equations 5 and 6, respectively.

$$N = F_V \cos\theta - F_H \sin\theta \tag{5}$$

$$T = F_H \cos\theta + F_v \sin\theta \tag{6}$$

where:

$\theta = 30°$

F_v = Force in the vertical direction, or F_Z

The researcher used a MATLAB® script to compute the amounts of total, compressive, and shear energy absorbed by the liners for each of the 15 cycles. Equations 7, 8, and 9 show how total, compression, and shear energy absorptions were determined, respectively.

$$Total\ Energy\ Absorption = \frac{enclosed\ area\ of F_Z\ versus\ total\ displacement\ plot}{area\ under\ the\ loading\ plot\ of\ F_Z\ versus\ total\ displacement} \quad (7)$$

$$Compression\ Energy\ Absorption = \frac{enclosed\ area\ of N\ versus\ total\ displacement\ plot}{area\ under\ the\ loading\ plot\ of\ N\ versus\ compression} \quad (8)$$

$$Shear\ Energy\ Absorption = \frac{enclosed\ area\ of\ T\ versus\ total\ displacement\ plot}{area\ under\ the\ loading\ plot\ of\ T\ versus\ shear\ displacement} \quad (9)$$

The researcher divided the amount of total, compression, and shear energy absorbed by the mass of the liner, in kilograms, to compute the total, compression, and shear specific energy absorption values. Equations 10, 11, and 12 show how total, compression, and shear specific energy absorptions were calculated, respectively. During this phase of the testing, a total of 12 different TPU designs were tested and compared to the standard goaltender liner. These TPU designs varied in shape, density, and structure, but were identical in overall size.

$$Total\ Specific\ Energy\ Absorption = \frac{Total\ Energy\ Absorption}{Mass\ of\ Sample} \quad (10)$$

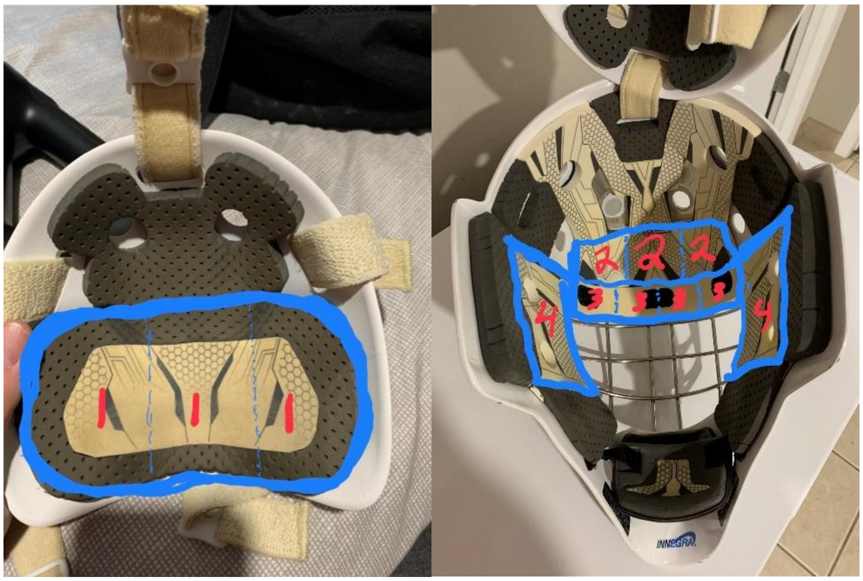
$$Compression\ Specific\ Energy\ Absorption = \frac{Compression\ Energy\ Absorption}{Mass\ of\ Sample} \quad (11)$$

$$Shear\ Specific\ Energy\ Absorption = \frac{Shear\ Energy\ Absorption}{Mass\ of\ Sample} \quad (12)$$

The researcher selected the TPU design that absorbed the most energy in J/kg, which was labelled as TPU_8. The researcher used TPU_8 to create four different sizes of TPU insert liners to match the standard helmet liner configuration for each location to conduct the dynamic testing. Figure 15 depicts the four locations and Table 2 outlines the mass and dimensions of the TPU sample used for each location.

Figure 15

TPU Locations



Note. Liner material specifications: 1- the back location, 2- the upper head location, 3-the headband location, and 4-the side location.

Table 2

Summary Information for Samples Used

Location	Mass (grams)	Length (mm)	Width (mm)	Depth (mm)	Number Used
Back	3.13	27.5	30	10	18
Upper Head	3.16	25	30	12	12
Headband	4.04	30	30	12	8
Side	2.39	25	25	10	12

Note. The TPU liner and standard liner are identical in length, width and depth.

Repeated Dynamic Impact Testing for Helmet Liners

The repeated impact testing involved the use of the pneumatic repeated impactor with AMTI® force plate and a mechanical surrogate headform mounted to the impacting rod and positioned to contact the desired location on the AMTI® force plate. Three liner conditions were assessed during the repeated impact trial including the bare head, TPU and standard. The bare head condition was selected to simulate the impact for an individual's head not wearing a helmet. The standard and TPU liners were selected to simulate the impact of an individual's head with just the liner material. The researcher attached the liner material to the force plate using tape on the edge of the liner material sample to hold it in place and ensure the mechanical head would impact the center of the material (see Figure 11).

The researcher selected 26 different impact velocities in the repeated impact trials to simulate the head collision and assess the performance of the liners during low-speed repeated impacts. The researcher adjusted the air pressure in increments of 1 psi, with corresponding impact velocities beginning at 0.2142 m/s and increasing at an average rate of 0.02 m/s to the highest velocity of 0.8642 m/s as shown in Table 3. Velocities greater than 0.87 m/s resulted in excessive vibration and noisy signal, and the recorded data were discarded. Three consecutive impacts were conducted for each liner condition across the 26 impact velocities. The researcher computed the average impact force (N) and acceleration (m/s^2) of the three trials for each helmet liner condition across the 26 impact velocities.

Table 3

Repeated Impactor Tank Pressure and Corresponding Velocity

	Pressure (psi)	Velocity (m/s)
1	10	0.2142
2	11	0.2402
3	12	0.2662
4	13	0.2922
5	14	0.3182
6	15	0.3442
7	16	0.3702
8	17	0.3962
9	18	0.4222
10	19	0.4482
11	20	0.4742
12	21	0.5002
13	22	0.5262
14	23	0.5522
15	24	0.5782
16	25	0.6042
17	26	0.6302
18	27	0.6562
19	28	0.6822
20	29	0.7082
21	30	0.7342
22	31	0.7602
23	32	0.7862
24	33	0.8122
25	34	0.8382
26	35	0.8642

Dynamic Testing Simulating Goaltenders' Head Collisions

The researcher performed the dynamic testing of the helmets using a modified version of the NOCSAE pneumatic ram test method for protective headgear and face guards (NOCSAE, 2018). Before conducting the dynamic head impact simulations, the researcher mounted the helmet to be tested on the NOCSAE headform and fitted it according to manufacturing instructions. The helmet was placed over the headform with the chin of the headform sitting securely in the chin cup of the helmet. The straps of the helmet were adjusted to the appropriate tension to hold the helmet in place during testing. The NOCSAE headform was connected to a mechanical neckform and it was positioned on the horizontal impactor to strike the front, rear, and side impact locations according to the NOCSAE protocol (NOCSAE, 2018).

The NOCSAE protocol was chosen for the current study because it is designed to provide reliable and repeatable measurements of linear acceleration experienced by the surrogate headform (NOCSAE, 2018). The dynamic testing consisted of horizontal impacts including different neck strength levels to represent male and female populations (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male), two different goaltender helmets (TPU and standard), and three impact locations (front, side, and back) across 18 different impact velocities ranging from 2.01 m/s to 5.13 m/s.

The velocity values shown in Table 4 simulated the goaltender ice hockey players using a similar approach as that by Jefferies et al (2017). Before implementing the horizontal impact testing protocol for each type of goaltender helmet, the researcher tensioned the mechanical neckform to the appropriate human neck strength level. The researcher re-torqued the neck when changing impact parameters.

Table 4

Tank Pressure and Corresponding Impact Velocity (Jeffries et al, 2017).

	Pressure (psi)	Impact Velocity (m/s)
1	24	2.01
2	26	2.14
3	28	2.42
4	30	2.62
5	32	2.83
6	34	3.12
7	36	3.25
8	38	3.47
9	40	3.61
10	42	3.64
11	44	3.86
12	46	3.94
13	48	4.11
14	50	4.26
15	52	4.48
16	54	4.56
17	56	4.65
18	58	5.13

Neck Strength. The researcher tensioned the mechanical neckform to accommodate the four neck strength levels addressed in the study. The female and male neck strength values implemented in this study were secondary data obtained from the studies conducted by Pennock (2018) and Broennle (2011), respectively. It is important to note that both studies determined the normative ranges of neck strength for ice hockey players in general and, therefore, were not specific to ice hockey goaltenders.

Similar to Pennock's (2018) methodology, a force gauge, torque wrench, and neck torque tensioning apparatus (see Figure 16) were used. The force gauge was attached perpendicularly to the end of the torque wrench arm to accurately measure the force applied to the bolt and add the required stiffness to simulate the cervical neck strength for the 30^{th} and 50^{th} percentile for females and the 50^{th} and 80^{th} percentile for males as shown in Table 5.

The tensioning apparatus ensured that the force was applied perpendicular to the wrench. This approach eliminated the possibility of error due to rotation of the mechanical head during the tensioning procedure. If there was a situation where an impact caused the helmet to shift or loosen, the researcher readjusted the neck strength setting within the subset of impacts.

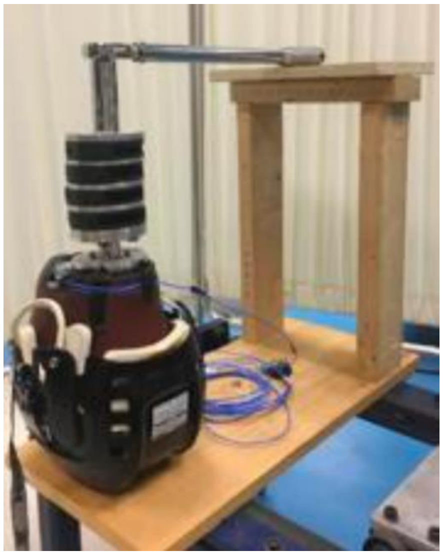
Table 5

Neck Strength (in Newtons) Data for Hockey Players

Gender	30th Percentile	50th Percentile	80th Percentile	SD
Female	66.83	76.01	88.81	15.52
Male	154.13	185.75	225.23	46.95

Figure 16

Neckform Torque Tensioning Apparatus

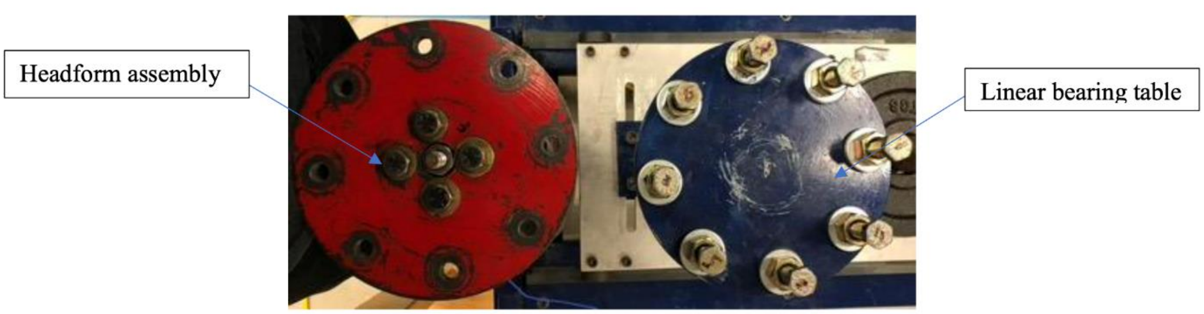


Note. Custom built tensioning apparatus used to accurately torque the longitudinal bolt of the mechanical neckform.

The headform and neckform were mounted to the linear bearing table of the horizontal impactor with the use of the mounting plates, which were bolted together as depicted in Figure 17. The helmet was then fitted to the headform using the manufacturer's instructions. The helmet was correctly aligned to the impacting rod for each helmet location being tested.

Figure 17

Steel Mounting Plates



Note. Steel mounting plates for attaching the headform assembly to the liner bearing table.

Impact Locations. The NOCSAE standards require the use of six impact locations (side, rear boss NC, rear boss CG, rear, front boss, and front). For the purposes of this study, the front, side, and rear locations were selected for impact trials. The three locations were selected because concussions occur frequently for ice hockey goaltenders from impacts in these locations (Clark, 2015). For the different impact locations, the head and neck were adjusted on the linear bearing table. Figures 19, 20, and 21 display the impact locations (side, rear, and front), respectively, as represented by NOCSAE (2018). Impacts were conducted for 18 different velocities ranging from 2.01 m/s to 5.13 m/s as shown in Table 4 for the front, side, and rear locations, respectively. Furthermore, the specific helmet impact locations when fitted to the surrogate headform were determined in accordance with NOCSAE (2018) standards for a medium sized headform as specified in Table 6.

Table 6

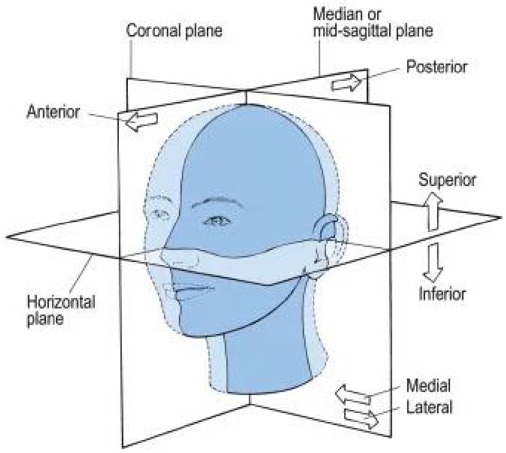
Impact Location Specifications for a Medium-sized Headform

Impact Location	α	β	Z axis relative to basic plane	Y axis
Side	7°	-90°	+60 mm	On the coronal plane
Rear	7°	-180°	+60 mm	64 mm posterior to the coronal plane
Front	7°	0°	+73 mm	On the midsagittal plane

Note. The starting position should be with the nose of the headform pointing towards the impact arm and all angles in Table 6 are relative to that position. The positive direction for α angle is tilting forward towards the impactor, and the positive direction for β angle is clockwise. The anatomical planes are outlined in Figure 18. The specific impact locations can be seen in Figures 19, 20 and 21.

Figure 18

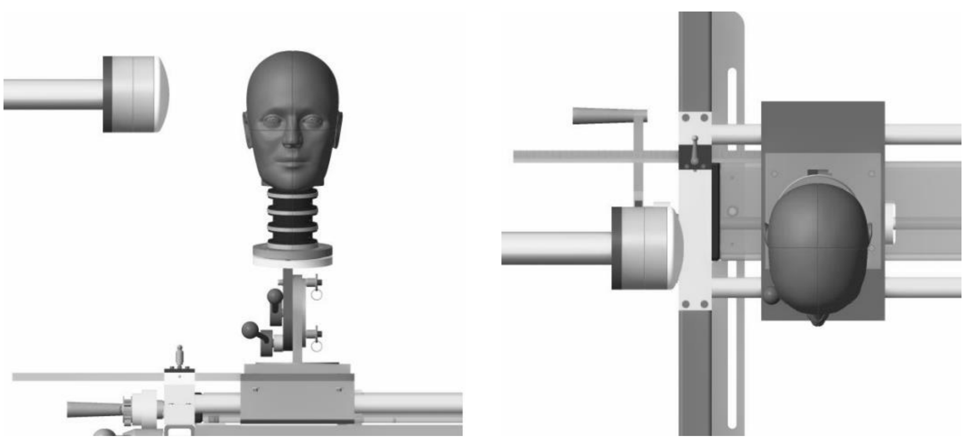
Anatomical Planes in Relation to the Head



Note. Anatomical planes of the human body (Rohen et al., 2015).

Figure 19

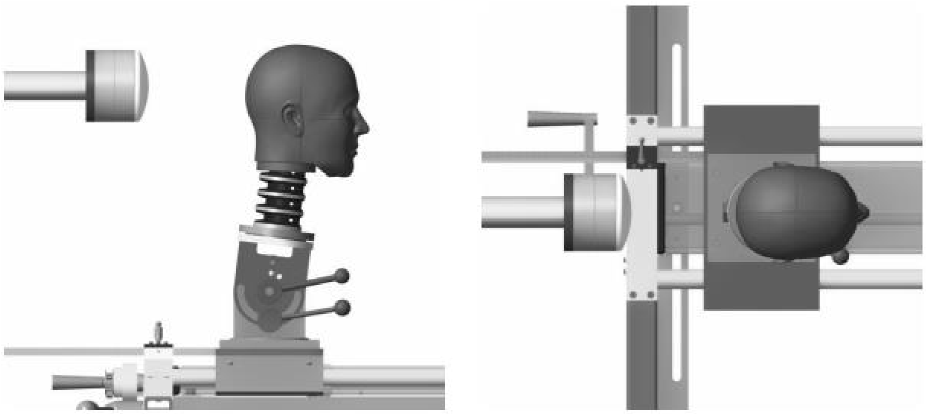
Side Impact Location



Note. Side impact location (NOCSAE, 2018). Adapted from "Standard pneumatic ram test method and equipment used in evaluating the performance characteristics of protective headgear and faceguards," NOCSAE, 2018, p. 3. Copyright by NOCSAE (2018).

Figure 20

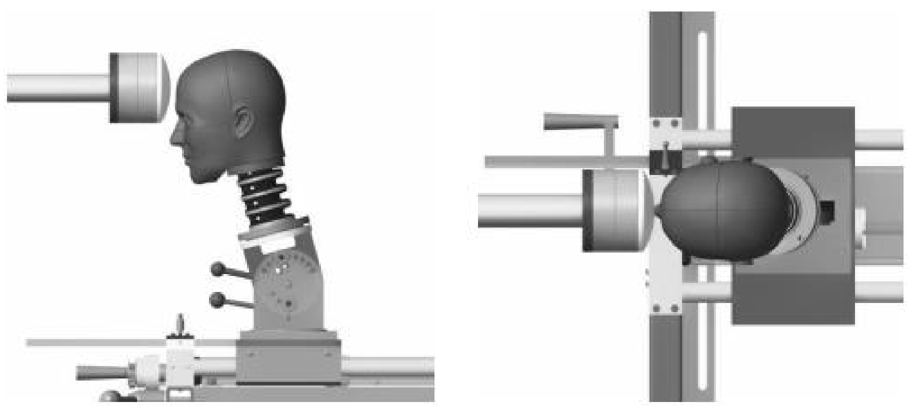
Rear Impact Location



Note. Rear impact location (NOCSAE, 2018). Adapted from "Standard pneumatic ram test method and equipment used in evaluating the performance characteristics of protective headgear and faceguards," NOCSAE, 2018, p. 3. Copyright by NOCSAE (2018).

Figure 21

Front Impact Location



Note. Front impact location (NOCSAE, 2018). Adapted from "Standard pneumatic ram test method and equipment used in evaluating the performance characteristics of protective headgear and faceguards," NOCSAE, 2018, p.3. Copyright 2018 by NOCSAE.

Linear Acceleration. Linear accelerations in the X, Y, Z directions were collected for each impact using LabChart® via the accelerometers positioned in the headform at the frequency of 20 kHz. The resultant linear acceleration (RLA) was calculated within LabChart® using Equation 3.

Risk of Injury. The HIC was calculated with Equation 1. To ensure the maximum HIC value was obtained for each set of data, a MATLAB® script was used to determine which period of t_2-t_1 resulted in the largest outcome value.

Dependent and Independent Variables

For the purposes of static testing, the independent variable was the type of helmet liner used (TPU and standard), and the dependent variable was the specific energy absorption. For the repeated impact testing, the independent variable was type of helmet liner (bare head, TPU, and standard) and the dependent variables were maximum force and acceleration. For the dynamic testing simulating goaltenders' head collisions, the independent variables were neck strength, impact location, and helmet liner type, while the dependent variables were peak resultant linear acceleration (PRLA) and HIC.

Data Analysis

Static Testing

The researcher addressed the first research question of this study by computing the specific energy absorption from each helmet liner type during static testing across 15 trials. This computation entailed dividing the energy absorbed by the mass of the material in kilograms for measures of compressive, shear, and total energy in Jules. The researcher used the mean values and standard deviations across trials 2 through 15 to compare the type of helmet materials on specific energy absorption.

Repeated Dynamic Testing for helmet liners

The researcher addressed the second question of this study by using repeated measure ANOVAs to determine differences in the independent variable of helmet liner condition (bare head, TPU, and standard) for each of the dependent variables (maximum force in newtons and acceleration in g's) respectively. In cases where significant differences were found, Tukey's post-hoc analyses were conducted to determine significant differences in liner condition.

Dynamic Testing Simulating Goaltenders' Head Collisions

The researcher addressed the third and fourth research questions using two-way mixed factorial ANOVAs with helmet type (standard and TPU) as a repeated factor and neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male) as an independent factor for measures of acceleration and risk of head injury for each helmet location (front, side, and back), separately. Helmet type was represented as different lines on the plot and neck strength level was represented on the horizontal axis. In situations where a significant interaction effect occurred, simple main effect analysis was conducted using ANOVAs and t-tests to help explain the interaction. If no interaction effect occurred, the researcher examined the effect of helmet type and neck strength level separately for measures of acceleration and risk of head injury. Tukey's post-hoc analysis was conducted for pair mean comparisons.

Post-hoc Exploration

Based on the results of the dynamic testing phase, the researcher decided to explore the trends that emerged between PRLA and HIC. The researcher used risk of injury curves to compare the likelihood of receiving a concussion based on PRLA and HIC at a given impact velocity in m/s.

Chapter 4 – Results

Static Testing

The first research question of this study was addressed by comparing different TPU liner designs and a standard VN goaltender helmet liner in terms of their capability to absorb energy per kilogram (kg) mass when loaded with compressive and shear forces during static testing. The TPU liner designs and VN liner sample varied in shape, structure, and density but were identical in overall length, width, and height. Table 7 outlines the samples the researcher analyzed in terms of mass, total energy absorbed, specific energy absorption, and percentage of energy absorbed.

Based on this analysis, the TPU_8 liner design was selected for further comparisons with the VN standard liner. The researcher selected the TPU_8 liner design because it produced the greatest specific energy absorption value compared to the other TPU liner designs, as shown in Table 7.

Table 7

Preliminary Static Testing Summary

Sample	Mass (g)	Absorbed Energy (N.mm)	Specific Energy Absorption (J/kg)	Percent Energy Absorbed (%)
Standard	1.40	46.10	32.93	63
TPU1	4.96	227.90	45.73	40
TPU2	4.73	277.70	58.74	43
TPU3	3.79	305.30	80.51	49
TPU4	4.17	320.30	76.81	48
TPU3 (35%)	4.19	268.90	64.24	47
TPU4 (35%)	3.82	318.20	83.32	47
TPU_9	2.10	105.70	50.26	47
TPU_10	2.10	104.50	49.69	48
TPU_11	2.68	157.80	58.92	50
TPU_12	2.68	148.10	55.30	50
TPU_7	3.48	417.80	120.09	49
TPU_8	2.99	372.10	124.16	48

Note. The standard sample contained the liner material from a goaltender helmet.

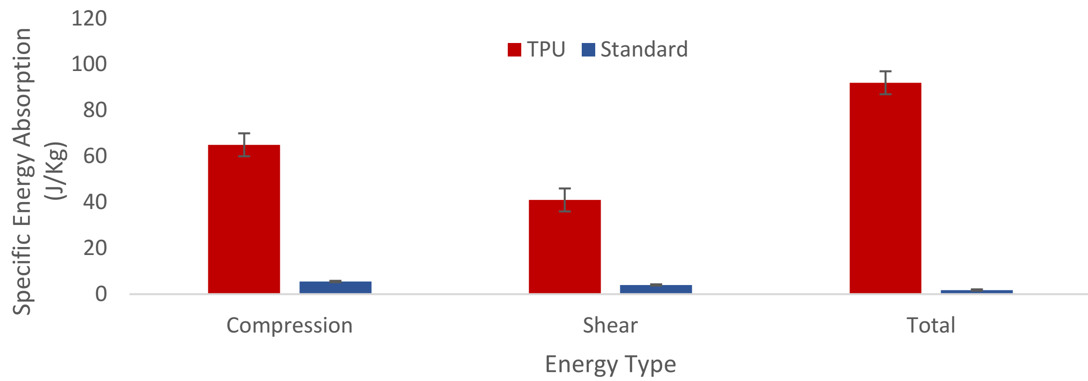
The researcher described the static energy results using means, standard deviations, and bar graphs. A summary table of the descriptive statistic information for all the liner samples used in this study can also be found in Appendix A.

Front Location Static

The TPU liner sample for the headband location had a mass of 4.04 grams, and the standard liner had a mass of 1.87 grams. The TPU liner for the headband location absorbed more compressive energy per kg (M = 65.80 J/kg, SD = 5.23) than the standard liner (M = 5.56 J/kg, SD = 0.22) over 14 trials. The TPU liner also absorbed more shear energy per kg (M = 41.53 J/kg, SD = 3.43) than the standard liner (M = 4.15 J/kg, SD = 0.15). Lastly, the TPU absorbed more total energy per kg (*M* = 91.82 J/kg, *SD* = 7.45) than the standard liner (*M* = 8.61 J/kg, *SD* = 0.31). Figure 22 illustrates the differences in energy absorption per kg between the headband TPU liner sample and the standard helmet liner.

Figure 22

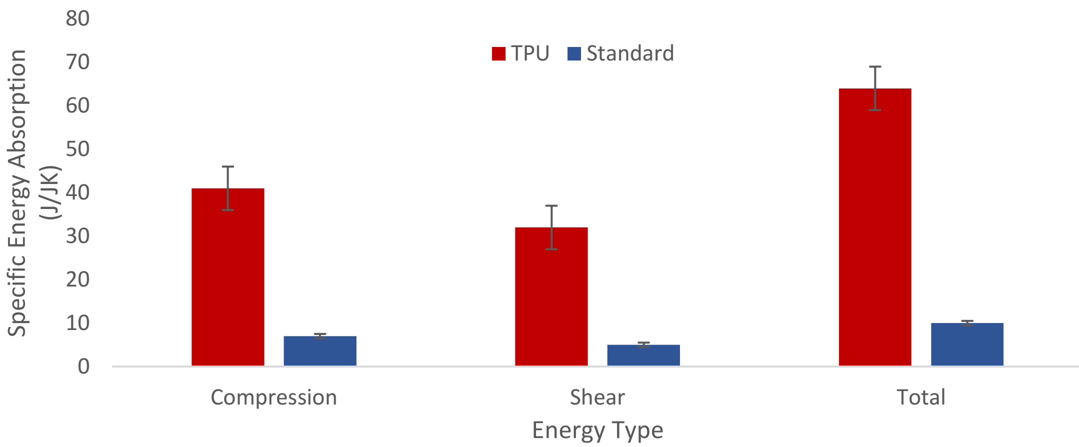
Headband Location for TPU vs. Standard Liner



The results also showed differences in the upper headband location between the TPU and standard liner. The TPU liner sample for the upper head location had a mass of 3.13 g, and the standard liner had a mass of 1.57 g. The TPU liner absorbed more compressive energy per kg for the upper head location (*M* = 41.14 J/kg, *SD* = 4.19) than the standard liner (*M* = 7.07 J/kg, *SD* = 0.29) over 14 trials. Similarly, the TPU absorbed more shear energy per kg (*M* = 32.14 J/kg, *SD* = 3.36) than the standard liner (*M* = 5.29 J/kg, *SD* = 0.21). Ultimately, the TPU absorbed more total energy per kg (*M* = 64.53 J/kg, *SD* = 6.67) than the standard liner (*M* = 10.81 J/kg, *SD* = 0.43). Figure 23 depicts the differences between the two helmet liners in energy absorption per kg for the upper headband location.

Figure 23

Upper Head Location for TPU Versus Standard Liner

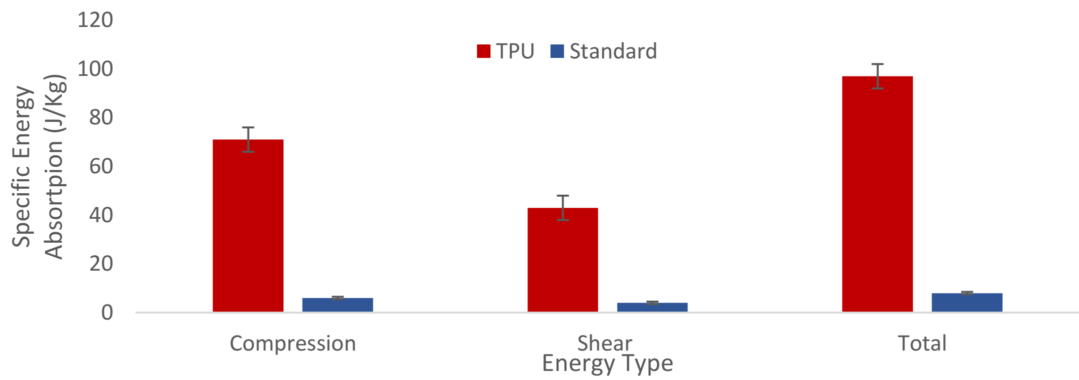


Back Location Static

The TPU liner for the back location of the goaltender helmet had a mass of 3.12 grams, and the standard liner had a mass of 1.28 grams. The TPU liner absorbed more compression energy per kg (M = 70.92 J/kg, SD = 5.97) than the standard liner (M = 5.95 J/kg, SD = 0.23) over 14 cycles. Similarly, the TPU liner absorbed more shear energy per kg (M = 43.86 J/kg, SD = 4.07) than the standard liner (M = 4.12 J/kg, SD = 0.14). Lastly, the TPU absorbed more total energy per kg (M = 97.92 J/kg, SD = 8.67) than the standard liner (M = 8.72 J/kg, SD = 0.30). Figure 24 compares the TPU and standard liners based on specific energy absorption per kg.

Figure 24

Back Location for TPU vs. Standard Liner

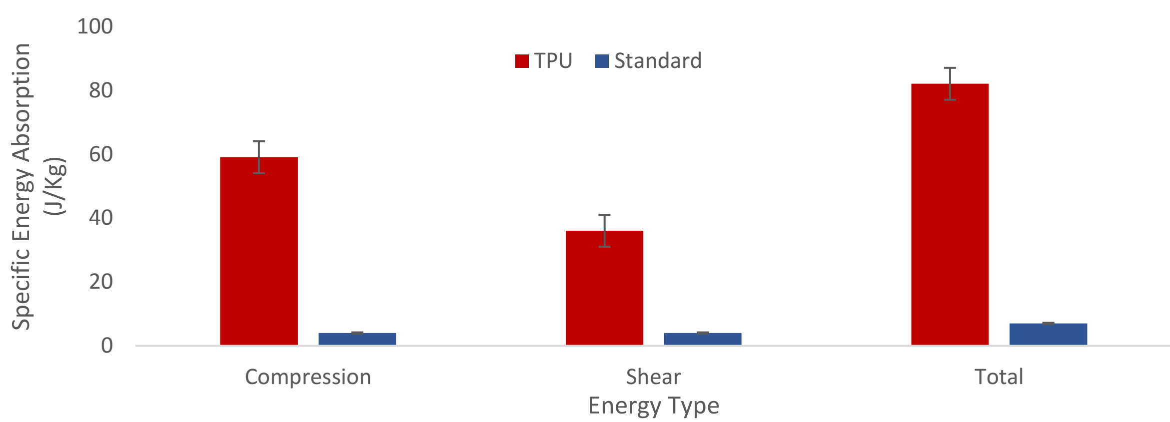


Side Location Static

The TPU liner for the side location of the goaltender helmet had a mass of 2.39 grams compared to the 0.97 grams of the standard liner. The TPU liner absorbed more compression energy per kg (M = 59.38 J/kg, SD = 4.71) than the standard liner (M = 4.71 J/kg, SD = 0.21). Similarly, the TPU absorbed more shear energy per kg (M = 36.84 J/kg, SD = 3.47) than the standard liner (M = 3.84 J/kg, SD = 0.14). Additionally, the TPU absorbed more total energy per kg (M = 82.12 J/kg, SD = 7.37) than the standard liner (M = 7.16 J/kg, SD = 0.29). Figure 25 compares the TPU liner and the standard liner based on specific energy absorption per kg.

Figure 25

Side Location for TPU vs. Standard Liner



Repeated Impact Testing for the Helmet Liners

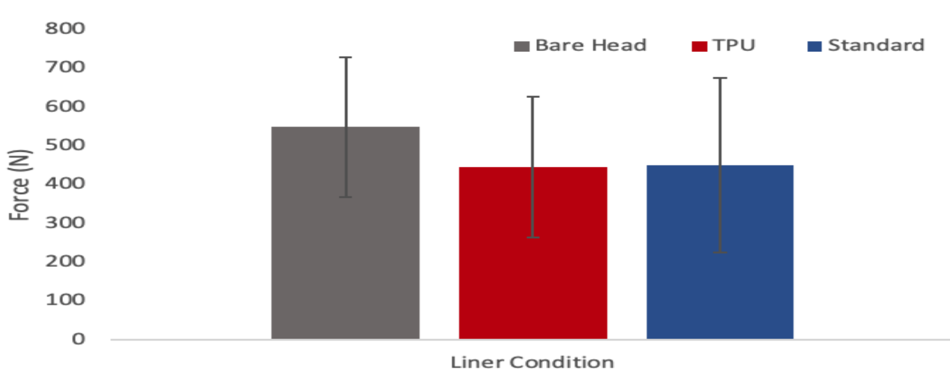
The researcher addressed the study's second research question by comparing the TPU and standard liners against bare head collisions to examine the capability of the liners to mitigate maximum impact force and acceleration during repeated head impacts.

Measures of Force

The results from the repeated measures ANOVA revealed statistically significant differences between helmet liner conditions for the measures of force with a large effect size, F (2, 25) = 107.43, $p < 0.001$, $\eta^2 = .707$. A Tukey's post hoc analysis revealed statistically significant differences between the bare head (M = 548.83 N, SD = 180.85) and the TPU liner (M = 445.87 N, SD = 185.52) at $p < 0.001$. There was also a statistically significant difference between the bare head (M = 548.83 N, SD = 180.85) and the standard helmet liner (M = 450.67 N, SD = 226.69) at $p < 0.001$. The post-hoc analysis, however, revealed no statistically significant differences between TPU and standard liners for measures of force at $p < 0.05$. Figure 26 outlines the significant differences between liner conditions.

Figure 26

Force Measures for Each Liner Condition

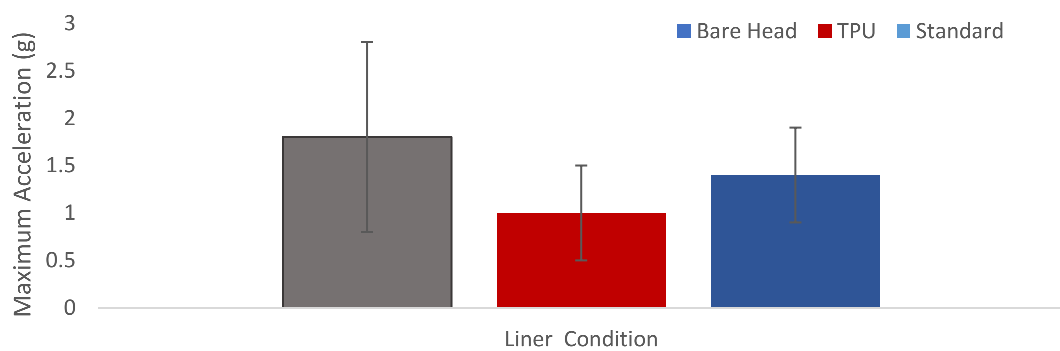


Measures of Acceleration

The results from the repeated measure ANOVA also revealed statistically significant differences for the measures of maximum resultant linear acceleration with a large effect size (in g, the gravitational acceleration) with respect to the three conditions, $F(2, 25) = 5.32$, $p = .008$, $\eta^2 = .176$. The Tukey's post hoc analysis, however, revealed statistically significant differences between the bare head ($M = 1.78$ g, $SD = 1.29$) and the TPU liner ($M = 0.98$ g, $SD = 0.49$) at $p = .025$. The results revealed no statistically significant differences between bare head and standard or TPU and standard at $p < .05$. Figure 27 illustrates the significant differences between the liner conditions.

Figure 27

Maximum Resultant Linear Acceleration for Each Liner Condition



Dynamic Testing Simulating Goaltenders' Head Collisions for Measures of Linear Acceleration

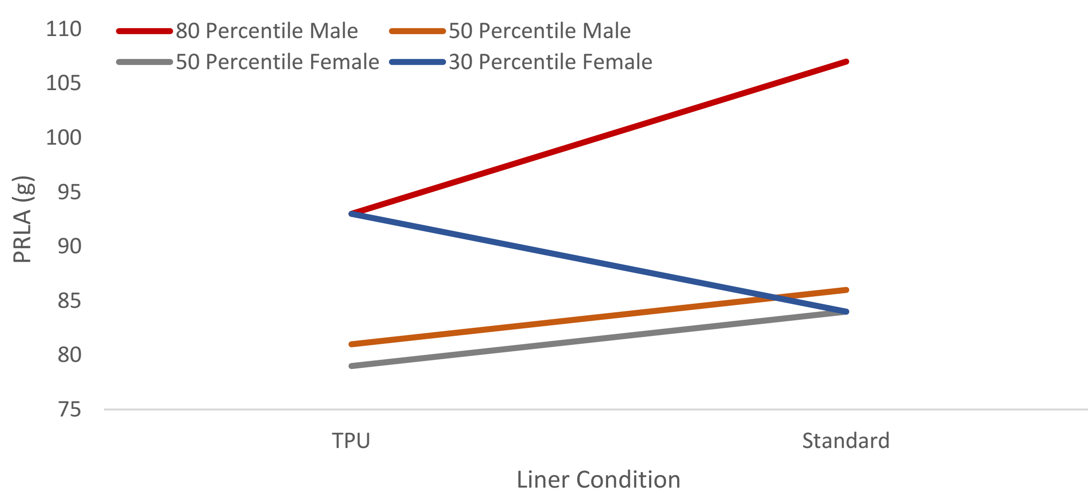
The researcher addressed the third research question by examining the interaction effect between neck strength level (30th percentile female, 50th percentile female, 50th percentile male, and 80th percentile male) and type of goaltender helmet liner (TPU and standard) for measures of linear acceleration during dynamic testing. The results are presented separately for each impact location (front, side, and back). A descriptive statistics summary table for measures of peak resultant linear acceleration (PRLA) regarding all helmet liners tested in this study is also provided in Appendix B.

Front Location PRLA

The two-way mixed factorial ANOVA results revealed a statistically significant interaction effect between neck strength level and helmet liner type for measures of PRLA with a large effect size, as shown in Figure 28 for the front location, F (3, 68) = 830.305, p = .001, η^2 = .205.

Figure 28

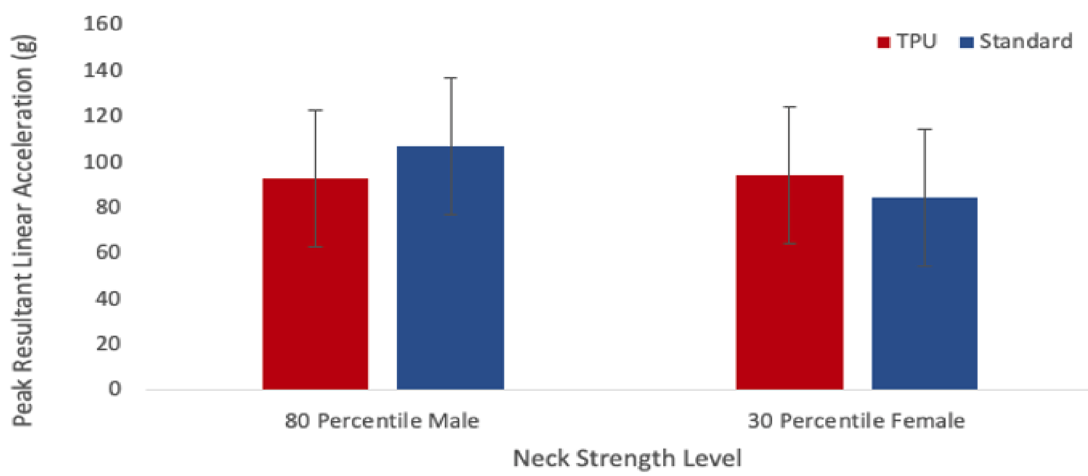
Interaction Between Neck Strength Level and Helmet Liner Type for Front Location



When examining the interaction effect by using a simple main effect analysis, the repeated measure t-tests revealed statistically significant differences between the TPU liner (M = 93.64 g, SD = 28.49) and standard liner (M = 107.34 g, SD = 42.32) for the neck strength level of the 80th percentile male with a large effect size, t (17) = -3.09, p = .007, d = -0.729, CI = [-1.24, -0.199]. The results also revealed statistically significant differences between the TPU liner (M = 93.73 g, SD = 34.83) and standard liner (M = 84.24 g, SD = 36.95) for the neck strength level of the 30th percentile female with a medium effect size, t (17) = 2.705, p = .015, d = 0.638, CI = [0.121, 1.14]. Figure 29 illustrates the differences between the helmet liners for the 80th percentile male and the 30th percentile female. To further explain the interaction, one-way ANOVAs were conducted to compare the neck strength levels for each helmet liner type separately, but no statistically significant differences were found.

Figure 29

Front Location PRLA Differences



Side Location PRLA

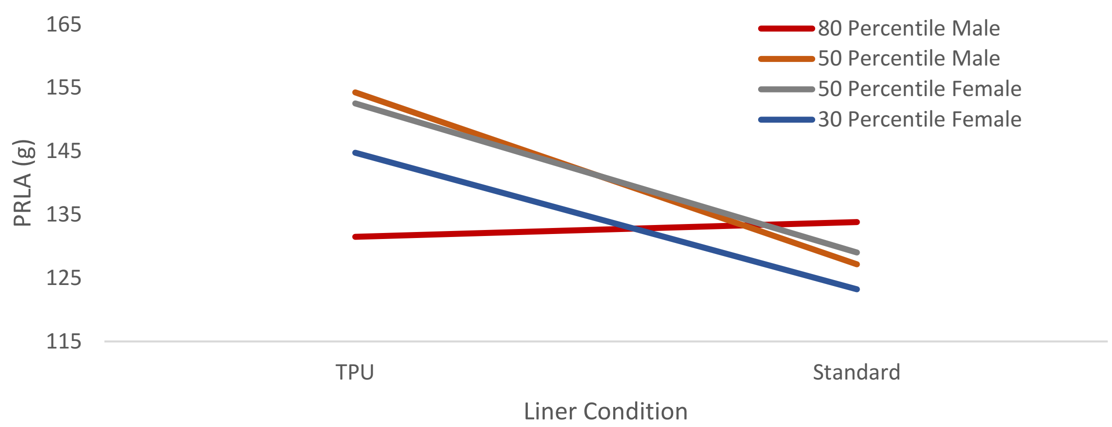
The results from the two-way ANOVA for the side location revealed no statistically significant interaction effect between neck strength level and helmet liner type for measures of PRLA, F (3, 68) = 2.33, $p > .05$. The results also revealed no statistically significant main effects for helmet liner type t (71) = -0.92, $p > .05$ and neck strength level F (3, 68) = 0.13, $p > .05$.

Back Location PRLA

The results from the two-way ANOVA revealed a statistically significant interaction effect between neck strength level and helmet liner type for measures of PRLA with a large effect size, F (3, 68) = 10.00, $p < .001$, $\eta^2 = .306$. Figure 30 illustrates the interaction effect.

Figure 30

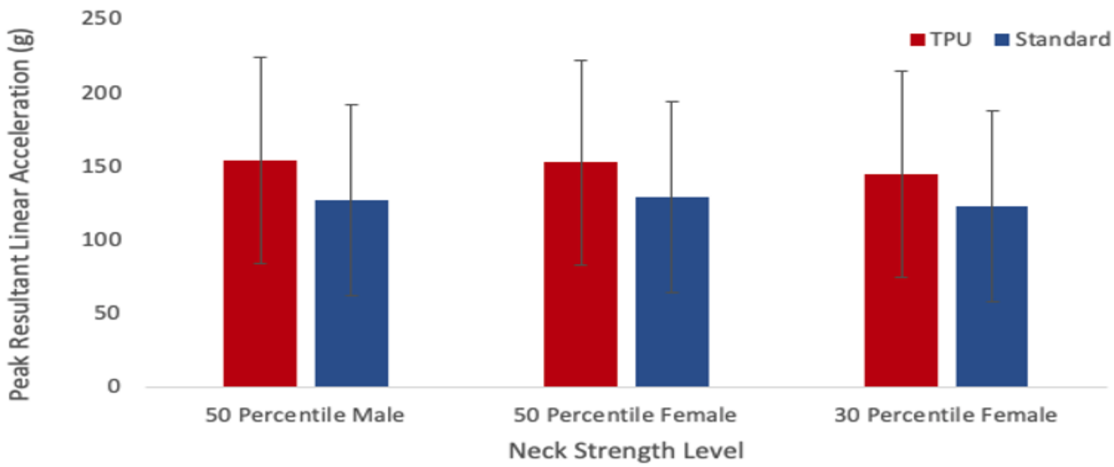
Back Location Interaction Effect



When examining the interaction effect by conducting a simple main effect analysis, the repeated measure *t*-tests revealed statistically significant differences between the TPU liner (M = 154.28 g, SD = 127.17) and standard liner (M = 127.17 g, SD = 57.94) for the neck strength level of the 50th percentile male with a large effect size, t (17) = 6.426 p < .001, d = 0.82, CI = [0.82, 2.19]. The *t*-test analysis also revealed statistically significant differences between the TPU liner (M = 152.54 g, SD = 65.47) and standard liner (M = 129.03 g, SD = 71.19) for the 50th percentile female with a large effect size, t (17) = 6.927, p <.001, d = 1.63, CI = [0.909, 2.34]. Furthermore, statistically significant differences were found between the TPU liner (M = 144.76 g, SD = 73.96) and standard liner (M = 123.23 g, SD = 61.85) for the neck strength level of the 30th percentile female with a large effect size, t (17) = 5.002, p <.001, d = 1.18, CI = [0.56, 1.78]. Figure 31 depicts the differences in PRLA for each neck strength level. To further explain the interaction effect, one-way ANOVAs were conducted separately across neck strength levels for each helmet liner type; however, no statistically significant differences were found.

Figure 31

Side Location PRLA Differences by Neck Strength Level



Dynamic Testing Simulating Goaltenders' Head Collisions for Risk of Injury Measures

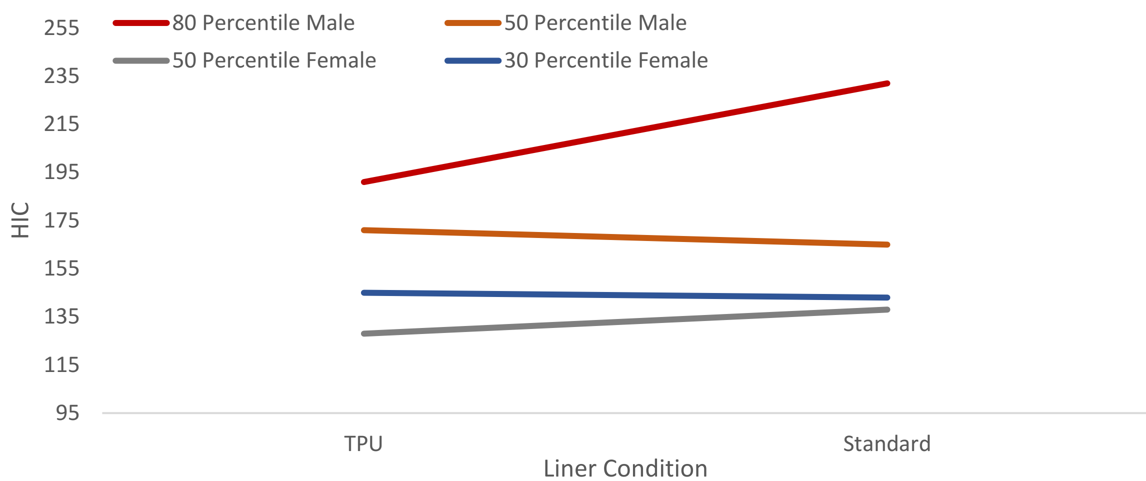
The researcher addressed the fourth research question of the study by examining the interaction effect between neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male) and the type of goaltender helmet liner (standard and TPU) for measures of HIC computed from the measures of linear impact accelerations during the dynamic head collisions. The researcher analyzed each impact location (front, back, and side), respectively.

Front Location HIC

The two-way mixed factorial ANOVA results revealed a statistically significant interaction effect between neck strength level and helmet liner type for measures of HIC with a large effect size, F (3, 68) = 4.213, p < .009, η^2 = 0.157. The interaction effect is shown in Figure 32.

Figure 32

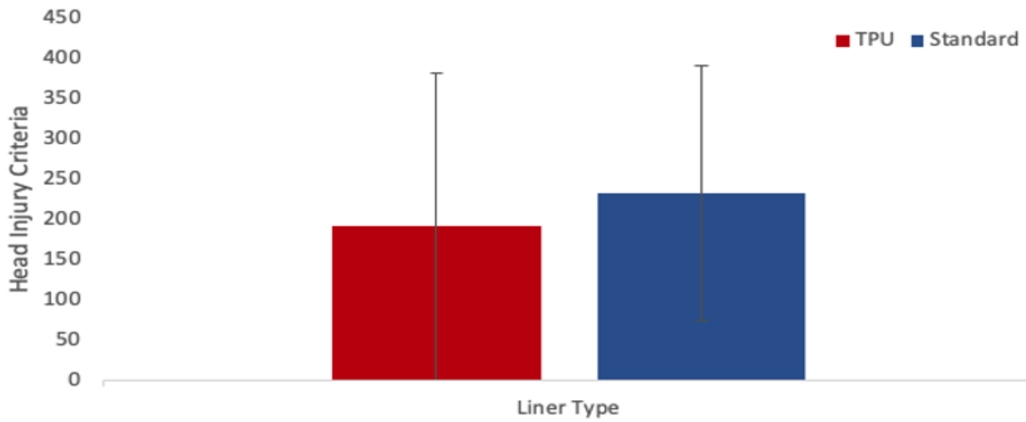
Interaction Effect for Front Location



When examining the interaction effect between liner type and neck strength level by using a simple main effect analysis, the *t*-tests for repeated measures revealed statistically significant differences between the TPU liner (M = 191.06, SD = 114.77) and the standard liner (M = 232.16, SD = 158.28) for the 80th percentile male with a medium effect size, t (17) = -3.17, p = .006, d = -0.75, CI = [-1.26, 0.214]. Statistically significant differences between liner types for HIC measures are illustrated in Figure 33 for the 80th percentile male neck strength level. To further explain the interaction, one-way ANOVAs were conducted to examine the effect of neck strength level for each helmet liner type separately, and no statistically significant differences were found for the measures of HIC at the front location.

Figure 33

Front Location 80th Percentile Male HIC Differences

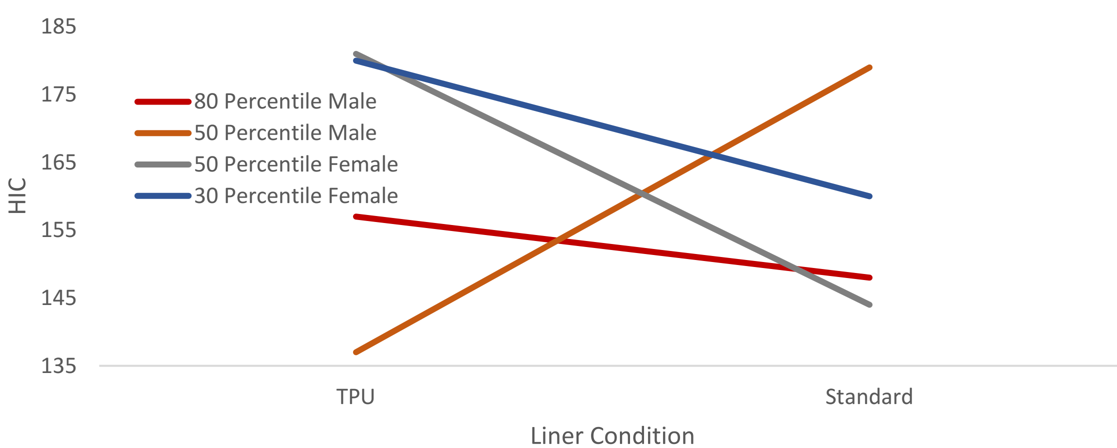


Side Location HIC

The results of the two-way mixed factorial ANOVA revealed a statistically significant interaction effect between neck strength level and helmet liner type for measures of the HIC measures with a large effect size, F (3, 68) = 3.961, p = .012, η^2 = 0.149, as shown in Figure 34.

Figure 34

Interaction Effect for Side Location

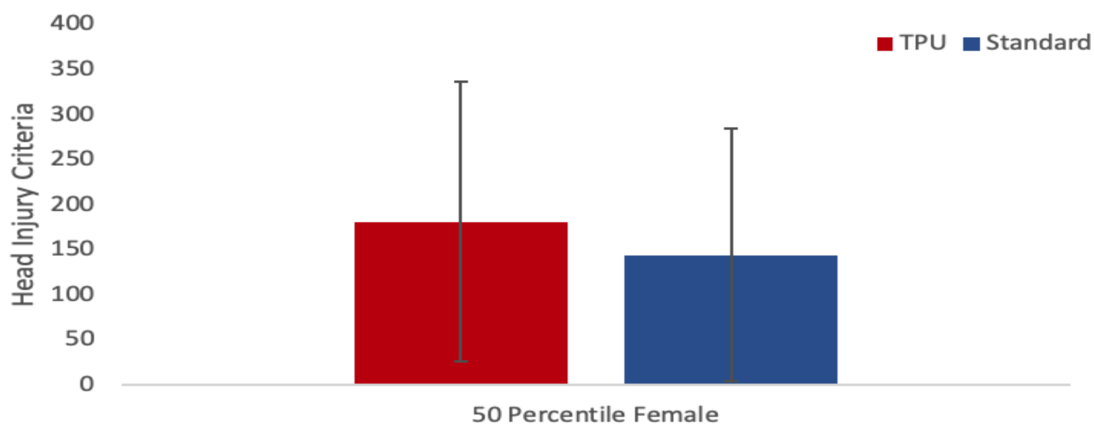


When examining the interaction effect by conducting a simple main effect analysis, the *t*-tests for repeated measures revealed statistically significant differences between the TPU liner

(M = 181.25, SD = 155.25) and the standard liner (M = 144.72, SD = 156.76) for the 50th percentile female with a small effect size, t (17) = 2.03, p = .05, d = 0.48, CI = [-0.018, 0.96] for the HIC measures (see Figure 35). To further explain the interaction, one-way ANOVAs were conducted to examine the effect of neck strength level for each helmet liner type separately, and no statistically significant differences were found for the measures of HIC at the side location.

Figure 35

Differences Between Helmet Liner for 50 Percentile Neck Strength

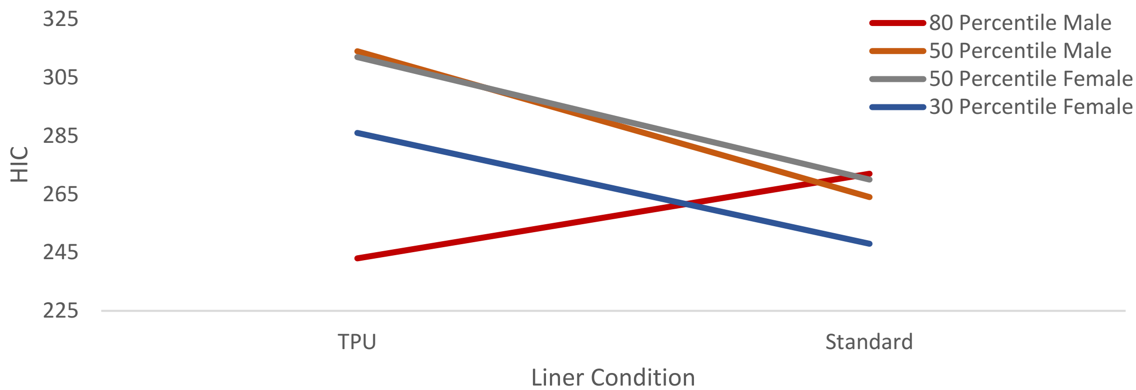


Back Location HIC

The two-way mixed factorial ANOVA results revealed a statistically significant interaction effect between neck strength level and helmet liner type for measures of HIC with a large effect size, F (3,68) = 9.285, p < .001 η^2 = 0.291. The interaction effect between helmet type and neck strength level for measures of HIC at the back location is illustrated in Figure 36.

Figure 36

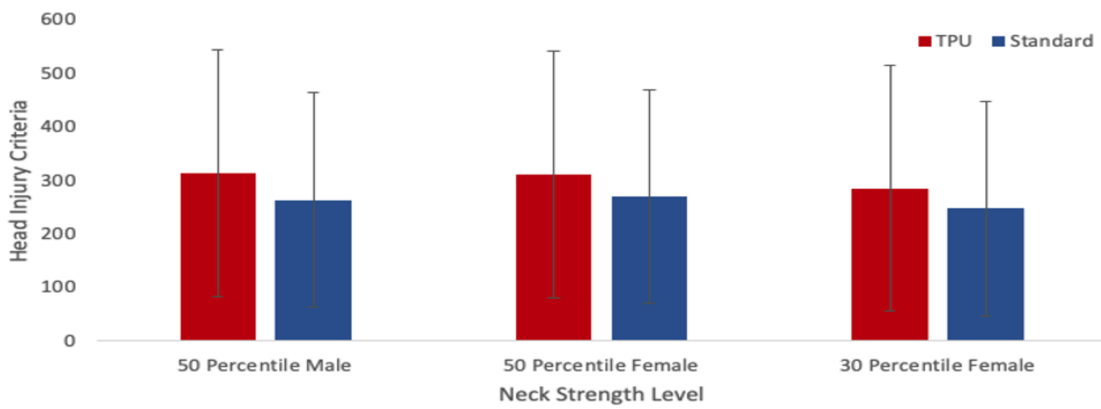
Interaction Effect for Back Location



When examining the interaction effect between helmet type and neck strength level for HIC measures by conducting a simple main effect analysis, the t-tests for repeated measures revealed statistically significant differences between the TPU liner (*M* = 314.99, *SD* = 223.32) and standard liner (*M* = 264.57, *SD* = 196.15) for the neck strength level of the 50th percentile male with a … effect size, *t* (17) = 6.127 *p* < .001, *d* = 1.44, *CI* = [0.77, 2.10]. The t-test for repeated measures analysis also revealed statistically significant differences between the TPU liner (*M* = 312.28, *SD* = 270.19) and standard liner (*M* = 270.19, *SD* = 224.04) for the 50th percentile female with a … effect size, *t* (17) = 6.90, *p* < .001, *d* = 1.63, *CI* = [0.90, 2.33]. Lastly, the t-test for repeated measures analysis revealed statistically significant differences between the TPU liner (*M* = 286.38, *SD* = 233.64) and standard liner (*M* = 248.76, *SD* = 191.80) for the neck strength level of the 30th percentile female with a … effect size, *t* (17) = 2.61, *p* = .018, *d* = 0.62, *CI* = [0.10, 1.13]. Figure 37 outlines the differences in the HIC measures based on neck strength level for each helmet liner type. To further explain the interaction effect, one-way ANOVAs were conducted to examine the effect of neck strength level for each helmet liner type separately, and no statistically significant differences were found for the measures of HIC at the back location.

Figure 37

HIC Based on Neck Strength Level



Post-hoc Exploratory Results

Peak Resultant Linear Acceleration and Head Injury Criteria Comparison

During the dynamic phase of the study, the researcher conducted a post-hoc exploratory analysis to examine a trend that emerged between PRLA and HIC values across the impact velocities for the injury threshold levels of the standard and TPU goaltender helmet liners. This trend is related to the association between the PRLA value of 82 gs (Zhang et al., 2004) and the HIC value of 250, representing the 50% chance of concussion normally used for football helmets (Gwin et al., 2008).

Interestingly, the impact velocity that caused the head to exceed the injury threshold level varied between the PRLA and the HIC measures for the TPU liners across impact velocities. The lowest velocity to apply more than 82 g of PRLA to the head, for example, was 3.25 m/s when using a goaltender helmet with a TPU liner, but was 3.61 m/s when using a goaltender helmet with a standard liner.

On the contrary, the lowest velocity to produce a HIC value of 250 due to a head impact was 3.95 m/s when using a goaltender helmet with a TPU liner and 4.11 m/s when using goaltender helmet with a standard liner. The impact velocities seem higher for HIC threshold values than PRLA threshold values. In other words: (1) the difference in impact velocity values related to 50% chance of concussion between PRLA and HIC measures was 0.69 m/s for the TPU helmet liner and 0.40 m/s for the standard liner; and (2) the impact velocity value related to 82 g of PRLA in the TPU helmet liner was 0.36 m/s lower than that for the standard liner; similarly the impact velocity value related to HIC value of 250 in the TPU helmet liner was 0.16 m/s lower than that for the standard liner. Figures 38 and 39 illustrate the comparison between the PRLA and HIC for the TPU liner and the standard liner, respectively.

Figure 38

PRLA and HIC versus Impact Velocity for TPU Liner

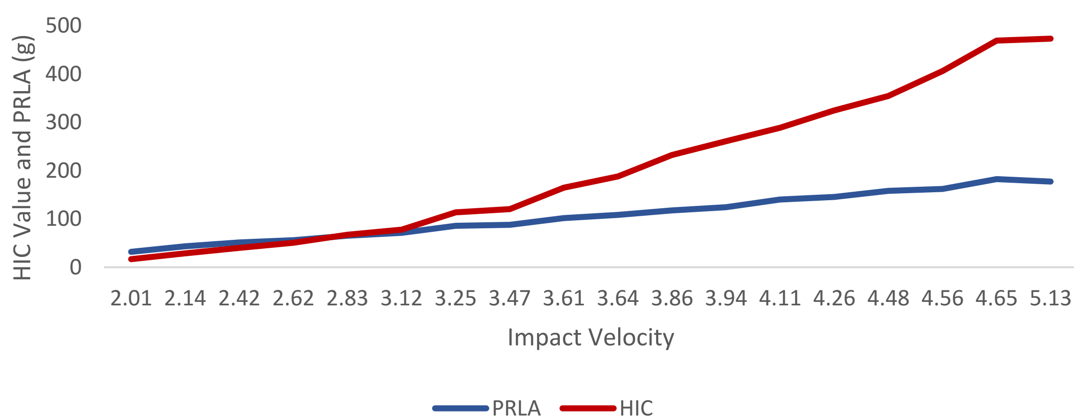
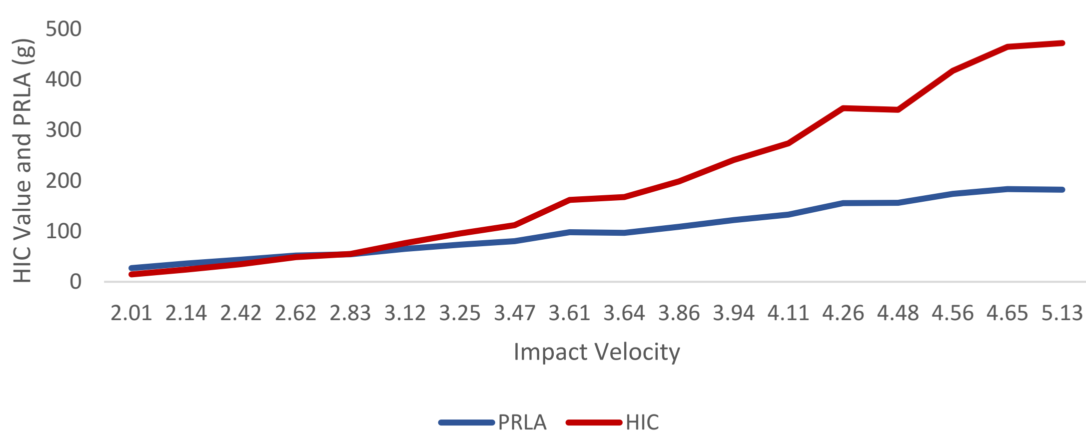


Figure 39

PRLA and HIC versus Impact Velocity for Standard Liner



Chapter 5 - Discussion

Static Testing

The aim of designing an effective helmet liner is to quantify the force needed to cause a specific amount of deformation in the liner material (Krzeminski et al., 2011). In the current study, the researcher statically quantified the amount of deformation of goaltender helmet liners (standard and TPU) to gain information about the capability of the liners to absorb energy. This approach permitted the calculation of energy absorption capacity of the VN goaltender helmet liner and TPU liner per unit of mass to mitigate compressive and shear forces during head collisions. As stated by Clark et al. (2018), a helmet liner must be able to attenuate both compressive and shear forces to decrease the risk of concussions.

More specifically, the researcher examined the energy absorbing properties of the TPU and VN liner materials and compared them in terms of their capacity to absorb energy in J/kg mass when loaded with compressive and shear forces. When comparing the TPU liner to the VN liner across the back, side, upper head, and headband locations, the results showed that the TPU liner absorbed more energy per kilogram in terms of compressive, shear and total energy. Specifically, the TPU liner absorbed 10.2 times more energy per kilogram than the VN liner for compressive energy across all locations. Similarly, the TPU liner absorbed 8.9 times more energy per kilogram than the VN liner for shear energy and 9.5 times more total energy than the VN liner.

The results support the research work conducted by McGillvray (2020), which found that the TPU liners absorbed a higher percentage of energy than the VN liners. The difference between the two studies, however, is that McGillvray (2020) compared the liner types based on the percentage of energy absorption instead of specific energy absorption per kilogram mass of the liner, which is a more standardized measure to compare energy absorption capacity across different types of helmet liner materials in relation to their mass (Bates et al., 2016). In the current study, for example, the TPU liner mass was 2.8 times more per unit volume than the VN liner, absorbing roughly 10 times more energy. In addition, the TPU liner increased the overall mass of the goaltender helmet by 87.78 grams or 6.5% more than when using the VN liner. This increase in mass remained within the range of 1.12 to 1.52 kg for goaltender helmets (Nur et al., 2015) and justified the use of specific energy absorption to conduct an appropriate comparison between the TPU and VN liners. As stated by Tancogne et al. (2016), when using the percentage of energy absorption instead of specific energy absorption, the output value of energy is relative to the loading energy and does not represent the specific energy absorbed relative to each kilogram mass of the liner (Tancogne et al., 2016).

Although the TPU liner absorbed more specific compressive, shear, and total energy per kilogram mass than the VN liner, it turned out to be the less compliant. It required roughly 10 times more force than the VN liner to be compressed 5 mm. This result indicated that the VN liner would saturate and cease to absorb energy under less force than the TPU liner. As Guidice et al. (2020) stated, the saturation point of a material to absorb energy relates to the material's compliance, which is the inverse of stiffness and measures how easily the material compresses under a load.

The reason that the VN liner compressed more under less force than the TPU liner in the current study might be related to the structural composition of the liner materials (Ramirez & Gupta, 2018). The VN liner, for example, consisted of a solid piece of foam. As the VN liner compresses, the density of the material increases until the foam material reaches a saturation point and cannot compress further. In the case of helmet liner technology, it means that once the material saturates, it does not absorb more force and, consequently, more force and energy is transferred to the goaltender's head (Jeong et al., 2014). This result supports the use of TPU as a potential goaltender helmet liner material based on Ouckama and Pearsall (2018) who found that stiffer liners mitigated impact forces more effectively than softer liners at high impact velocities that can cause concussions.

The TPU liner material, on the other hand, consisted of freely moving air-filled cells, which require more force than the VN liner to reach a saturation point. This microscopic characteristic of the TPU material might have been why the TPU liner performed better in the current study in terms of force attenuation and energy absorption across all locations of the goaltender helmet during the static testing. Ramirez and Gupta (2018) stated that when the TPU liner material compresses, the air-filled cells collapse, forcing the air to escape through perforations in the cells to dissipate force and energy more efficiently.

The findings of the static testing indicated that the TPU liner outperformed the VN helmet liner in energy absorption per kilogram for the goaltender helmet's back, side, upper head, and headband locations. The TPU liner absorbed approximately ten times more energy per kilogram than the VN liner. This outcome provided enough evidence for the researcher to proceed with the TPU_8 liner design to conduct dynamic impact testing to further compare the TPU liner design with the VN goaltender liner.

Repeated Impact Testing for the Helmet Liners

According to Mane et al. (2017) and Bhinder et al. (2021), materials can behave differently when loaded statically as compared to dynamically under different strain rates. Based on this notion, the current study used repeated impact testing to assess the changes in the mechanical properties of the TPU and VN liners and compared them to a bare surrogate head without including the shell. The shell was excluded because the repeated impact testing only aimed to compare the helmet liners' capacity to mitigate force and acceleration dynamically. As stated by Spyrou et al. (2000), the local shell geometry could significantly influence the impact absorption of a helmet liner, consequently, requiring the testing of more helmet locations to assess the performance of the liner under different strain rates. Assessing the liners without including the shell helped identify which helmet liner condition (bare head, TPU, or VN) performed the best during repeated head impact testing to mitigate force and acceleration. This approach also helped examine the helmet liners' capabilities to rebound and mitigate impact forces and accelerations in rapid succession. These capabilities of helmet liner materials are important in the sport of ice hockey as impacts occurring in rapid succession during a fall for example, cause the goaltender's head to bounce off the ice and strike the ice again. This type of fall mechanism has the potential to cause a second impact syndrome, which may lead to further swelling of the brain from previous head impacts and, in some instances, death (Wetjen et al., 2010). As stated by Clark (2018), if the liner material cannot rebound, it means that the helmet liner is not absorbing impact forces and accelerations, and consequently, a second impact can be much more dangerous for the athlete.

The results of the current study indicated that the bare head experienced significantly more force than the TPU and VN liners. This outcome highlighted the role of compressible helmet liners under single and successive impacts to mitigate forces that cause concussions or brain injuries (Clark et al., 2018). The bare head impact condition involved non-compliant material when colliding with the force plate under low speeds, and consequently, the entire force transferred directly to the head. On the contrary, the TPU and VN helmet liners had more compliance under dynamic impacts and absorbed roughly 18 % of the impact forces. Ouckama and Pearsall (2014) also found a similar trend when testing different helmet liners. The researchers identified that softer liners reduced impact forces more favorably than rigid liners at low impact speeds.

In the current study, the results revealed no statistically significant differences between the two liners for measures of force, which indicated that both liners rebounded in a similar way under repeated impacts for low speeds. This outcome supported the notion that the TPU material has elastic properties enabling it to rebound rapidly to its original shape to maintain its force absorption capacity (Ramirez & Gupta, 2018) similarly to the VN liner during successive impacts. Since both the TPU and VN liners rebounded rapidly during successive impacts, this outcome further justified their use and implementation in goaltenders' helmet liner designs as an avenue for head injury prevention. A potential reason for the lack of significant differences between the TPU and VN liners for the measures of force can be explained under the notion of material strain and impact velocity. The repeated impact testing occurred at low speeds ranging from 0.21 m/s to 0.86 m/s due to mechanical constraints of the impactor. The impacts generated a maximum acceleration of 4.69 g for the VN liner and 1.65 g for the TPU liner, which was not high enough for the TPU liner to trigger its material absorption capabilities to mitigate the impact force. Ramirez and Gupta (2018) stated that when the TPU liner material experiences higher material strain, the material can dynamically bend and twist the cell walls to mitigate forces and accelerations more efficiently. This suggests that a higher impact speed, or a higher impact force, is needed to experience the dynamic bending and twisting mechanism of the TPU material.

For the measures of maximum acceleration, the TPU liner experienced significantly less acceleration than the bare head. This outcome indicated that the TPU performed better under successive impacts when compared to the bare head condition to mitigate linear impact accelerations for low-speed collisions. This result was consistent with the research work of Ouckama & Pearsall (2014), which found that softer materials performed better in mitigating accelerations during low-speed impacts. This finding, however, was expected as the role of a helmet liner is to decrease the amount of acceleration experienced by the head (Clark, 2018). The VN liner, on the other hand, performed the same as the bare head condition. One potential reason for this result relates to the loading rate of impact acceleration of the VN liner as compare to the TPU liner (Ramirez & Gupta, 2018). The VN liner's material strain rate changes throughout the course of an impact and increases as the structure collapses and compresses, making the material denser (Ramirez & Gupta, 2018). As the VN material approaches its saturation point, it can lead to higher acceleration because the impact acceleration applied to the head increases when the material collapses. The TPU material, however, limits the changes in strain rates by releasing air from the large cells of the material as the impact occurs, which allows the TPU material to maintain a constant density for a longer period to produce gradual deceleration before reaching a saturation point. (Ramirez & Gupta, 2018

In summary, the findings from the repeated dynamic testing supported the concept that helmet liners can mitigate impact force and acceleration (Clark, 2018). This outcome led the researcher to proceed with the use of the current TPU liner design to examine its viability as goalie helmet liner by simulating ice hockey goaltender horizontal head collisions at high impact velocities.

Dynamic Impact Testing Simulating Goaltenders' Head Collisions

Peak Resultant Linear Acceleration

Peak resultant linear acceleration is the measure currently used for the certification process of ice hockey goaltender helmets (Clark, 2015). Hockey goaltender helmets must limit

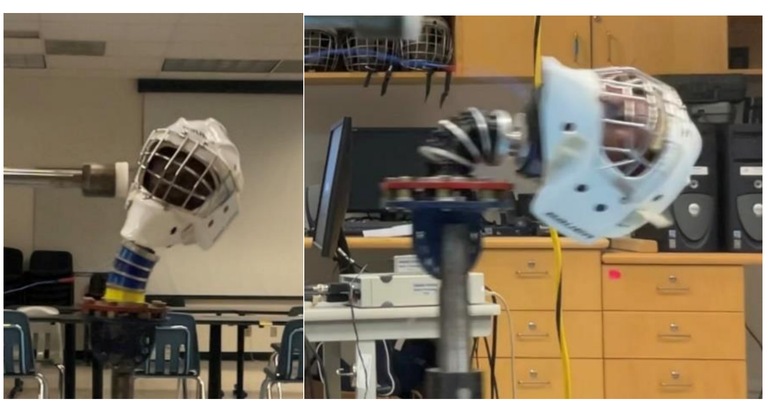
PRLA to a value below 275 g when dropped from a vertical height of 1.5 m for the helmet to be certified (Clark, 2015). The current study chose to implement horizontal impacts to recreate collisions that cause concussions for ice hockey goaltenders, as most of the concussion occur from horizontal head collisions instead of falls for this population group (Clark et al., 2018). The researcher also based his decision on the research work conducted by Jefferies et al. (2017), which found evidence of strong correlations for measures of PRLA between horizontal and vertical head impacts at equivalent velocities.

The aim of the current study was to identify the combined effect of neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male) and type of goaltender helmet liner (TPU and VN) on measures of PRLA during simulated dynamic impact (side, front, and back) locations of the head, respectively. Statistically significant interactions were observed between helmet liner materials and neck strength levels with a large effect size on the PRLA measures for the front and back locations, respectively. The findings suggest that for the front location the TPU liner mitigates force more effectively than the VN liner for the highest neck strength level. Whereas for the back location, the VN liner is more favorable for lower neck strength levels. These findings highlight the need for a multifaceted approach that accounts for the type of helmet liner material and the neck strength level to mitigate PRLA and decrease the risk of concussion for ice hockey goaltenders at the front and back locations of the head (Carlson, 2016; Clark, 2018; Pennock, 2018).

When further examining the interaction effect between neck strength level and helmet type on measures of PRLA for the front and back locations, the results indicated no statistically significant main effects across neck strength levels for each helmet liner type, respectively. This outcome supported the research work of Carlson (2016) and Pennock (2018), which also did not find statistically significant effects across neck strength levels for measures of PRLA. The results of the current study along with the research work of Carlson (2016) and Pennock (2018) did not support the hypothesis that a stiffer neck would mitigate linear accelerations experienced during an impact and, consequently, concussion risks (Mihalik et al., 2011). The reason for this trend may be related to the compliance of the neckform. If the neck is stiff then it limits the ROM of the head, which decreases the amount of angular displacement that occurs during the impact and more loading energy transfers directly to the head (Pennock, 2018). Figure 40, for example, shows more peak lateral flexion for the 80th percentile male neck strength level (a) than the 30th percentile female neck strength level (b) during a 5.15 m/s head impact when using a goaltender helmet. This observation suggests that simply measuring the linear acceleration of a head collision may not fully assess the risk of head injury or concussion when examining helmet liner materials, as rotational accelerations also contribute to the risk of concussion (Giudice et al., 2011). In the current study, however, the researcher did not measure the angular acceleration of the head due to a failure of the sensory technology instrumented in the surrogate headform.

Figure 40

Comparison Between Highest and Lowest Neck Strength Levels



Note. Image (a) represents the 80th percentile male neck strength level, and image (b) represents the 30th percentile female neck strength level.

On the contrary, when further examining the interaction effect between helmet types for each neck strength level on PRLA for the front location, the TPU liner resulted in significantly lower PRLA values than the VN liner for the 80th percentile neck strength level. The VN liner, however, had significantly lower PRLA values than the TPU liner for the 30th percentile female neck strength level. A potential reason for this finding may be related to the material properties of the liners (Ramirez & Gupta, 2018) in relation to the neck strength level. In the current study, the static testing revealed that the TPU liner absorbed on average 10.2 times more energy than the VN liner. The freely moving air-filled cells of the TPU material, however, required more force than the VN liner to compress and reach a saturation point (Ramirez & Gupta, 2018). With reference to Figure 40, the higher the neck strength level, the higher the resistance to the impactor rod, and the higher the impact force on the helmet. It seemed that the impact force associated with the 80th percentile male neck strength level might be high enough to cause the air in the TPU material to be squeezed out through perforations in the cells to dissipate force and energy more efficiently (Ramirez & Gupta, 2018). Eckner et al. (2014) and Nagy et al. (1974) found that when less force was applied to the TPU material, less activation of the freely moving air-filled cells occurred to reduce the impact force. For the lowest neck strength level examined in this study (30th percentile female), due to the high compliance of the neck, lower impact forces would be applied to the helmet. This lower force might have been low enough for the VN liner to not reach its saturation such that it might be effective to absorb impacts and mitigate PRLA.

When examining the interaction effect between helmet types for each neck strength level on PRLA at the back location, the results indicated significant differences between the TPU and VN liners for the 50th percentile male, 50th percentile female, and 30th percentile female neck strength levels on PRLA. The VN liner had lower levels of PRLA than the TPU at all three neck strength levels. These results, however, are different from the research work conducted by Gimbel and Hoshizaki (2008), which found that the TPU liner materials reduced loading energy and PRLA significantly better than the VN liner materials at impact. A possible reason for these differences might be that Gimbel and Hoshizaki (2008) used impact velocities between 5.3 m/s and 9.0 m/s compared to the current study where the velocities ranged from 2.01 m/s to 5.13 m/s. As stated by Ramirez and Gupta (2018), the TPU material required sufficient impact force to active more of the freely moving air-filled cells of the material to absorb energy and consequently minimize the magnitude of the impact. Gimbel and Hoshizaki (2008) also found minor differences between TPU and other liners at velocities similar to the current study, and higher significant differences between TPU and other liners at velocities up to 9.0 m/s.

The results of the current study for the PRLA at the back location are also different from the research work conducted by McGillvray (2020), which compared TPU to VN liners in horizontal collisions and found that the TPU liners outperformed the VN liner at different impact locations of the helmet including the back location. A potential reason for this difference might be related to the geometry of the helmet and the design of liners. McGillvray (2020) tested ice hockey player helmets instead of goaltender helmets with different TPU liner design from that used in the current study. Hockey player helmets are designed entirely differently from goaltender helmets due to the differences in the positional demands of players in ice hockey (Clark, 2018). Goaltender helmets are designed with a rigid shell and soft foam liner (Clark, 2018). Compared to hockey player helmets that consist of a plastic shell with a more rigid liner and are designed to be light weight (Clark, 2018). Furthermore, there are several geometric differences between a goaltender helmet and a hockey player helmet, which may cause the helmets to mitigate PRLA impacts differently. For example, a primary function of goaltender helmets is to deflect projectiles traveling from in front of the goaltender leading to highly specialized and directionally dependent shell geometry. However, due to the positional differences, hockey player helmets are more evenly distributed to mitigate impacts equally from different directions (Clark, 2018).

No statistically significant interaction effect between helmet liner material and neck strength level or main effects were found at the side location on PRLA. This outcome supported the research work of Pennock (2018) and Carlson (2016), which found no statistically significant differences in PRLA based on neck strength level during vertical drop tests. This result is contradictory to clinical outcomes that find increases in neck strength are related to decreased risk of concussion (Mihalik et al., 2011). Therefore, it is necessary to continue to develop mechanical neckforms that more accurately recreate these clinical outcomes to improve our understanding of cervical muscle strength and concussions. The results of the current study, however, differed from the research conducted by Rousseau and Hoshizaki (2009) which found that for horizontal collisions, PRLA increased with greater level of neck strength. A potential reason for these differences might be related to the impact velocities. Rousseau and Hoshizaki (2009) conducted impact velocities ranging from 5 m/s to 9 m/s, with a soft neckform condition, which produced significantly less PRLA than when the impacts were conducted with a stiff neck condition at higher velocities. In the current study, the greatest impact velocity was 5.13 m/s. The trend observed by Rousseau and Hoshizaki (2009) might be present at greater impact velocities for the side location of the helmet when using a TPU liner as compared to a VN liner. The underlying reason for this finding will be a worthy topic for further investigation.

It is important to note from the results of the current study, however, that the TPU and VN liners reduced the impact accelerations to a value below the standard certification threshold level of 275 g for all velocities, locations, and neck strength levels. This outcome supported the research work conducted by Nur et al. (2015) and Clark et al. (2018) suggesting impact acceleration threshold values of 275 g may be too high. Some of the PRLA values in the current study, however, were above the threshold levels of concussion stated by Zhang et al. (2004). These researchers suggested that a PRLA value of 66 g, 82 g, and 106 g at the head's center of gravity corresponds to a 25%, 50% and 80% probability of sustaining a concussion. This result indicates that although the helmets are being certified for use, they are still putting goaltenders at risk for concussion. Meaning that revision of the certification standard may be necessary. Continued research is required to develop goaltender helmets that are able to sufficiently mitigate impact forces.

The results of the current study added to the concern that despite goaltender helmet liners' capacity to limit PRLA to a value below the threshold level recommended by helmet manufacturers, concussions still occur for ice hockey goaltenders, which suggests that current helmet certification standards fail to address the concussion issue and risk for ice hockey goaltenders. In addition to linear acceleration, angular acceleration and risk of head injury due to angular acceleration measures must be addressed when determining the safety of a goaltender helmet. Finally, the outcome of this study supported the research work conducted by Clark (2015), which found that ice hockey goaltenders' concussions occur most frequently from horizontal impacts, leading to high magnitudes of linear and rotational accelerations experienced by the brain. Currently, goaltender helmets are being assessed with drop tests which produce different levels of linear and angular acceleration than horizontal impacts (Clark, 2018). These two findings highlight the need to modify and improve the current goaltender helmet certification process to include horizontal impacts as well as angular acceleration-based risk of injury measures in helmet testing as an avenue to help design better helmets to decrease concussion risk for goaltenders.

Risk of Injury Measures

When determining the effectiveness of an ice hockey goaltender helmet liner, it is critical to consider the capability of the material to mitigate the risk of head injury for the goaltender. The current study examined the HIC scores because this risk of head injury index not only considers the magnitude of the impact acceleration but also the duration of impact that the brain experiences to assess the severity of the head injury more accurately (Schmitt et al., 2010). More specifically, this study aimed to identify the interaction effect between neck strength level (30^{th} percentile female, 50^{th} percentile female, 50^{th} percentile male, and 80^{th} percentile male) and helmet liner type (TPU or VN) on measures of HIC during simulated dynamic head collisions. Statistically significant interactions were found between the helmet liner type and neck strength level on HIC scores at the front, side, and back locations of the goaltender helmet, respectively. These statistically significant interactions indicated that neck strength level affected the capability of the goaltender helmet liners in mitigating the HIC scores during horizontal head collisions differently depending on the location of the impact. As stated by Clark (2018), the use of helmets to protect the head without improving the athletes' neck strength levels might not be enough to mitigate the risk of concussion.

When examining the effect of the interaction across neck strength levels for each helmet liner type separately, the results revealed no statistically significant differences on HIC scores for the any of the impact locations. This outcome supported the research of Carlson (2016) and Pennock (2018), which also found no differences on HIC scores across different neck strength levels. On the other hand, when examining the effect of the interaction with respect to helmet liner type for each neck strength level separately, the results revealed statistically significant differences between helmet liner types on HIC scores. As depicted in Figure 32 and Figure 33 the TPU liner performed better in reducing the HIC scores than the VN liner for the 80^{th} percentile male neck strength level at the front location. This outcome supported the research work conducted by Gimbel and Hoshizaki (2008) and McGillvray (2020), which found that TPU liners outperformed the VN liners in terms of mitigating concussion risk. The VN liner, however, performed better than the TPU liner in reducing HIC scores for the 50^{th} percentile male neck strength level at the side location. Similarly, the VN liner performed better in reducing HIC scores than the TPU liner for the 50^{th} percentile of male and female neck strength levels, and the 30^{th} percentile female neck strength level at the back location. These outcomes do not support the research work conducted by Gimbel and Hoshizaki (2008) and McGillvray (2020). These discrepancies might be related to lower levels of neck strength implemented in the current study as compared to the level of neck strength implemented in the research work conducted by Gimbel and Hoshizaki (2008) and McGillvray (2020). As previously stated, the TPU material requires sufficient impact force to active more of the freely moving air-filled cells of the material to absorb energy and consequently minimize the magnitude of the impact (Ramirez & Gupta, 2018). The neck strength levels used in the current study ranged from the 30^{th} percentile female to the 80^{th} percentile male when comparing the TPU and VN liners. The neck strength level used in the research work conducted by Gimbel and Hoshizaki (2008) and McGillvray (2020) included a torque valued at 1.356 N-m (Post et al., 2019). This torque value corresponded to the 90^{th} percentile neck strength level of a male ice hockey player (Bronnle, 2011), when flexing the neck 5 degrees (Pennock, 2018). Consequently, the high level of neck strength used by Gimbel and Hoshizaki (2008) and McGillvray (2020) might have facilitated the production of more impact acceleration and force for the TPU material to absorb energy and consequently minimize the magnitude of the impact (Ramirez & Gupta, 2018).

Another potential reason for the VN liner performing better that the TPU liner at the back location in the current study might be related to the tightness of the elastic straps holding the helmet to the head. The VN liner is more compliant than the TPU liner and it requires less force to compress the material to fit it snuggly to the face of the goaltenders. Hence, the elastic straps holding the back plate of the VN helmet did not have to be as tight to hold it in the correct position as compared to the TPU helmet. As stated by Clark (2018), goaltender helmets are designed with elastic straps and soft liners to be easily adjustable for a proper fit to the face of the goaltender. A softer helmet liner will conform to the face more easily and will require less force to compress the material leading to comfort for the athlete (Jeong et al. 2014). Hao et al. (2019) also determined that the softness and tightness of the fitting mechanism of a helmet are critical components, which contribute to the overall comfort of the helmet. It is plausible to assume then that goaltender helmet manufacturers choose more compliant helmet liners to increase the viability and comfortability of the helmet as important factors for commercialization of the product without sufficiently assessing the effect of the elastic straps on the risk of concussion for goaltenders (Colello et al., 2018). Based on the current study, it is important to continue to develop helmet liners, so protection is not sacrificed for comfort. There may be potential for a liner to be constructed of a rigid base layer situated against the shell with a soft exterior layer to facilitate comfort.

During the horizontal impacts to the back of the head in the current study as shown in Figure 41, the impactor rod struck the back plate of the helmet, causing the helmet to translate forward. In the situation when the straps were loose for the VN liner as depicted as (a) and (b) of

Figure 41, the helmet moved further away from the face of the surrogate head. Although this loose fit led to lower levels of PRLA and HIC experienced by the head and helmet according to the data in the current study, in the real world, this outcome might not translate to increased protection of the head against concussions due to a possible escalation of rotational acceleration. Jadischke (2012), for example, found that loose-fitting American football helmets produced lower linear accelerations of the athletes' head but higher rotational accelerations than tight-fitting helmets. Moghaddam and Kwok (2019) also found that loose-fitting helmets resulted in higher variation in linear accelerations during simulated head collisions. Therefore, despite the VN liner outperforming the TPU liner in terms of measures of PRLA and HIC in the current study, it should be stated that the VN liner might not be the safer option when closely examining the elastic properties of the helmet straps. Despite being the best available liner on the market at the time of the current study, the VN liner used in this study failed to adequately mitigate impact forces and further development is required to improve goaltender safety.

Figure 41

Back Impact Location Difference in Tightness

Note. Image (a) is the tight-fitting TPU-lined helmet. Image (b) on the right is the loose-fitting VN standard-lined helmet.

Comparing the outcomes for measures of PRLA and HIC

This study identified two interesting results when comparing the outcomes for the measures of PRLA and HIC. The first result was the similarity in the outcomes when determining significant differences in liner type for measures of PRLA and HIC. Of the 24 individual impact conditions, 22 were consistent with the outcomes for the measures of PRLA and HIC. These results differ from the research conducted by McGillvray (2020) and Rybak (2021), which found that the outcomes for the measures of PRLA and HIC were similar for all impact conditions tested in their studies. The differences in results may be attributed to the type of helmets used in the studies. McGillvray (2020) used ice hockey player helmets and Rybak (2021) used boxing headgear in their respective studies. Both helmet types vary dramatically from ice hockey goaltender helmets. This discrepancy between the outcomes of the current study and the outcomes of research work conducted by McGillvray (2020) and Rybak (2021) highlighted the importance of including measures of HIC in combination with PRLA to better assess the capacity of helmet liners in mitigating the risk of concussion. As stated by Schmitt et al., (2010), it is important to include the HIC measures to account for the duration of the loading effect that the brain experiences with the impact to assess the severity of the head injury more accurately.

The second result was the difference between the outcomes of the current study for the threshold measures of HIC and PRLA when comparing the TPU and VN liners. As stated by Zhang et al. (2004), 50% risk of sustaining a concussion corresponds to an impact acceleration of 82 g. Hutchinson et al (1998), on the other hand, stated that HIC score of 250 corresponded to a 50% risk of mild brain injury. Interestingly, in the current study, the PRLA and HIC thresholds occurred at different impact velocities for both the TPU and VN helmet liners, respectively. The impact velocity required to reach the threshold value of 82 g for the measures of PRLA was lower than the impact velocity required to reach a threshold value of 250 for the measures of HIC when examining the TPU and VN liners, respectively. This trend highlighted the need for goaltender helmet manufacturing companies to establish more stringent threshold measures of PRLA and HIC when assessing the capability of the helmet liners to reduce the risk of concussion as compared to skull fractures (Clark et al., 2018). In current helmet designs, manufacturing companies used threshold measures of PRLA and HIC more associated with TBI injuries (e.g., skull fractures) than concussions (Kleiven, 2013). The idea of using a more stringent threshold measures of PRLA and HIC for concussion was supported by the research work of Duma et al. (2007) which found that the risk of injury curve produced by Zhang et al. (2004) predicted a concussion rate 100 times more than what is seen in the National Football League. Exploring injury thresholds and the relationship between PRLA and HIC measures based on impact velocity may aid at improving the safety of ice hockey goaltender helmets.

It is important to note that based on the outcomes of this study for measures of PRLA, the TPU liner passed the certification standard for a goaltender helmet because all the PRLA values were under 275 g (Clark, 2015). These findings, however, suggest that further investigation and improvement of the TPU design are necessary to decrease the risk of concussion for ice hockey goaltenders caused by linear acceleration impacts across different neck strength levels.

Chapter 6 - Conclusion

The overall purpose of this study was to examine the combined effect of cervical muscle strength and helmet liner type in mitigating the magnitude of impact accelerations and risk of concussion for ice hockey goaltenders during simulated horizontal head collisions. This study was conducted because there are gaps in the literature in terms of determining how cervical muscle strength and helmet liner material influence the risk of concussion injury due to the amount of linear acceleration applied to the head of ice hockey goaltenders during impact. This study aimed to address this gap by identifying other liner materials such as TPU and comparing it to a VN commercial liner for different neck strength levels to examine the capabilities of the helmet liners to reduce the risk of concussion for goaltenders during simulated head collisions. To understand the material properties of the helmet liners, the liner materials were first tested on their capacity to absorb energy per kilogram during static loading. The liners were then tested during repeated impacts to understand their capacity to absorb impact force and rapidly recover to the original shape. Finally, horizontal impacts were conducted to measure the liners' capacity to mitigate impact acceleration and risk of injury during the simulation of concussion causing events for ice hockey goaltenders.

The results showed that the TPU liner was able to load 10.2 times more energy per kilogram than the standard helmet liner during static testing. This result, however, did not translate to all the conditions tested during the dynamically simulated goaltender head collisions. The TPU liner performed better than the standard liner for one of the 12 combinations of neck strength level and location during the dynamic head impact simulations. The VN liner, on the other hand, outperformed the TPU in four combinations of neck strength level and location. The seven remaining combinations revealed no statistically significant differences between the two liners. Additionally, differences in neck strength did not lead to any significant differences in PRLA or HIC.

The data presented in the current study, however, help provide a better understanding of the risk of injury and highlight possible strategies to develop more effective TPU helmet liners as a prevention strategy for ice hockey goaltenders during horizontal head collisions. The data from the current study also provide insight into the need to create more standardized measures of neck strength levels for male and female populations when examining measures of linear acceleration and risk of head injury during horizontal head impact simulations for goaltenders.

Strengths

A strength of the current study was that it included measures of neck strength levels based on male and female populations for ice hockey players. The researcher of the current study did not find prior studies that used measures of neck strength levels to simulate goaltenders' head impacts. In previous studies, the neckform tension level used in simulations for goaltender head impact only represented the 90th percentile for a male ice hockey player based on the use of a standard neckform tensioned to 1.356 N-m (Pennock, 2018; Bronnle, 2011). Consequently, the researcher believes that this is the first study to use different levels of neck tension to simulate goaltenders' neck strength levels during goaltender helmet impact testing. Another strength of the current study was the implementation of three testing phases, which included the static testing of the liners, dynamic testing of liners, and testing of the goaltender helmet and liners based on dynamic simulations of goaltenders' head collision which may lead to concussions while playing the sport of ice hockey.

Limitations

In the current study, it was not feasible to the test the liners at high impact velocities for repetitive head collisions during the second stage of data collection due to technology constraints as the surrogate head caused excessive vibrations at impact. The researcher, however, eliminated noise induced in the data for the range of lower impact velocities conducted in the study by implementing digital filters using a MATLAB® script.

The study did not include measures of angular acceleration, which is considered an important measure to assess concussion risk during horizontal head collisions. Due to a failure in the hardware, the researcher was not able to include this measure in the third stage of the current study.

During the third stage of the study, only one impact was conducted for each combination of neck strength level and helmet liner type across the 18 velocities tested. Ideally, the average of three impacts trials would be taken per combination of neck strength level and helmet type at each velocity. Due to financial constraints caused by the cost of the goaltender helmet and the risk of the deterioration of the helmet shells, however, only one impact per condition was conducted. The researcher attempted to limit this potential risk of error by analyzing the acceleration values of every impact and retesting trials that were abnormally high or low.

The neck strength values used for the simulations of the current study were secondary data from two different studies that included male and female ice hockey players respectively but no ice hockey goaltenders specifically. It is important to note that goaltenders were not excluded from participation in either study, meaning that goaltenders might have been included in the data. It is plausible to assume, however, that ice hockey goaltenders may have different neck strength levels than ice hockey players due to the different positions they play in the game.

Future Directions

Future studies should continue to explore different TPU shapes, densities and structures. As outlined in the current study, slight changes in TPU structure, design and density can lead to different static loading capabilities. The second area of focus for future research is to include measures of angular acceleration and energy absorption for the helmet liners across different levels of neck strength for goaltenders and impact locations.

The final area of focus for future researchers would be to simulate other goaltender concussion causing events such as projectiles. For a goaltender helmet to effectively reduce the risk of concussions, it must be tested in the manner that is specific to how goaltenders are injured. Because the impact parameters are significantly different for projectile impacts compared to horizontal collisions, it is necessary to test goaltender helmets under different mechanism of injury.

References

Anderson, E. D., Giudice, J. S., Wu, T., Panzer, M. B., & Meaney, D. F. (2020). Predicting concussion outcome by integrating finite element modeling and network analysis. *Frontiers in Bioengineering and Biotechnology*, *8*, 1–13. https://doi.org/10.3389/fbioe.2020.00309

Andriessen, T. M. J. C., Jacobs, B., & Vos, P. E. (2010). Clinical characteristics and pathophysiological mechanisms of focal and diffuse traumatic brain injury. *Journal of Cellular and Molecular Medicine*, *14*(10), 2381–2392. https://doi.org/10.1111/j.1582-4934.2010.01164.x

Bates, S., Farrow, I., Trask, R. (2016). 3D printed polyurethane honeycombs for repeated tailored energy absorption. *Materials & Design 112*(15), 172-183. https://doi.org/10.1016/j.matdes.2016.08.062

Beckwith, J. G., Greenwald, R. M., & Chu, J. J. (2012). Measuring head kinematics in football: Correlation between the head impact telemetry system and hybrid III headform. *Annals of Biomedical Engineering*, *40*(1), 237–248. https://doi.org/10.1007/s10439-011-0422-2

Bell, G. J., Snydmiller, G. D., & Game, A. B. (2008). An investigation of the type and frequency of movement patterns of National Hockey League goaltenders. *International Journal of Sports Physiology and Performance*, *3*(1), 251-265. https://doi.org/10.1123/ijspp.3.1.80

Bežák, J., & Přidal, V. (2017). Upper body strength and power are associated with shot speed in men's ice hockey. *Acta Gymnica*, *47*(2), 78–83. https://doi.org/10.5507/ag.2017.007

Bhinder, J., Kumar Verma, S., Agnihorti, K. (2021). Qualifying carbon nanotube reinforced polyurethane foam as helmet inner liner through in-situ, static and low velocity impact testing. *Materials Science and Engineering D: Solid-State Materials for Advanced Technology.* 78-90. 10.1016/j.mseb.2021.115496. 300-312.

Biasca, N., Wirth, S., & Tegner, Y. (2002). The avoidability of head and neck injuries in ice hockey: An historical review. *British Journal of Sports Medicine*, *36*(6), 410–427. https://doi.org/10.1136/bjsm.36.6.410

Bogduk, N., & Mercer, S. (2000). Biomechanics of the cervical spine. I: Normal kinematics. *Clinical Biomechanics*, *15*(9), 633–648. https://doi.org/10.1016/S0268-0033(00)00034-6

Broennle, M. (2011). *The Relationship Between Maximal Isometric and Isotonic Neck Strength of Hockey Players and Wrestlers.*

Bussey, M. D., McLean, M., Pinfold, J., Anderson, N., Kiely, R., Romanchuk, J., & Salmon, D. (2019). History of concussion is associated with higher head acceleration and reduced cervical muscle activity during simulated rugby tackle: An exploratory study. *Physical Therapy in Sport*, *37*, 105–112. https://doi.org/10.1016/j.ptsp.2019.03.012

Cagnie, B., Cools, A., De Lose, V., Cambier, D., & Danneels, L. (2007). Healthy controls and women with chronic neck pain: The use of a reliable measure. *Archives of Physical Medicine and Rehabilitation, 88*, 1441-1445. doi: 10.1016/j.apmr.2007.06.776

Carlson, S., Zerpa, C., Hoshizaki, T., Patemon, G., Przysucha, E., & Sanzo, P. (2016). Evidence of reliability and validity for the use of a helmet impact drop system. *34rd International Conference on Biomechanics in Sports, Tsukuba, Japan*, 18–22.

Clark, J. M. (2015). *Evaluation of the Protective Capacity of Ice Hockey Goaltender Masks for Three Accident Events using Dynamic Response and Brain Stress and Strain In-depth Analysis and Reconstruction of Equestrian Jockey Accidents and Associated Headgear View project*. https://doi.org/10.13140/RG.2.2.12401.07524

Clark, J. M., Hoshizaki, T. B., & Gilchrist, M. D. (2020). Event-specific impact test protocol for ice hockey goaltender masks. *Sports Biomechanics*, *19*(4), 510–531. https://doi.org/10.1080/14763141.2018.1510975

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. https://doi.org/10.1007/s10439-018-2017-7

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. *Journal of Primary Prevention*, *35*(5), 309–319. https://doi.org/10.1007/s10935-014-0355-2

Colello, R., Colello, I., Abdelhamedia, D., Cresswell, Kellen., Merchant, R., Beckett, E. (2018). Making football safer: Assessing current national football league policy on the type of helmets allowed on the playing field. *Journal of Neurotrauma, 35*(11), 1-12. https://doi.org/10.1089/neu.2017.5446

Coulson, N. (2011). *The influence of impact mass, inbound velocity and system compliance on the dynamic response of a hybrid III head form.* http://hdl.handle.net/10393/28886

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. https://doi.org/10.1123/jab.2020-0070

Decloe, M. D., Meeuwisse, W. H., Hagel, B. E., & Emery, C. A. (2014). Injury rates, types, mechanisms and risk factors in female youth ice hockey. *British Journal of Sports Medicine*, *48*(1), 51–56. https://doi.org/10.1136/bjsports-2012-091653

Duma, S., Funk, J. (2007). Biomechanical risk estimates for mild traumatic brain injury. *Association for the advancement of automotive medicine.*

Elkin, B. S., Elliott, J. M., & Siegmund, G. P. (2016). Whiplash injury or concussion? A possible biomechanical explanation for concussion symptoms in some individuals following a rear-end collision. *Journal of Orthopaedic and Sports Physical Therapy*, *46*(10), 875–885. https://doi.org/10.2519/jospt.2016.7049

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. *The American Journal of Sports Medicine, 42*, 566-576. doi: 10.1177/0363546513517869

Gennarelli, T. A., Thibault, L. E., Adams, J. H., Graham, D. I., Thompson, C. J., & Marcincin, R. P. (1982). Diffuse axonal injury and traumatic coma in the primate. *Annals of Neurology*, *12*(6), 564–574. https://doi.org/10.1002/ana.410120611

Gennarelli, T., Thibault, L., & Ommaya, K. (1972). Pathophysiological responses to rotational and translational accelerations of the head. *Proceedings of the 16th Stapp Car Crash Conference*.

Gimbel, G., & Hoshizaki, T. (2008). Compressive properties of helmet materials subjected to dynamic impact loading of various energies. *European Journal of Sport Science*, *8*(6), 341–349. https://doi.org/10.1080/17461390802438763

Giudice, J., Caudillo, A., Mukherejee, S., Kong, K., Park, G., Kent, R., Panzer, M. (2020). Finite element model of a deformable American football helmet under impact. *Annals of Biomedical Engineering, 48*(5). 45-56. 10.1007/s10439-020-02472-6

Gutierrez, G. M., Conte, C., & Lightbourne, K. (2014). The relationship between impact force, neck strength, and neurocognitive performance in soccer heading in adolescent females. *Pediatric Exercise Science*, *26*(1), 33–40. https://doi.org/10.1123/pes.2013-0102

Gwin, J,. Chu, J., McAllister, T., Greenwald, R. (2008). In situ measures of head impact acceleration in NCAA division 1 men's ice hockey helmets: implications for ASTM F1045 and other ice hockey helmet standards. *Journal of ASTM International, 6*(6), 1-10.

Hao, W., Shan, L., Yang, W., Jingjing, W., Cheyan, Li., Yun, S. (2019). Analysis of the influence factors of safety helmet comfort. *Earth and Environmental Science.* 1-12.

Harmon, K., Clugston, J., Dec, K., Herring, S., Kane, S., Kontos, A., Leddy, J., McCrea, M., Poddar, S., Putukian, M., Wilson, J., Roberts, W. (2019). American Medical Society for Sports Medicine Position Statement on Concussion in Sport. *British Journal of Sports Medicine, 53*(4), 213-225.

Hodgson, V. R., Thomas, L. M., & Prasad, P. (1970). Testing the validity and limitations of the severity index. *SAE Technical Papers*, 2672–2684. https://doi.org/10.4271/700901

Hoshizaki, T. B., Post, A., Oeur, R. A., & Brien, S. E. (2014). Current and future concepts in helmet and sports injury prevention. *Neurosurgery*, *75*, 136–148. https://doi.org/10.1227/NEU.0000000000000496

Hoshizaki, T. B., Walsh, E., Post, A., Rousseau, P., Kendall, M., Karton, C., Oeur, A., Foreman, S., & Gilchrist, M. D. (2012). The application of brain tissue deformation values in assessing the safety performance of ice hockey helmets. *Journal of Sports Engineering and Technology*, *226*(3–4), 226–236. https://doi.org/10.1177/1754337112448765

Hutchinson, J., Kaiser, M. J., & Lankarani, H. M. (1998). The Head Injury Criterion (HIC) functional. *Applied Mathematics and Computation*, *96*(1), 1–16. https://doi.org/10.1016/S0096-3003(97)10106-0

Jadischke, R. (2012). *Football helmet fitment and its effect on helmet performance.*

Jeffries, L., Zerpa, C., Przysucha, E., Sanzo, P., & Carlson, S. (2017). The use of a pneumatic horizontal impact system for helmet testing. *Journal of Safety Engineering*, *2017*(1), 8–13.

Jeong, K. Y., Cheon, S. S., & Munshi, M. B. (2012). A constitutive model for polyurethane foam with strain rate sensitivity. *Journal of Mechanical Science and Technology*, *26*(7), 2033–2038. https://doi.org/10.1007/s12206-012-0509-1

Kendall, M., Post, A., Rousseau, P., Oeur, A., Gilchrist, M. D., & Hoshizaki, B. (2012). A comparison of dynamic impact response and brain deformation metrics within the cerebrum of head impact reconstructions representing three mechanisms of head injury in ice hockey. *2012 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury*, 430–440.

Kendall, M., Walsh, E. S., & Hoshizaki, T. B. (2012). Comparison between Hybrid III and Hodgson-WSU headforms by linear and angular dynamic impact response. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology*, *226*(3–4), 260–265. https://doi.org/10.1177/1754337112436901

Kimpara, H., & Iwamoto, M. (2012). Mild traumatic brain injury predictors based on angular accelerations during impacts. *Annals of Biomedical Engineering*, *40*(1), 114–126. https://doi.org/10.1007/s10439-011-0414-2

King, J. A., McCrea, M. A., & Nelson, L. D. (2020). Frequency of primary neck pain in mild traumatic brain injury/concussion patients. *Archives of Physical Medicine and Rehabilitation*, *101*(1), 89–94. https://doi.org/10.1016/j.apmr.2019.08.471

Kleiven, S. (2003). Influence of impact direction on the human head in prediction of subdural hematoma. *Journal of Neurotrauma*, *20*(4), 365–379. https://doi.org/10.1089/089771503765172327

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. https://doi.org/10.3389/fbioe.2013.00015

Krzeminski, D., Goetz, J., Janisse, A., Lippa, N., Gould, T., Rawlins, J. (2011). Investigation of linear impact energy management and product claims of novel American football helmet liner component. *Sports Technology 4*(1). 65-76. https://doi.org/10.1080/19346182.2012.691508

Laprade, R. F., Wijdicks, C. A., & Spiridonov, S. I. (2009). A prospective study of injuries in NCAA intercollegiate ice-hockey goaltenders. *Journal of ASTM International*, *6*(3), 310–384. https://doi.org/10.1520/JAI101862

Liberman, M., & Mulder, D. S. (2007). Airway injuries in the professional ice hockey player. *Clinical Journal of Sport Medicine*, *17*(1), 61–67. https://doi.org/10.1097/01.jsm.0000168072.44003.98

MacAlister, A. (2013). Surrogate head forms for the evaluation of head injury risk. *Brain Injuries and Biomecanics*, *37*(12), 1–13. http://hdl.handle.net/10919/23818

Mane, J., Chandra, S., Sharma, S., Ali, H., Chavan, V., Manjunath, B., Patel, R. (2017). Mechanical property evaluation of polyurethane foam under quasi-static and dynamic strain rates. *Procedia Engineering,* 173, 726-731. 10.1016/j.proeng.2016.12.160

Marcotte-L'heureux, V., Charron, J., Panenic, R., & Comtois, A. S. (2021). Ice hockey goaltender physiology profile and physical testing: A systematic review and meta-analysis. *International Journal of Exercise Science*, *14*(6), 855–875.

McGillivray, K. (2020). *Comparison of Innovated Thermoplastic Polyurethane and Commercial Helmet Lining Technology in Mitigating Concussions as Measured by Linear Accelerations and Risk of Head Injury During Simulated Horizontal Impacts to the Head*.

McLean, A., & Anderson, R. W. (1997). Biomechanics of closed head injury. *Head Injury*, 25–37.

Meaney, D. F., & Smith, D. H. (2011). Biomechanics of concussion. *Clinics in Sports Medicine*, *30*(1), 19–31. https://doi.org/10.1016/j.csm.2010.08.009

Mesfin, F. B., Gupta, N., Hays Shapshak, A., & Taylor, R. S. (2017). *Diffuse axonal injury*. StatPearls Publishing, Treasure Island (FL). http://europepmc.org/abstract/MED/28846342

Mihalik, J. P., Guskiewicz, K. M., Marshall, S. W., Greenwald, R. M., Blackburn, J. T., & Cantu, R. C. (2011). Does cervical muscle strength in youth ice hockey players affect head impact biomechanics? *Clinical Journal of Sport Medicine*, *21*(5), 416–421. https://doi.org/10.1097/JSM.0B013E31822C8A5C

Moghaddam, H., Wok, W. (2019). Role of helmet fit on angular and linear accelerations of head in ice hockey. *International Journal of Science Engineering. 7*(2), 26-32. doi: 10.11648/j.ijbse.20190702.11

Morin, M., Langevin, P., & Fait, P. (2016). Cervical spine involvement in mild traumatic Brain injury: A review. *Journal of Sports Medicine*, *2016*, 1–21. https://doi.org/10.1155/2016/1590161

Nagy, A., ko, W. L., & Lindholm, U. S. (1974). Mechanical behavior of foamed materials under dynamic compression. *Journal of Cellular Plastics*, *10*(3), 127–134. https://doi.org/10.1177/0021955X7401000306

Namjoshi, D. R., Good, C., Cheng, W. H., Panenka, W., Richards, D., Cripton, P. A., & Wellington, C. L. (2013). Towards clinical management of traumatic brain injury: A review of models and mechanisms from a biomechanical perspective. *DMM Disease Models and Mechanisms*, *6*(6), 1325–1338. https://doi.org/10.1242/dmm.011320

Netter, F. H. (2019). Atlas of human anatomy (7th ed.). In *Elsevier Inc*.

NOCSAE. (2018). Standard Pneumatic Ram Test Method and Equipment Used in Evaluating the Performance Characterizes of Protective Headgear and Face Guards. Retrieved from https://nocsae.org/wp-content/uploads/2018/05/ND081-18am19-005.pdf

Nordin, M., & Frankel, V. H. (2012). Basic biomechanics of the musculoskeletal system. In *Basic Biomechanics of the Musculoskeletal System*. https://doi.org/10.1136/bjsm.26.1.69-a

Nur, S., Kendall, M., Clark, J., Hoshizaki, B. (2015). A comparison of the capacity of ice hockey goaltender masks for the protection from puck impacts. *Sports Biomechanics,* 14, 459-469. https://doi.org/10.1080/14763141.2015.1084035

Oeur, A., Blaine Hoshizaki, T., & Gilchrist, M. D. (2014). The influence of impact angle on the dynamic response of a Hybrid III headform and brain tissue deformation. *The Mechanism of Concussion in Sports*, 23–29. https://doi.org/10.1520/STP155220120160

Ouckama, R., & Pearsall, D. J. (2014). Projectile impact testing of ice hockey helmets: Headform kinematics and dynamic measurement of localized pressure distribution. *2014 IRCOBI Conference Proceedings - International Research Council on the Biomechanics of Injury*.

Panchuk, D., Vickers, J. N., & Hopkins, W. G. (2017). Quiet eye predicts goaltender success in deflected ice hockey shots†. *European Journal of Sport Science*, *17*(1), 93–99. https://doi.org/10.1080/17461391.2016.1156160

Pellman, E. J., Viano, D. C., Tucker, A. M., Casson, I. R., Waeckerle, J. F., Maroon, J. C., Lovell, M. R., Collins, M. W., Kelly, D. F., Valadka, A. B., Cantu, R. C., Bailes, J. E., & Levy, M. L. (2003). Concussion in professional football: Reconstruction of game impacts and injuries. *Neurosurgery*, *53*(4), 799–812. https://doi.org/10.1093/neurosurgery/53.3.799

Pennock, B. (2018) *The Effect of Isometric Cervical Strength, Head Impact Location, and Impact Mechanism on Simulated Head Impact Measures in Female Ice Hockey Players.*

Pennock, B., Kivi, D., & Zerpa, C. (2021). Effect of neck strength on simulated head impacts during falls in female ice hockey players. *International Journal of Exercise Science*, *14*(1), 446–461. http://www.intjexersci.com

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. https://doi.org/10.1080/10255842.2019.1577827

Post, A., Oeur, A., Hoshizaki, 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. https://doi.org/10.1080/10255842.2011.627559

Pujalte, G. G. A., Dekker, T. M., Abadin, A. A., & Jethwa, T. E. (2020). Signs and symptoms of concussion. In *Concussion Management for Primary Care* (pp. 19–30). Springer International Publishing. https://doi.org/10.1007/978-3-030-39582-7_3

Qi, H. J., & Boyce, M. C. (2005). Stress-strain behavior of thermoplastic polyurethanes. *Mechanics of Materials*, *37*(8), 817–839. https://doi.org/10.1016/j.mechmat.2004.08.001

Ramirez, B. J., & Gupta, V. (2018). Evaluation of novel temperature-stable viscoelastic polyurea foams as helmet liner materials. *Materials and Design*, *137*. https://doi.org/10.1016/j.matdes.2017.10.037

Richards, J. (2008). *Biomechanics in clinic and research.* Philadelphia, PA: Elsevier Limited.

Rivara, F. P., & Graham, R. (2014). Sports-related concussions in youth. *JAMA*, *311*(3), 239–249. https://doi.org/10.1001/jama.2013.282985

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. N. (2014). Research methods in biomechanics. In *Research Methods in Biomechanics*. https://doi.org/10.5040/9781492595809

Rohen, J., Lutjen-Drecoll, E., & Yokochi, C. (2015). *Color atlas of anatomy a photographic study of the human body* (7th ed.). Lippincott Williams & Wilkins.

Rousseau, P., Post, A., & Hoshizaki, T. B. (2009). A comparison of peak linear and angular headform accelerations using ice hockey helmets. *Journal of ASTM International*, *6*(1), 1–13. https://doi.org/10.1520/JAI101877

Rybak, T. (2020). *Towards examining the effectiveness of boxing headguards with thermoplastic polyurethane in mitigating acceleration and risk of head injury using a dynamic head model*.

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. *American Journal of Sports Medicine*, *42*(9), 2056–2066. https://doi.org/10.1177/0363546514536685

Spyrou, E., Pearsall, D. J., & Hoshizaki, T. B. (2000). Effect of local shell geometry and material properties on impact attenuation of ice hockey helmets. *Sports Engineering*, *3*(1), 25–35. https://doi.org/10.1046/j.1460-2687.2000.00045.x

Strich, S. J. (1961). Shearing of nerve fibres as a cause of brain damage due to head injury. A Pathological Study of Twenty Cases. *The Lancet*, *278*(7200), 443–448. https://doi.org/10.1016/S0140-6736(61)92426-6

Tancogne, T., Spieringe, A., Mohr, D. (2016). Additively manufactured metallic micro-lattice materials for high specific absorption under static and dynamic loading. *Acta Materilia.* 119, 14-28. https://doi.org/10.1016/j.actamat.2016.05.053

Tortora, G. J., & Nielsen, M. (2017). Principles of human anatomy. In *Angewandte Chemie International Edition, 6(11), 951–952.*

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. https://doi.org/10.1007/s12283-011-0060-9

Wetjen, N. M., Pichelmann, M. A., & Atkinson, J. L. D. (2010). Second impact syndrome: Concussion and second injury brain complications. *Journal of the American College of Surgeons*, *211*(4), 553–557. https://doi.org/10.1016/j.jamcollsurg.2010.05.020

Zerpa, C., Carlson, S., Sanzo, P., Przysucha, E., Hoshizaki, T., & Kivi, D. (2017). The effect of angle of impact, neck stiffness, and impact location on measures of shear forces during Helmet Testing. *ISBS Proceedings Archive*, *35*(1), 1020–1023.

Zerpa, C., Liu, M., & Shashankdhwaj, P. (2020). An improved cycling helmet technology to mitigate head injuries. *38th International Society of Biomechanics in Sport Conference*, 80–83.

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. https://doi.org/10.1115/1.1691446

Appendix A

Summary Tables for Research Question 1

Table A.1
Summary Information for Samples Used

Location	TPU Mass (grams)	Standard Mass (grams)	Sample Length (mm)	Sample Width (mm)	Sample Depth (mm)	Number of TPU Used
Back	3.13	1.28	27.5	30	10	18
Upper Head	3.16	1.57	25	30	12	12
Headband	4.04	1.87	30	30	12	8
Side	2.39	0.97	25	25	10	12

Note. The TPU liner and standard liner are identical in length, width and depth.

Table A.2
Summary Results for Question 1

Location	Energy	TPU (SD)	Standard (SD)
Headband	Compressive	65.80 (5.32)	5.56 (0.22)
	Shear	41.53 (3.43)	4.15 (0.15)
	Total	91.82 (7.45)	8.61 (0.31)
Upper Head	Compressive	41.14 (4.19)	7.07 (0.29)
	Shear	32.14 (3.36)	5.29 (0.21)
	Total	64.53 (6.67)	10.81 (0.43)
Back	Compressive	70.92 (5.97)	5.95 (0.23)
	Shear	43.86 (4.07)	4.12 (0.14)
	Total	97.92 (8.67)	8.72 (0.30)
Side	Compressive	59.38 (4.71)	4.71 (0.21)
	Shear	36.84 (3.47)	3.84 (0.14)
	Total	82.12 (7.73)	7.16 (0.29)
Total	Compressive	59.31 (5.04)	5.82 (0.23)
	Shear	38.59 (3.58)	4.35 (0.16)
	Total	84.09 (7.54)	8.825 (0.33)

Appendix B

Descriptive Results for Research Question 3 and 4

Table B.1
Descriptive Statics for Research Question 3 and 4

Helmet Liner	Impact Location	Neck Strength	PRLA (g)	HIC
TPU	Front	80PM	93.64 (28.48)	191.06 (114.77)
		50PM	81.30 (28.17)	171.17 (116.41)
		50PF	79.28 (34.97)	128.92 (92.67)
		30PF	93.72 (93.73)	146.33 (92.55)
		Total	86.99 (31.61)	159.38 (104.35)
	Side	80PM	84.37 (42.81)	157.95 (131.40)
		50PM	79.46 (39.78)	137.17 (109.63)
		50PF	86.01 (49.44)	181.25 (155.25)
		30PF	90.24 (52.37)	180.12 (159.32)
		Total	85.027 (46.10)	164.13 (138.90)
	Back	80PM	131.50 (67.97)	243.16 (219.26)
		50PM	154.28 (64.60)	314.99 (233.32)
		50PF	152.52 (65.47)	312.28 (227.89)
		30PF	144.76 (73.96)	286.84 (233.64)
		Total	145.77 (68.01)	289.32 (228.53)
	Total	80PM	103.17 (46.42)	197.39 (155.15)
		50PM	105.01 (44.18)	207.78 (153.12)
		50PF	105.94 (49.96)	207.49 (158.93)
		30PF	109.57 (53.72)	204.43 (161.83)
		Total	105.92 (48.57)	204.27 (157.26)
Standard	Front	80PM	107.33 (43.32)	232.16 (158.27)
		50PM	86.13 (33.44)	165.08 (114.13)
		50PF	84.30 (40.30)	135.57 (108.88)
		30PF	84.23 (36.11)	143.44 (102.55)
		Total	90.50(38.29)	169.07 (120.95)
	Side	80PM	83.16 (61.66)	148.15 (153.53)
		50PM	96.89 (65.23)	176.15 (164.28)
		50PF	81.78 (57.94)	144.71 (156.74)
		30PF	90.35 (58.91)	160.57 (151.86)
		Total	88.05 (60.93)	157.40 (156.63)
	Back	80PM	133.83 (55.81)	272.80 (194.15)
		50PM	127.16 (58.94)	264.57 (196.152)
		50PF	129.03 (71.19)	270.18 (224.04)
		30PF	123.22 (61.85)	248.78 (191.82)
		Total	128.31 (61.95)	264.09 (201.73)
	Total	80PM	108.11 (53.60)	217.71 (168.91)
		50PM	103.40 (52.53)	201.94 (158.22)

		50PF	98.38 (56.48)	183.50 (163.22)
		30PF	99.27 (52.29)	184.27 (148.74)
		Total	102.29 (48.57)	196.85 (159.77)
Total	Front	80PM	100.49 (35.9)	211.61 (136.52)
		50PM	83.72 (30.80)	168.13 (115.27)
		50PF	81.79 (37.63)	132.26 (101.27)
		30PF	88.98 (35.47)	144.89 (97.55)
		Total	88.75 (34.95)	164.22 (112.65)
	Side	80PM	83.77 (52.23)	153.05 (142.46)
		50PM	88.18 (52.50)	156.67 (137.00)
		50PF	83.90 (53.68)	162.99 (155.99_
		30PF	90.30 (55.64)	170.35 (155.59)
		Total	86.54 (55.52)	160.76 (147.76)
	Back	80PM	132.67 (61.89)	257.98 (207.11)
		50PM	140.72 (61.77)	289.78 (214.73)
		50PF	140.78 (68.33)	291.24 (225.96)
		30PF	134.00 (67.91)	267.82 (212.13)
		Total	137.04 (64.98)	276.70 (215.13)
	Total	80PM	105.64 (50.01)	207.55 (162.03)
		50PM	104.21 (48.36)	204.86 (155.67)
		50PF	102.16 (53.22)	195.49 (161.08)
		30PF	104.43 (53.00)	194.35 (155.28)
		Total	104.11 (51.15)	200.57 (158.52)

Table B.2
Key findings for PRLA and HIC

Measure	Location	Neck Strength Level	Liner
PRLA	Front	80th Percentile Male	TPU
		30th Percentile Female	Standard
	Back	50th Percentile Male	Standard
		50th Percentile Female	Standard
		30th Percentile Female	Standard
HIC	Front	80th Percentile Male	TPU
	Side	50th Percentile Female	Standard
	Back	50th Percentile Male	Standard
		50th Percentile Female	Standard
		30th Percentile Female	Standard