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

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Abstract

Advances in helmet designs and technology have not completely eliminated the incidence of mild traumatic brain injuries (mTBI) in the sport of ice hockey. One issue that concerns researchers is that the vinyl nitrile (VN) and expanded polypropylenes (EPP) materials used to manufacture hockey helmets have remained the same since the implementation of mandatory helmets in the 1970s, due to the cost and capability of these materials to meet the aesthetic demands of the public. One potential avenue to address this concern is to use new thermoplastic polyurethane (TPU) lining materials in existing helmet shells to better mitigate impact forces while maintaining the aesthetic structure of the helmet. The TPU material has the desirable qualities of being tough, meaning that it is resistant to tearing. The TPU material also has highly elastic properties meaning that it can deform to absorb energy and quickly return to its original shape. Hence, the first purpose of this study was to compare the capability of the 3D printed TPU lining inserts to absorb energy during static loading to commercially available liners made from VN and EPP. The second purpose of this study was to determine the effectiveness of 3D printed TPU liners in reducing linear acceleration and risk of head injury from a simulated impact when compared to traditional liner materials during dynamic testing. The results from the static testing using a Chatillon® TCD1100 Force Tester, indicated that the TPU liner absorbed 38.6% of the loading energy, which fell between the EPP (41.8%) and VN (15.8%) indicating that it performs approximately as well as, commercially available liners. The results from the simulated dynamic impacts to the head at five different locations, as defined by the NOCSAE protocol, revealed a significant interaction effect between impact location and liner type on measures of linear acceleration and risk of head injury. When examining the interaction effect, the results indicated that the EPP liner performed the best at the front and front boss locations,

while the TPU liner performed the best at the side, rear boss, and rear locations. The primary implication of these results is that a hockey helmet lining material with multilayer characteristics composed of TPU and EPP structures would better mitigate impact forces from all locations to minimize the risk of concussions.

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Introduction

The sport of ice hockey is very popular in Canada, the United States, Russia; and Central and Eastern Europe. The use of skates allows players to move on the ice at high speeds. The combination of high speeds, collisions with other players, and hard surfaces such as the ice, boards and net, increase the risk of injuries for players (Goodman, Gaetz, & Meichenbaum, 2001). One injury commonly sustained among athletes while playing the sport of ice hockey relates to concussions (Cantu, 1996). A concussion is a mild traumatic brain injury (mTBI), which originates due to a pathophysiological process caused by biomechanical forces that affect the brain functioning (Benson, Meeuwisse, Rizos, Kang, & Burke, 2011).

Currently, helmets offer the best protection against concussions for ice hockey players and are mandatory in organized leagues (Hockey Canada, 2018). Helmets also protect ice hockey players against skull fractures and lacerations. Manufacturers assess the protective capability of the helmets using testing protocols. These helmet testing protocols exist to create a minimum standard to certify the safety of the hockey helmets to protect the head against brain injuries before selling them to consumers.

These hockey helmet testing protocols entail mounting the helmet on a surrogate headform designed to simulate the response of an actual human head when impacted at six different locations using either a vertical or horizontal impactor. Accelerometers mounted inside the headform measure the helmet's ability to limit peak linear acceleration from impacts to the head.

While the majority of the helmet testing standard protocols are based on the use of free falling vertical impactors, horizontal pneumatic impactors have also become more common in recent research studies as they more accurately recreate mechanisms of injury causing concussions due to a collision with an opposing player (Delaney, Al-kashmiri, & Correa, 2014).

Unfortunately, despite the introduction of helmets, testing protocol and new improved designs in helmet technology over the last 40 plus years, the incidence rate of concussions remains unacceptably high (Benson et al., 2011; Goodman et al., 2001). Part of the issue relates to limitations in current helmet designs and materials to accommodate public demands to not only protect the head against concussion but also satisfy consumers in terms of helmet shell shape/geometry for aesthetic purposes, which continues to happen at a much faster rate than the development of better liner materials to improve helmet protective capabilities.

One potential avenue to improve the performance of current hockey helmet technologies in mitigating impacts to the head involves the use of lining inserts made of thermoplastic polyurethane (TPU) material in combination with the inner attenuation liners of existing hockey helmets. The TPU material belongs to a unique category of plastic created when a polyaddition reaction occurs between a disocyanate and one or more diols (Huntsman Ltd., 2010). The TPU material is a versatile polymer first developed in 1937, which gets soft and becomes malleable when heated and hard when cooled. The TPU material is known for its high elongation and tensile strength as well as its elasticity (Huntsman Ltd., 2010). This approach seems to be more cost effective in ameliorating the performance of current helmet technologies rather than making drastic changes to the entire helmet designs to accommodate public demands.

To address this gap in existing hockey helmet technologies, this study had two purposes. The first purpose was to examine the ability of the TPU lining inserts to absorb energy during static loading compared to the commercially available liners made from vinyl nitrile (VN) and expanded polypropylene polymers (EPP) materials. The second purpose of this study was to determine the effectiveness of three dimensional (3D) printed TPU hockey helmet liners in reducing acceleration and risk of head injury from a linear impact when compared to traditional helmet liner materials such as VN and EPP during dynamic testing.

The static and dynamic results of this study indicate that the TPU material seems promising in its ability to mitigate impact forces. The static testing showed that the TPU material was able to absorb a similar amount of energy as the commercial liners during static loading. The dynamic testing showed that the TPU material was able to reduce the peak resultant linear acceleration and the head injury criteria (HIC) better than the commercial liners at 3 of the 5 tested locations. The locations where the TPU outperformed the VN and EPP liners were the side, rear boss, and rear locations, which have been linked to a greater risk of concussion (Liao, Lynall, & Mihalik, 2016).

From the theoretical perspective, this study adds to the knowledge base on the effects of different ice hockey helmet liner materials, on measures of peak resultant linear acceleration and risk of head injury during simulated dynamic impacts to the head to mitigate concussion risk. From the practical perspective, this study opens an avenue for potential advancements in helmet performance using 3D printed TPU as a liner material to improve helmet performance in mitigating the risk of head injury without changing the structure of the commercial helmet. This study also provides evidence that multi material liners could be the most effective at mitigating peak resultant linear acceleration and risk of head injury depending on the location of the impact to the head.

Literature Review

Ice hockey is a popular sport played around the world. Players' safety has become a major priority at professional and recreational levels due to the number of head injuries that transpired and continue to occur while playing the sport (Goodman et al., 2001). Ice hockey is considered a fast-paced, contact sport played by over 630,000 people in Canada alone (International Ice Hockey Federation, 2017). The attribute that separates ice hockey from other contact sports (i.e., football, lacrosse) is the combination of players moving at high speeds, around 30 mph skating and up to 15 mph sliding (Goodman et al., 2001), as well as, the number of hard surfaces players come in contact with such as the ice, glass, boards, goal posts, and other players (Goodman et al., 2001). This combination of high speeds and hard surfaces leads to a high incidence of head and neck injuries (Goodman et al., 2001).

Head Injuries

Head injuries are common in sports such as ice hockey and can be classified into focal or diffuse injuries based on the impact mechanism that causes the head trauma (Graham, Rivara, Ford, & Spicer, 2014). Focal injuries originate by direct blunt trauma to the head (Graham et al., 2014). This trauma occurs when a stick or puck strikes the player's head, the player falls to the ice, or the player hits the boards or another player. The trauma to the head can in some cases, result in a skull fracture or intracranial haematoma caused by bleeding inside the skull (Biasca, Wirth, & Tegner, 2002). Furthermore, improper use of sticks (high sticking) seems to be another major cause of this type of head injury in the sport of ice hockey (Murray & Livingston, 1995).

Diffuse injuries, on the other hand, originate by the inertial effect of a direct impact to the head (Graham et al., 2014). Although cerebral spinal fluid is present inside the cranium to act as a shock absorber, it does not effectively mitigate the shearing and tensile forces acting on the brain (Biasca, Wirth, Maxwell, & Simmen, 2005). This type of head injury produces more widespread disruptions of neurological function (Biasca et al., 2002) such as concussions, which are the most commonly reported athletic head injury (Cantu, 1996).

Concussions in the Sport of Ice Hockey

A concussion is a mild traumatic brain injury (mTBI), which results in a pathophysiological process caused by biomechanical forces that affect the brain functioning (Benson et al., 2011). Initially, it was thought that concussions resulted in only a temporary change in brain function caused by neuronal, chemical, and neuro-electrical changes without structural damage. Later, researchers determined that structural damage causing irreversible loss of brain cells occurred in some concussions, but not always (Cantu, 1996). According to the American Centers for Disease Control and Prevention (CDC) concussion symptoms generally fall into one of four categories. First is "Thinking/Remembering", this includes things like difficulty thinking clearly, feeling slowed down, difficulty concentrating, and difficulty remembering new information. The second category is "Physical" which includes things like headache, fuzzy or blurry vision, nausea or vomiting, dizziness, sensitivity to noise or light, balance problems, feeling tired, and having no energy. The third category is "Emotional/Mood" and includes things like irritability, sadness, being more emotional, and nervousness or anxiety. The last category is "Sleep" which includes sleeping more than usual, sleeping less than usual, and trouble falling asleep.

Concussion Management

Concussions are serious head injuries and should be treated as such. They can affect an athlete's health as well as his/her career if symptoms persist, limiting his/her ability to return to play. Time lost due to injury can cost players and teams millions of dollars at the professional level. Even with recent pushes for increased concussion awareness, concussion incidence rates are likely higher than what is seen in research, since many concussions go unreported (Graham et al., 2014). Many reasons exist for not reporting concussions. For example, athletes may withhold their symptoms for fear of being pulled from a big game or may not even recognize the signs and symptoms of a concussion themselves. Teammates, coaches, or parents may also place pressure on athletes to return to the field of play while experiencing symptoms of a concussion.

In Ontario over the last year, there has been a push to increase awareness of not only signs and symptoms of concussions, but the importance of being removed from play under the guidance of *Rowan's Law*. It states that it is mandatory for sports organizations to do three things at the start of every season: first, ensure that athletes under 26 years of age, parents of athletes under 18 years, coaches, team trainers, and officials confirm every year that they have reviewed Ontario's Concussion Awareness Resources. Second, establish a concussion code of conduct that sets out rules and behaviours to support concussion prevention. Third, establish a removal-from-sport and return-to-sport protocol ("Rowan's Law: Concussion Safety", 2019). Rowan's Law is named after a high school rugby player from Ottawa, Ontario named Rowan Stringer, who died in 2013 as a direct result of second impact syndrome; she was believed to have suffered three concussions over a period of six days ("Rowan Stringer's Story", 2019). Neither she, nor her teachers, coaches, or family knew that her brain needed time to heal. Second impact syndrome occurs when an athlete experiences a second blow to the head soon after suffering an acute concussion. There is strong evidence to suggest that after a concussion, brain cells can remain alive, but in a vulnerable state for an undetermined amount of time, which predisposes the athlete to an increased risk of severe brain injury from a second impact (Biasca et al., 2005).

Concussion Incidence Rates

The incidence rates of concussions in the sport of ice hockey can be difficult to accurately determine, especially at an amateur level due to the lack of resources such as dedicated team physicians and medical staff. One study published in 2006, examined injury incidence rates among minor hockey players in Alberta. In this study, 986 players between the ages of 8 and 17 years were monitored throughout the regular season. A certified athletic therapist performed weekly assessments of any hockey injuries (Emery & Meeuwisse, 2006). The researchers found that 21.9% of players reported at least one injury, with an overall injury incidence rate of 30.02 injuries per 100 players per season. Of the 296 recorded injuries, concussions accounted for 18%. Although the concussion injuries were slightly less common than ligament sprains and contusions, they were more prevalent than muscle strains, fractures, dislocations, and cuts/bleeding (Emery & Meeuwisse, 2006). The prevalence of injuries by type can be seen in Figures 1 and 2 below.

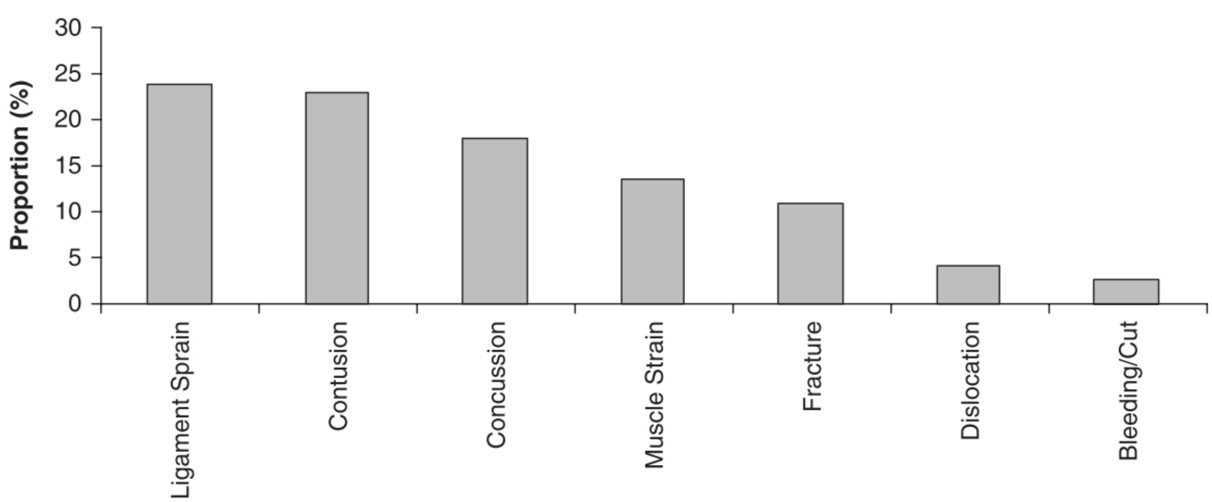

Figure 1. Injury by type among minor hockey players in Alberta. This bar graph depicts the proportion of the total reported injuries that can be classified into each of these main categories. Adapted from "Injury Rates, Risk Factors, and Mechanisms of Injury in Minor Hockey", C. Emery, W. Meeuwisse, 2006 *American Journal of Sports Medicine 34*(12), p. 1960-1969.

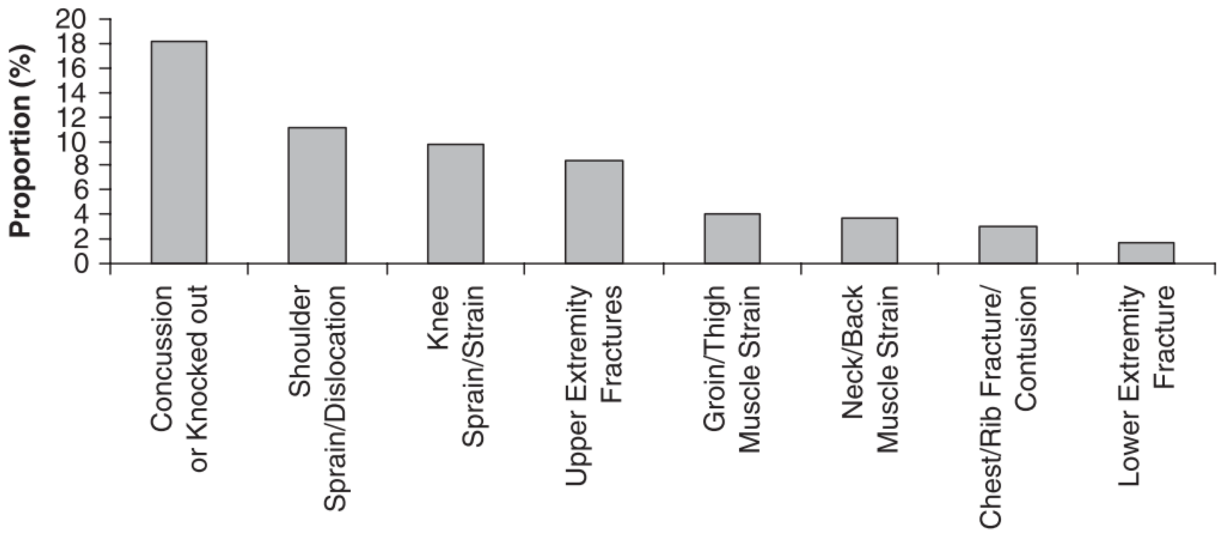

Figure 2. Injury by specific injury type among minor hockey players in Alberta. This bar graph depicts the proportion of total injuries that are made up by each specific injury type and region. Adapted from "Injury Rates, Risk Factors, and Mechanisms of Injury in Minor Hockey", C. Emery, W. Meeuwisse, 2006 *American Journal of Sports Medicine 34*(12), p. 1960-1969.

A meta-analysis published in 2016 aimed to determine concussion incidence rates across many different sports (Pfister, Pfister, Hagel, Ghali, & Ronksley, 2016). The researchers initially found 698 studies and narrowed them down to 81 for a full text review. After a full text review of each article, 58 were excluded leaving 23 unique studies. The researchers found that 12 different sports were included in more than one study. Of the 12 sports, only rugby had a higher concussion incidence rate than ice hockey (Pfister et al., 2016). Furthermore, the studies indicated that incidence rates in ice hockey were even higher than many sports traditionally associated with concussions such as American football and lacrosse. A graphical representation of the concussion incidence rates for each sport can be seen in Figure 3. Each study included in the meta-analysis is listed on the left side and grouped by sport. The sports with the higher incidence rates of concussions appear higher on the list. The X-axis indicates the pooled incidence rate of concussions, with higher values being further to the right. Each study's observed value for concussion incidence rate is depicted as a dot on the spectrum.

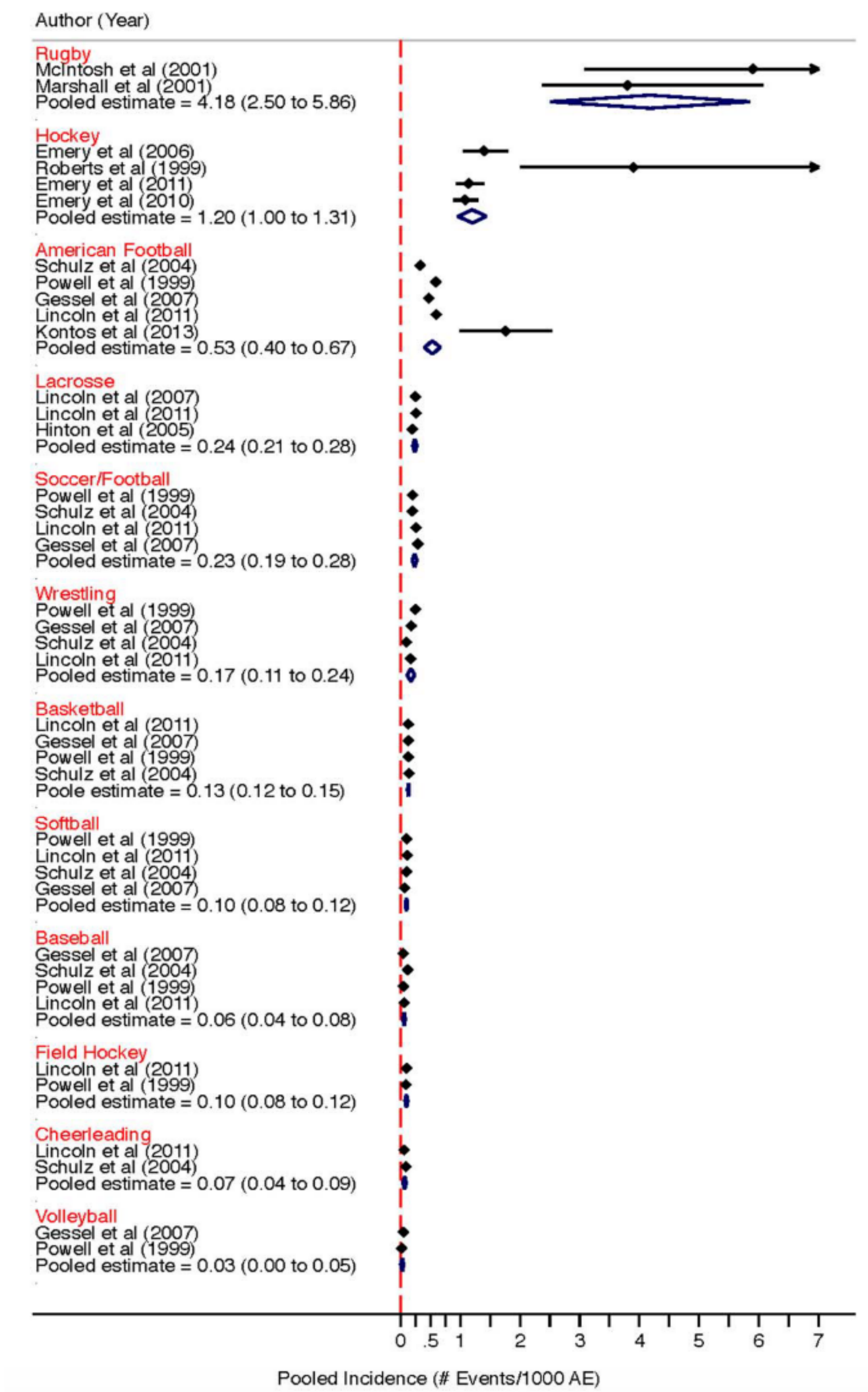

Figure 3. Comparing the pooled estimate of concussion incidence rate across different sports. The results of the meta-analysis are summarized in this graph, the sports with the highest incidence rates of reported concussions are closer to the top of the list. Adapted from "The incidence of concussion in youth sports: a systematic review and meta-analysis", T. Pfister, K. Pfister, B. Hagel, W. Ghali, and P. Ronksley, 2016, *British Journal of Sports Medicine, 50*, p. 292-297.

Another study published in 2011, with the approval of the National Hockey League (NHL) and the NHL Players Association, provided an approach to collect information on the incident rate of concussions via the team physicians from all 30 NHL teams (Benson et al., 2011). The physicians reported every concussion sustained in a regular season game from the start of the 1997/98 season, through the end of the 2003/04 seasons, for a total of 7 seasons. At the end of the data collection process, the physicians reported a total of 559 diagnosed concussions (Benson et al., 2011). This outcome worked out to a mean of 80 diagnosed concussions per year with 5.8% of players being diagnosed per season and an incidence rate of 1.8 diagnosed concussions for every 1000 game player hours (Benson et al., 2011). The researchers concluded that the severity of the concussion depended on the magnitude of the impact and the mechanism of injury.

Mechanisms of Injury Causing Concussions

In the sport of ice hockey, two primary mechanisms of injury seem to result in mTBI causing mental and physical impairment to players. These mechanisms of injury include a direct impact to the head from a puck, stick, or another player, or an indirect impact such as a whiplash associated disorder causing acceleration/deceleration of the head and neck (Graham et al., 2014). Whether the head is directly impacted or the impact occurs elsewhere on the body, acceleration of the head can occur and in both cases, the injury results from the inertia produced by the manner in which the head is moved (Biasca et al., 2005). Delaney, Al-Kashmiri and Correa (2014) examined the mechanisms of injury for concussions in university football, soccer, and ice hockey players. Over a 10-year period, there were a total of 226 concussions reported for 170 different athletes. The most common impact location for a concussive impact was the side of the head, with 41%, 37.3%, and 50% of concussions in football, ice hockey, and soccer, respectively, coming from blows to the side of the head (Delaney et al., 2014). Eight of the 226 concussions occurred due to impacts to parts of the body other than the head, meaning that the overwhelming majority of concussions reported in this study were due to direct impacts to the head. Among the ice hockey players, 52.9% of the concussions came as a result of being struck by an opponent, and 45.1% were as a result of striking another object, such as the boards, goalpost, ice, or stick (Delaney et al., 2014). Since most concussions appeared to be a result of a direct impact to the head, this outcome highlighted the potential for an improvement in helmet technology to make a meaningful impact on concussion incidence rates in ice hockey.

Concussions originating from these mechanisms of injury, however, result in not only short-term impairments such as dizziness, headache, memory loss, lack of ability to concentrate, and irritability (McEntire & Whitley, 2005) but also long-term impairments such as headaches, attention loss, mood disorders, cognitive impairments, and motor impairments (Maroon, Winkelman, Bost, Amos, & Mathyssek, 2015). Multiple concussions, on the other hand, seem to be more closely associated with chronic traumatic encephalopathy (CTE) due to tau protein deposits in areas of the brain (Caron & Bloom, 2015). Chronic traumatic encephalopathy has been linked to memory disorders such as dementia and depression (Caron & Bloom, 2015). For instance, Steve Montador was a defenceman who played in the NHL for 10 years. He died in 2015 and became the fifth NHL player to be diagnosed with CTE post-mortem (Nezwek & Lee, 2016). This case offers support to the claims that multiple concussions may put NHL players at greater risk of developing CTE (Nezwek & Lee, 2016). The frequent occurrence of concussions in the sport of ice hockey due to different mechanisms of injury opened an avenue to develop new equipment and testing measurement protocols to assess the helmet technologies capabilities to minimize the risk of brain injury and protect the well-being of the athlete.

Measures of acceleration are used in research and manufacturing to assess the performance of helmets and consequently, help improve the helmet designs. Via measures of acceleration, it is possible to determine the ability of the helmet material to mitigate the magnitude of the impact and protect the head against a traumatic brain injury. Linear acceleration is the most common measure used to assess a helmet's ability to mitigate impact forces. This measure provides a standard for helmet certification before shipping the helmets to the market (NOCSAE, 2016).

Concussions and Measures of Linear and Rotational Accelerations for Helmet Testing

Researchers also established a link between concussions and both linearly and rotationally-induced strains to the brain tissue (Forero Rueda, Cui, & Gilchrist, 2011; Kleiven, 2006). Linear acceleration is the rate of change of linear velocity produced due to a straight impact to the head (McGinnis, 2005). Rotational or angular acceleration, on the contrary, is the rate of change of angular velocity and occurs when the net force is applied as a tangential impact to the head (McGinnis, 2005). In helmet testing, accelerations are measured in units of gravitational acceleration "g", where 1g is equal to 9.8 m/s^2.

When an impact occurs to the head, the linear and rotational accelerations causing a concussion originate from three distinct forces including compressive, tensile, and shear forces (Cantu, 1996). Compressive and tensile forces act perpendicularly to the surface and relate mostly to linear accelerations and focal type injuries. Shear forces, on the other hand, are forces that act parallel to the surface and relate more to angular acceleration and diffuse type injuries (Cantu, 1996). Both linear and angular accelerations have been linked to concussions.

Rowson and Duma (2013) aimed to create a predictive threshold for suffering a concussion, using measures of both linear and angular acceleration. The researchers used a data set of previously collected data on measures of linear and rotational acceleration using instrumented helmets during play in football. The data set consisted of 62, 974 sub-concussive impacts and 37 diagnosed concussive impacts (Rowson & Duma, 2013). The researchers used a multivariate logistical regression analysis to create a risk curve as shown in Figure 4. The risk curve depicts the probability of sustaining a concussion based on the amount of linear and angular acceleration experienced during the impact.

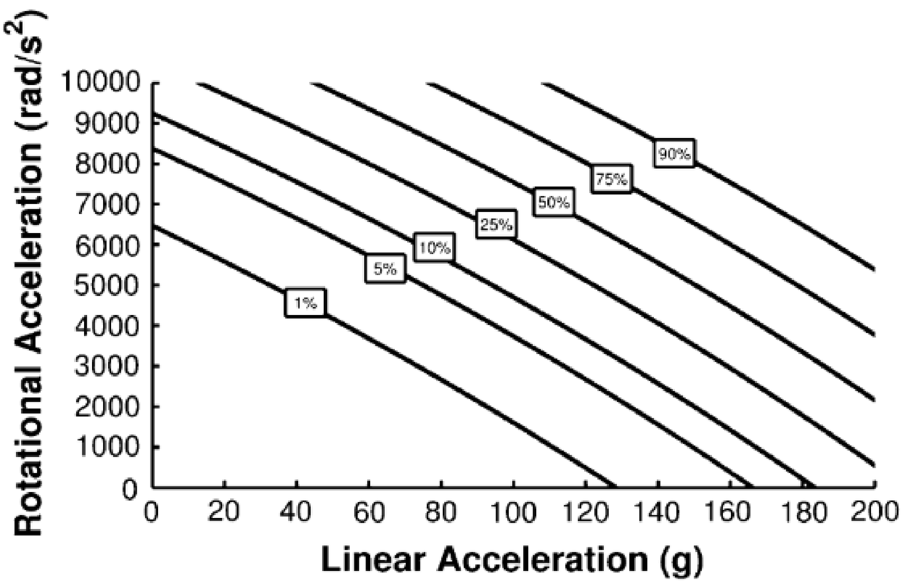

Figure 4. Risk of sustaining a concussion based on linear and angular acceleration of the head. This graph depicts the risk of injury thresholds for sustaining a concussion when including both linear and angular acceleration of the head. Adapted from “Brain Injury Prediction: Assessing the Combined Probability of Concussion Using Linear and Rotational Head Acceleration”, S. Rowson, and S. Duma, 2013, *Annals of Biomedical Engineering, 41*(5), p. 873-882.

While both measures of linear and rotational accelerations correlate to the occurrence of concussions in the real world (King, Yang, Zhang, & Hardy, 2003; Post, Karton, Hoshizaki, & Gilchrist, 2014), the combination of these two types of measures are not considered in current helmet testing standard protocols. Current helmet testing protocols for hockey helmets use measures of linear impact acceleration and risk of injury indices as standard measures to evaluate helmet performance in protecting the head against concussions (King et al., 2003). These standards also include measures of linear acceleration and risk of head injury across helmet impact locations.

Ice Hockey Helmets

As previously stated, in the sport of ice hockey, helmets offer the best form of head protection against traumatic brain injuries for hockey players (Biasca et al., 2002). Helmet designs vary greatly depending on their intended use.

A standard ice hockey helmet contains two parts: the hard plastic outer shell, which protects the head against focal injuries such as skull fractures, and the inner attenuation liner, that protects the head against rapid acceleration/deceleration (Graham et al., 2014). An image displaying the different parts of an ice hockey helmet is shown in Figure 5.

Figure 5. Parts of an ice hockey helmet. This image depicts a typical ice hockey helmet with a VN attenuation liner.

Plastic outer shell. Since the popularization of ice hockey helmets in the 1970s, the outer shells have been made with materials such as injected polyethylene, polycarbonate, or acrylonitrile butadiene styrene (ABS) plastics (Clement & Jones, 1989). The thickness of these shells in modern helmets varies depending on helmet impact location. The thickness of the shell at the front location, for example, is generally in the 2.5 – 3 mm range (Rousseau, Post, & Hoshizaki, 2009). The shape of the shells are designed to be both aesthetically pleasing and effective at spreading the energy of an impact over a larger area, which effectively prevents focal injuries and consequently concussions (Rousseau et al., 2009). Researchers found that the local geometry of the helmet shell can make a significant difference in the impact absorption properties of the helmet. More specifically, the researchers found that geometric changes in the shell could account for 4-35% of the variability in impact absorption depending on the location and angle combinations (Spyrou, Pearsall, & Hoshizaki, 2000).

Attenuation liner. The liner materials are either made of VN or EPP (Rousseau et al., 2009). The VN liners are created by cutting out specific shapes from large sheets of material, whereas EPP liners are created by pouring the material into a mould to create the desired shape (Rousseau et al., 2009). The thickness of the liner can also vary depending on the location on the helmet. For the VN lined helmets, the liner thickness is generally in the 14 - 20 mm range, while the EPP helmets generally have a liner that varies more from location to location, anywhere from 14 - 24 mm (Rousseau et al., 2009). The purpose of these liners is to undergo elastic deformation, and consequently absorb the most amount of energy possible from a collision, while returning to their original shape post impact (Rousseau et al., 2009).

Research on helmet designs for other purposes such as motorcycle or equestrian sports have recently led researchers studying ice hockey helmets to the conclusion that a liner material made of functionally graded foam could be significantly more effective at mitigating impact forces over a wider variety of impact speeds when compared to traditional single density foam liners (Cui, Kiernan, & Gilchrist, 2009). Functionally graded materials were defined as "a new generation of engineered materials wherein the microstructural details are spatially varied through a non-uniform distribution of the reinforcement phase(s), ……. with different properties, and shapes, as well as by interchanging the roles of reinforcement and matrix phases in a continuous manner" (Aboudi, Pindera, & Arnold, 1999, p 105). This essentially means that the foam will strategically vary in density throughout the thickness of the liner. The less dense foam will be able to deform and absorb energy from lower speed impacts. As the foam gets progressively denser, it will be more effective at higher speed impacts. The effectiveness of functionally graded foam is a principle that can be carried over to 3D printing technologies. Khosroshahi, Tsampas and Galvanetto (2018) used 3D printing to mimic the same principle of functionally graded foam but with polyacetic acid (PLA), which is a commonly used plastic. They used a hierarchical lattice structure to achieve the same multi-density properties as the previously used foam. This study focused on motorcycle helmets, which are designed to be used for a single impact, as such, part of the energy absorption comes from the plastic deformation of the liner material. For this reason, PLA would not be an effective material for hockey helmet liners as they need to be able to withstand multiple impacts. The majority of research conducted on hockey helmet technology includes VN liners and EPP liners.

Rousseau et al. (2009) compared three different pairs of helmets from top manufacturers. Each model offered a VN, and an EPP options. The purpose of the study was to examine the material properties of these helmets in mitigating linear acceleration, angular acceleration, and HIC, which is a risk of injury metric based on linear acceleration to the head. The researchers compared identical models with different liners to eliminate the effect of confounding variables related to the geometry or material properties of the shell, and, therefore, isolated the performance of the attenuation liner. They performed three helmet impact trials at three different speeds: 5 m/s, 7 m/s, and 9 m/s, and at three different impact locations including the front, front boss, and deflected. In this study, the front location was 0° of rotation 30 mm +/- 1 mm above the intersection of the longitudinal plane and the reference plane. The front boss location was rotated 0.916 rad clockwise, 30 mm +/- 1 mm through the centre of mass. The deflected location was created by translating the headform 6.35 cm to the right to produce an impact tangential to the centre of mass. The researchers chose these locations to be able to examine the performance of the helmets during both perpendicular and angled impacts. The results of this study indicated that the EPP liner maintained lower linear acceleration values. The VN liner, on the other hand, had lower angular acceleration values. Based on the outcome of this study, it is important to consider that although concussions correlate to measures of angular and linear accelerations as stated by King et al., (2003), helmet designs seem to perform better at mitigating linear impacts as opposed to angular impacts (Rousseau et al., 2009). Furthermore, current testing/certification protocols pass or fail criteria revolve around linear acceleration and consequently, it stands to reason that both VN and EPP mitigated linear acceleration reasonably well when the helmet was tested at different locations (Post et al., 2014).

Ice Hockey Helmet Testing Protocols and Similarities with Other Sports

Hockey helmets undergo testing before being certified (NOCSAE, 2016). Several different testing protocols exist, each with their own methodologies and pass/fail thresholds. In Canada, every player participating in a league or event under the umbrella of Hockey Canada must wear a Canadian Standards Association (CSA) certified helmet as defined by rule 3.6 b

(Hockey Canada, 2018). For a helmet to pass the CSA certification, it must limit the resultant linear acceleration to less than 275 g from a 40 Joules (J) impact, which indicates the loading energy transferred to the helmet during the collision (Halstead, Alexander, Cook, & Drew, 1998). Other helmet standard testing protocols across different sports are outlined in Table 1, which require different acceleration levels to pass or fail when compared to hockey helmets.

Table 1

Pass/Fail criteria for different helmet testing protocols

Standard	Impact Energies (J)	Pass/Fail Limit (g)
CSA Z262.1 (CSA Ice Hockey)	40	275
F1045-90a (ASTM Ice Hockey)	51	300
F429-92 (ASTM Football)	62	300
F1163-90a (ASTM Horseback)	90	300
F1447-94 (ASTM Bicycling)	96	300
F1751 (ASTM Roller Skating)	96	300
F1492-93 (ASTM Skateboarding)	52	300
M90 (SNELL Motorcycle)	150	300
NOCSAE DOC.002-96[3] (New Football helmets)	115	215
NOCSAE Lacrosse[3]	115	215

Note. Notably missing is the *National Operating Committee on Standards for Athletic Equipment (*NOCSAE) hockey helmet testing protocol which uses a threshold severity index score of 1200 (NOCSAE, 2016).

The pass/fail criteria limits in Table 1 for the CSA ice hockey and American Society for Testing and Materials (ASTM) ice hockey (first and second rows) seem to be relatively stringent. When comparing impact energies by CSA Ice Hockey and ASTM Ice Hockey to those by the NOCSAE*,* however, it becomes evident that the CSA and ASTM use a lower impact energy for helmet testing protocols. The *NOCSAE* protocol for testing lacrosse helmets, for example, allows for a maximum of 215 g (22% less than the CSA) from a 115 J impact, which is almost three times greater than the 40 J helmet impact standard used by the CSA. Ice hockey and lacrosse are similar sports in terms of the physical demands of the game, and the protective equipment used (Halstead et al., 1998). Based on these similarities, one would argue that helmet testing protocols should follow similar standards (Halstead et al., 1998). The standards for ice hockey helmets, however, differ from other sports due to the magnitude of the head impacts encountered by the athletes while playing ice hockey. In some cases, due to similarities in head impacts in different sports, some sports abide by similar hockey helmet standards with additional modifications. Box lacrosse players in Canada, for example, are permitted to use hockey helmets certified by CSA with a lacrosse specific facial shield (Canadian Lacrosse Association, 2020)

Impact Locations

Despite limitations of hockey helmet protocols when compared to other sports, hockey helmet testing standards and protocols continue to improve to minimize the risk of head injuries. In 2017, the CSA began funding research with the goal of creating a more comprehensive hockey helmet impact simulation testing protocol to be able to more accurately examine helmet's properties to mitigate the risk of sustaining a concussion. One hockey helmet protocol, which continues to evolve, is the NOCSAE protocol. Manufacturers and researchers use this protocol for testing commercial hockey helmets and studying mechanics of injury causing concussions. The NOCSAE protocol includes six impact locations defined as the front, front boss, side, rear boss, rear, and top. The front location is located at 0° (facing the impactor directly) and 25 mm above the reference plane. The front boss location is rotated 45° and 25 mm above the reference plane. The side location is rotated 90° and on the reference plane. The rear boss location is rotated 135° and on the reference plane. The rear location is rotated 180° on the reference plane. The NOCSAE protocol for newly manufactured hockey helmets also includes a "top" impact location, which is at the intersection of the coronal and median planes, with an impact angle of 90 degrees. This location is not usually assessed or included in research studies as it is responsible for a relatively small percentage of impacts among hockey players. Wilcox and colleagues (2014) found that over a period of three seasons, the impacts to the top of the head among female collegiate ice hockey players occurred only 9.5% of the time. Male collegiate ice hockey players in the same study sustained impacts to the top of their head only 7.5% of the time (Wilcox et al., 2014).

Surrogate Headforms and Impactors Used for Testing Helmets

Helmet testing protocols also include a surrogate headform to fit the helmet, which is then struck either horizontally or vertically to assess the protective capability of the hockey helmet material (McIntosh & Janda, 2003). Two main headforms are commonly used to conduct helmet testing, the Hybrid III and NOCSAE headforms as shown in Figure 6 (MacAlister, 2013).

Figure 6. Visual representation of the Hybrid III headform (left) and NOCSAE headform (right). Adapted from "Quantitative comparison of Hybrid III and National Operating Committee on Standards for Athletic Equipment headform shape characteristics and implications on football helmet fit", B. Cobb, A, MacAlister, T. Young, A, Kemper, S. Rowson, and S. Duma, 2014, *Journal of Sports Engineering and Technology, 1*(8), p. 1-8.

Initially, the Hybrid III headform was most commonly used during motor vehicle collision testing, but as sport-related concussions became more prevalent, researchers began using it along with the surrogate neckform to investigate concussions in sports such as football and ice hockey (MacAlister, 2013). The NOCSAE headform, however, is more commonly used for testing commercial helmets according to NOCSAE standards. The NOCSAE headform was developed based on research conducted at Wayne State University to more accurately simulate a human head with appropriate facial features and bone structure (MacAlister, 2013). The NOCSAE headform also comes in three adult sizes including small, medium, and large with the medium size headform being representative of a 50^{th} percentile of the male head (MacAlister, 2013). Both headform types, however, are used in helmet research under NOCSAE testing standards. Due to the anatomically correct facial features and the ability to represent a 50^{th} percentile male, the NOCSAE headform is the standard used by manufacturers along with NOCSAE protocols to test helmets in reducing risk of head injury before shipping the helmets to the market.

When testing helmets under NOCSAE protocols, researchers may either use a horizontal pneumatic impactor (Pellman et al., 2006; Post et al., 2014; Rousseau et al., 2009) or a vertical drop impactor (Di Landro, Sala, & Olivieri, 2002; Zerpa, Carlson, Elyasi, Przysucha, & Hoshizaki, 2016; Zhang, Yang, & King, 2004). The horizontal impactor operates on compressed air to propel an impact rod into the helmeted headform. The vertical impactor, on the other hand, uses gravity to accelerate the headform to strike the helmet on a solid surface such as a steel to assess the performance of the helmet according to NOCSAE protocols. Both impactors have benefits. The vertical impactor allows the researcher to simulate mechanisms of injury to the head due to the player falling on the ice (NOCSAE, 2018; Zerpa et al., 2017). The horizontal impactor, on the other hand, allows the researcher to simulate mechanism of injury to the head due to collision between players, boards, or net (Zerpa et al., 2017). Research has shown that the vast majority (88%) of concussions suffered by collegiate ice hockey players were as a result of being struck on the head by an opposing player, while only 7% of concussions were as a result of a fall (Clark, Post, Hoshizaki, & Gilchrist, 2016). Since most concussions are a result of being struck by an opposing player, the horizontal pneumatic impactor is a more accurate representation of the majority of concussive impacts in ice hockey players than the vertical drop impactor.

Both impactors allow the collection of acceleration data through the surrogate headform which contains an array of nine accelerometer sensors arranged in a 3-2-2-2 array that captures the full motion of the head in three dimensions (Rousseau et al., 2009). The impact velocities range from 3.46 - 5.46 m/s according to NOCSAE standard protocols (NOCSAE, 2016). The linear acceleration data is manipulated to compute HIC.

Head Injury Severity Indices

Different methods of estimating the severity of head injuries have been used in past research. Some of these methods include the Gadd Severity Index (GSI) and HIC. The GSI was developed in 1966 and was derived from peak linear acceleration and the Wayne State Tolerance Curve (WSTC) (Gadd, 1966). Severity indices were created as scientists began to discover that peak linear acceleration alone was not a strong predictor of concussion incidence (Hoshizaki et al., 2017). An important factor to consider when attempting to assess the risk of head injury is the duration of the acceleration. The longer a head is exposed to an acceleration, the larger the change in velocity. The higher the acceleration, the shorter an impact must be for it to be considered safe, which is why one of the primary mechanisms of helmets is to spread out the energy of the impact, causing the acceleration to be experienced at a lower magnitude, over a longer period of time (Hoshizaki et al., 2017). The equation for the GSI can be seen in Equation 1 below.

$$SI = \int_{t2}^{t1} A^{2.5}\, dt \tag{1}$$

where:

A = head acceleration impulse function; and

t_1 = impulse duration.

Gadd determined that an index of 1000 is the threshold for severe, possibly life-threatening injury. The NOCSAE protocol for newly manufactured hockey helmets indicates that no impact may exceed a score of 1200 SI, and none of the 3.46 m/s impacts may exceed a score of 300 SI (NOCSAE, 2016). The HIC, on the other hand, was initially introduced to fit the WSTC (Kleiven, 2006) and is calculated using Equation 2 (Rousseau et al., 2009).

$$HIC = \max_{t_1,t_2}\left\{(t_2 - t_1)\left[\frac{1}{t_2-t_1}\int_{t_1}^{t_2} a(t)dt\right]^{5/2}\right\} \quad (2)$$

where:

t_1= initial time in seconds;

t_2= final time in seconds; and

a= resultant linear acceleration in g.

Note that t_1 and t_2 should be chosen such that HIC is maximized. The time duration, $t_2 - t_1$, is limited to a value of 15 ms (Chichester et al., 2001). The main difference, and advantage to using the HIC instead of the GSI is that the HIC considers the duration of the acceleration (time to peak linear acceleration), which is an important variable when examining head injury risk. Previous research suggested that HIC scores of 151, 240, and 369 represent a 25%, 50%, and 80% probability of sustaining a mTBI or concussion (Zhang et al., 2004).

Research Problem

In the sport of ice hockey, however, impacts to the head do not result in only linear or angular acceleration, but the combination of both (King et al., 2003). Unfortunately, the majority of hockey helmet technologies are designed to mitigate linear impacts but not rotational impacts (King et al., 2003). Furthermore, current helmet testing protocols only include measures of linear acceleration but not rotational acceleration and consequently, new helmet technologies need to be tested on measures of linear acceleration when comparing them to existing commercial testing standards. Researchers, however, believe that developing a helmet technology that proficiently reduces both linear and angular accelerations is the ideal technology for preventing mTBI across different mechanisms of injury (King et al., 2003).

One avenue to improve current hockey helmet performance in reducing both types of acceleration is 3D printed helmet linings without the need to replace the helmet shell. New lining materials, and more importantly new lining shapes and designs, could prove to be more effective in mitigating both linear and angular accelerations.

Recent research by Soe, Martin, Jones, Robinson, and Theobald (2015), for example, showed that thermoplastic elastomer cellular structures produced by additive manufacturing for bicycle helmets were significantly more effective in mitigating impact forces. Finite element modelling, which uses computer simulated impacts to observe the behaviour of brain tissue loads (Forero Rueda et al., 2011) was used to compare different densities of cellular structures to determine which density would be optimal. The outcome of this research indicated that additive manufacturing may be an effective tool for creating high performance hockey helmet liners.

The main concern, however, is that concussion incidence rates in ice hockey continue to increase despite advances in technology for protective equipment (Halstead et al., 1998). The implications of this issue have led researchers to suggest that players are not adequately protected by current helmet technologies and consequently, there is a need to develop other liner helmet structures made out of thermoplastic material as an avenue to better mitigate linear and rotational accelerations. The initial testing of these new thermoplastic liner technologies, however, must abide by the NOCSAE testing standard protocols to be able to compare the new technology to existing commercial helmet technologies. The NOCSAE helmet testing standards only include measures of linear acceleration and risk of head injury to assess the performance of the helmets across impact locations.

Based on these issues and concerns, this study aimed to examine the effectiveness of a hockey helmet liner made from 3D printed TPU structures in reducing peak linear impact acceleration and risk of head injury from a horizontal impact. To accomplish this task, the helmet material was assessed statically by measuring the energy absorption capability before causing any damage to the material. It was also assessed dynamically to examine the capability of the helmet material to mitigate impact acceleration during inertial loading.

The first purpose of this study was to test different ice hockey helmet liner materials to compare their energy absorption abilities during static testing. Cleveland and Morris' book *Dictionary of Energy* defines energy as "the capacity to do work more generally as the potential ability of a system to influence changes in other systems by imparting work (forced directional displacement)" (Cleveland & Morris, 2015, p 196). The energy absorbed was calculated by subtracting the unloading energy from the loading energy. The final outcome was expressed as a percentage of energy stored resulting from the division of the absorbed energy by the loading energy as shown in Equation 3.

$$\%\ energy\ absorbed = \frac{loading\ energy - unloading\ energy}{loading\ energy} \tag{3}$$

The second purpose of this study was to determine the effectiveness of 3D printed TPU hockey helmet liners in reducing acceleration and risk of head injury from a linear impact when compared to traditional helmet liner materials such as VN and EPP polymers during dynamic testing. The following research questions guided the study.

Research Questions

1. What type of helmet liner material (TPU, VN, or EPP) absorbed the most energy during static testing**?**
2. Was there a significant difference between TPU, VN, and EPP liners across helmet impact locations on measures of peak linear acceleration during dynamic testing?
3. Was there a significant difference between TPU, VN, and EPP liners across helmet impact locations on measures of risk of head injury (HIC) during dynamic testing?

Method

Instruments

3D Printer. Three-dimensional printing technology is a common form of manufacturing 3D objects that can be beneficial in the design of new ice hockey helmet liners. Three dimensional printing is not all that different from traditional two dimensional printing as it creates layers on top of each other to build a 3D object (Paulo & Brans, 2013). This approach allows for helmet liners to be more geometrically complex than before and different designs/shapes can be created and tested within a relatively short period of time. This 3D printer technology was used in the current study to manufacture TPU liner inserts. These TPU helmet inserts were tested to examine their capabilities in mitigating linear acceleration and risk of injury during horizontal impacts to a surrogate headform.

Chatillon® TCD1100 Force Tester. The Chatillon® TCD1100 force tester was used to measure the stiffness and energy absorption properties of the 3D printed TPU lining inserts. The stiffness of the TPU lining inserts was determined by compressing the sample against the TLC series load cell, which measured how much force was required to deform the inserts by a given distance.

The energy absorption property was determined by compressing then releasing (or uncompressing) the TPU inserts. The set up for static testing of the VN sample is shown in Figure 5 below.

Figure 7. Set up used for static testing of the VN sample. Close up image of the VN sample during static testing with the Chatillon® TCD1100 force tester.

Plots of compressive force versus material compression were obtained for both the compressing and uncompressing stages. The enclosed area between the plots was evaluated to indicate the energy absorbed. The TCD1100 machine can be seen below in Figure 6.

Figure 8. Chatillon® TCD1100 force tester by AMETEK Inc. Image of the force tester used in this study. Adapted from "Chatillon® TCD Series Console User's Guide", Ametek Industries**,** 2008, p. 1-2

Headform. A medium sized NOCSAE headform, representative of a 50th percentile of a human head, functioned as a surrogate head to perform all impact trials. The NOCSAE headform was used for this study because it has anatomically correct facial features and bone structure, and, therefore, it is more desirable when studying equipment designed to engage the human head (MacAlister, 2013). More specifically, this headform creates a human structure that permits a more accurate simulation of the dynamic response of a human head during impact than previously used metal headforms and it is more rugged and repeatable than cadaver heads (Hodgson, 1975). The anthropometric measurements of the NOCSAE headform can be seen in Table 2 below.

Table 2

Anthropometric measurements of the NOCSAE headform in inches and (mm);(NOCSAE, 2018).

POINTS OF MEASURE	HEADFORM SIZES		
	SMALL (6 5/8)	MEDIUM (7 ¼)	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 head) …...……........	7.09 (180)	7.87 (200)	8.15 (207)
Outside eye corner (external canthus) to back of head…………	6.22 (158)	6.81 (173)	7.32 (186)
Ear hole (tragion) to back of head……………………………...	3.50 (89)	3.86 (98)	4.25 (108)
Ear hole to outside corner of eye (tragion to ext. canthus) ...….	2.72 (69)	2.95 (75)	3.07 (78)
Ear hole to top of head (tragion to vertex) ……………...……...	4.72 (120)	5.24 (133)	5.67 (144)
Eye pupil to top of 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 nose to point of chin (subnasal to menton) …...……	2.56 (65)	2.80 (71)	3.03 (77)
Top of nose to point of 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…………………........	9.08lb (4.12kg)	10.8 lb (4.90kg)	13.08lb (5.93kg)

The NOCSAE headform as shown in Figure 8 was instrumented with a mechanical neckform rather than a rigid arm (MacAlister, 2013). This approach was used in this study as current helmet testing protocols in research studies often include a mechanical neckform to be used in conjunction with the headform. The customized neckform used in this study and described below was previously validated by Carlson et al. (2016) to more accurately represent an impact to a human head.

Figure 9. Photograph of the NOCSAE headform used in all impact trials.

Mechanical neckform. This mechanical neckform was created by the Lakehead University Mechanical Engineering Department in conjunction with the School of Kinesiology to be representative of a 50^{th} percentile human neck. Neoprene and steel disks alternate to simulate the intervertebral joints. A small cut-out on the anterior side, and a larger cut-out on the posterior side as shown in Figure 9 allow the neckform to move through a range of motion to simulate the movement and muscle strength of a human head during impact.

Figure 10. Computer model of the mechanical neckform used for all impact trials.

A steel cable runs longitudinally through these disks and can be tightened or loosened by a bolt at the end. The amount of torque applied to this bolt affects the stiffness of the neckform and can be used to represent participants with different neck strengths. This neckform has been used in previous research (Carlson, 2016; Jeffries, Carlson, Zerpa, Przysucha, & Sanzo, 2017; Zerpa et al., 2016). A standard torque value was used in this study based on previous research (Carlson, 2016; Rousseau & Hoshizaki, 2009). Carlson (2016) calculated the low and high stiffness values to represent the 30th percentile and 80th percentile from the 50th percentile of standardized neck stiffness data. The results of these calculations can be found in Table 3 below.

Table 3

Torques required for different stiffness conditions (Carlson, 2016).

Stiffness	Torque (in-lb)	Torque (N-m)
Low	8.4	0.949
Standard	12	1.356
High	15	1.763

Interlink accelerometers and Labchart7 software. Wireless accelerometers arranged in a cube formation were instrumented in the NOCSAE headform shown in Figure 7 to measure linear acceleration in three mutually perpendicular x, y, and z directions. Accelerometers are sensors that measure acceleration from reactionary impact forces (Winter, 2009). That is, a mass is accelerated against a force transducer that produces a signal voltage (Winter, 2009). The high frequency Interlink Electronics accelerometers measure the change in velocity of the surrogate headform over time in three directions (x, y, and z). The accelerometers were connected to a power supply and amplifier unit model 482A04 to acquire the three analog signals as shown in Figure 10.

Figure 11. Power supply unit with amplifier for three channel signal processing.

The analog signals were interfaced to a Powerlab™ unit with an analog to digital converter to acquire the data for each impact test. The Labchart7 software was used to collect, analyze, and interpret the raw accelerometer impact data. The three acceleration signals (a_x, a_y, and a_z) were combined to produce the resultant acceleration as shown in Equation 4 to assess each impact to the surrogate headform during the helmet testing procedure.

$$a_r = \sqrt{a_x^2 + a_y^2 + a_z^2} \tag{4}$$

where:

a_r = resultant acceleration;
a_x = acceleration in the x axis;
a_y = acceleration in the y axis; and
a_z = acceleration in the z axis.

A lowpass filter of 1000 Hz was applied to the resultant acceleration to eliminate any possible high frequency noise that may originate on the impactor frame during each head collision, which can affect the data.

Pneumatic linear impactor. The horizontal impactor was designed in a collaboration between the Lakehead University School of Kinesiology and Mechanical Engineering Department. To provide evidence of reliability, Jeffries et al. (2017) performed 100 impacts each at three different locations (front, side, and back). The split half method was then used to show evidence of reliability between the even and odd numbered trials. An intraclass correlation showed strong correlations between the even and odd numbered trials, with correlation values of .86, .79, and .81 for the front, side, and rear locations, respectively. To establish evidence of validity, the measures of linear acceleration obtained from the horizontal pneumatic impactor were compared to those obtained from the previously validated (Carlson et al., 2016) vertical drop impactor with an impact speed of 4.39 m/s. The intraclass correlation coefficients (ICC) were .95, .85, and .88 for the front, side, and rear, respectively, providing strong evidence of validity.

The horizontal impactor consists of a main frame as shown in Figure 11, secured to the floor. It contains a headform assembly as shown in Figure 12, which is connected to a sled on rollers to allow for realistic after impact mechanics. The main frame holds a 3-gallon compressed air tank, a quick release solenoid valve pressure switch, and the impacting rod. The mechanical solenoid valve is used to control the release of the compressed air in the tank, which propels the impact rod into the head assembly. A digital pressure gauge mounted to the air tank allows the user to fill in the tank with the appropriate compressed air to generate accurate impact speeds ranging from 0-7 m/s, accurate to +/- 1% (Jeffries, 2017). Releasing the compressed air propels the 13.1 kg impact rod horizontally towards the target. The greater the pressure, the greater the impact speed. Weights can be added to the sled to adjust the inertia of the head assembly.

Figure 12. Main frame of the horizontal pneumatic impactor.

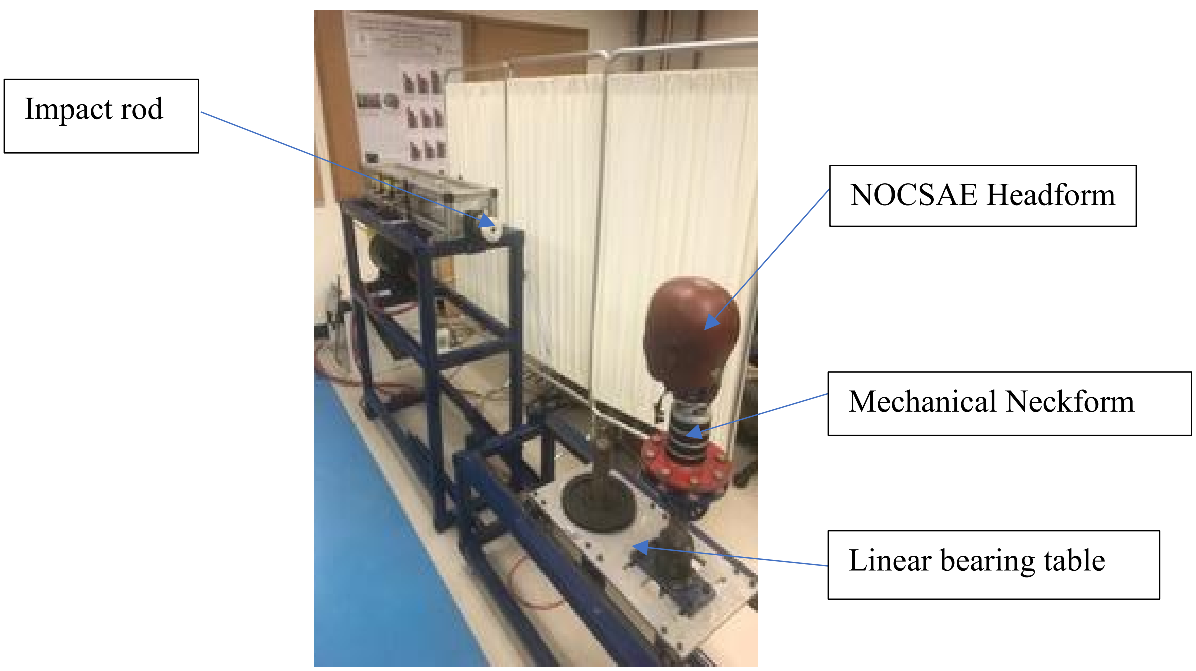

Figure 13. Main frame and headform assembly of the horizontal pneumatic impactor.

Helmets. Three CCM hockey helmets were used for testing. One helmet contained a VN foam liner, the second helmet contained an EPP liner, and the third helmet contained a liner made of 3D printed TPU. The third helmet was created by first removing the liner from an identical helmet shell, then attaching the TPU inserts individually using adhesive Velcro. A photo of the helmet with the TPU liner can be seen in Figure 13 below.

Figure 14. TPU lined helmet used for this study.

The VN and EPP helmets were used to establish baseline values for helmets that are currently commercially available. The performance of the helmet with the 3D printed TPU liner was then compared to the commercially available helmets.

Procedures

Static testing. First, the three liner material samples were tested using the Chatillon® TCD1100 force tester. The samples were compressed by 10 mm against a force gauge.

The protocol included 10 trials per sample to compress and decompress the material with the application of force during static testing. The raw force versus displacement data was exported to Microsoft® Excel®. The loading and unloading energy were calculated for each cycle. The unloading energy was subtracted from the loading energy to calculate the energy absorbed by the material. This value was presented as a percentage of loading energy as shown in Equation 3. The energy absorbed was then compared between the TPU, VN, and EPP liners.

Dynamic testing. Before beginning the trials, the stiffness of the surrogate neckform was adjusted by torqueing the longitudinal cable to the "standard" 12 inch-pounds of torque or 1.356 Nm as used in previous research (Carlson, 2016). A custom jig built by Pennock (2018), depicted in Figure 14 was used to keep the headform from rotating, as well as, ensuring that the force was applied perpendicular to the nut to adjust the neck to the appropriate torque.

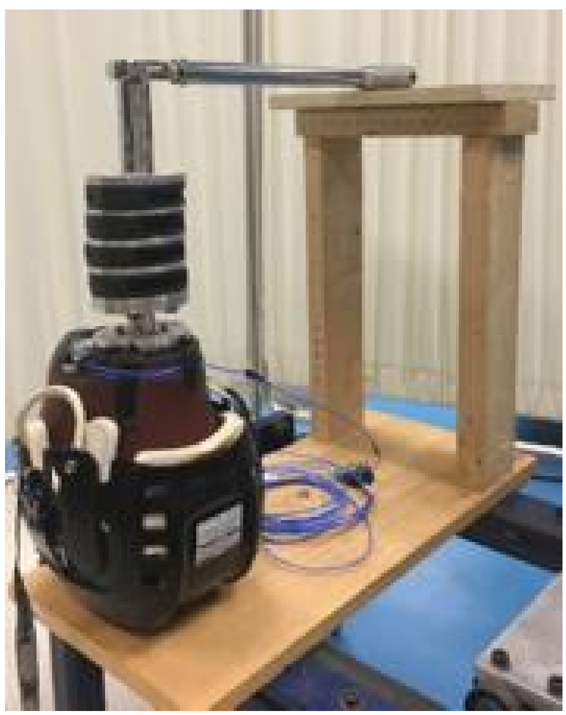

Figure 15. Custom built jig used to accurately torque the longitudinal bolt of the mechanical neck form.

Since a torque of 12 in-lb (1.36 N-m) is too low to measure with a standard torque wrench, a force gauge was used to ensure the proper amount of torque was being applied to the nut. The force gauge only reads whole numbers, so some calculations were necessary to ensure the proper amount of torque was applied. The integer of force that can be applied closest to the end of the wrench without going over, is 4 N.

$$T = F \cdot D \tag{5}$$

$$1.36\ Nm = (F)(0.44\ m)$$

$$F = 3.09\ N$$

where:

T = torque;
F = applied Force; and
D = length of the moment arm.

Once it was determined that 4 N needed to be used, Equation 5 was used again to determine the necessary length of the moment arm.

$$1.36\ Nm = (4\ N)(D)$$

$$D = 0.34\ m$$

Therefore, to achieve 12 in-lbs of torque, 4 N of force was applied to the wrench, 34 cm from the axis of rotation. Once the neck was appropriately torqued, it was attached to the headform assembly using 8 bolts as shown in Figure 15.

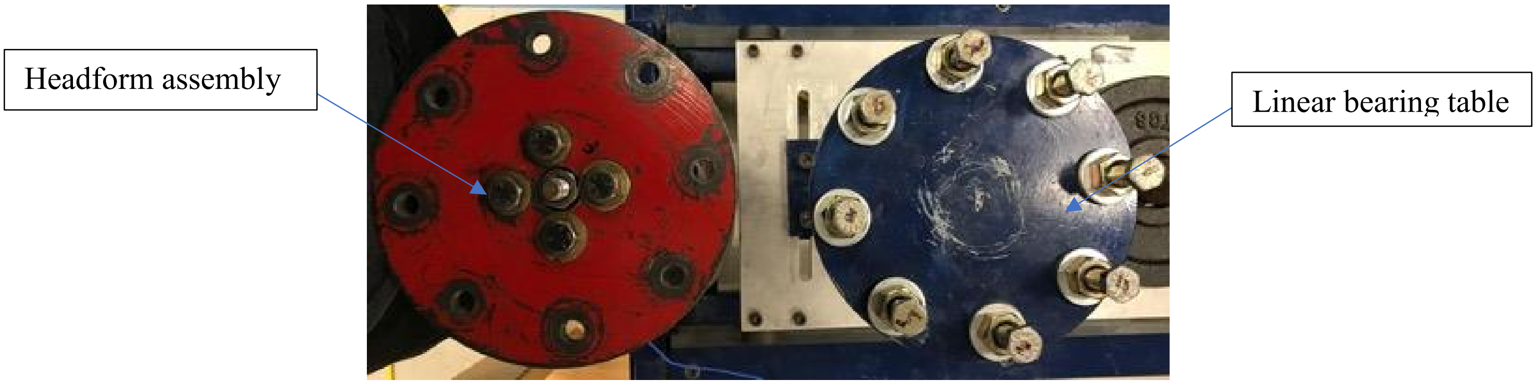

Figure 16. Steel mounting plates for attaching the headform assembly to the linear bearing table.

The neck was re-torqued when switching impact locations (front to front boss, front boss to side, side to rear boss, and rear boss to rear) to ensure consistency in the neck stiffness across trials. All five impact locations are depicted in Figure 16.

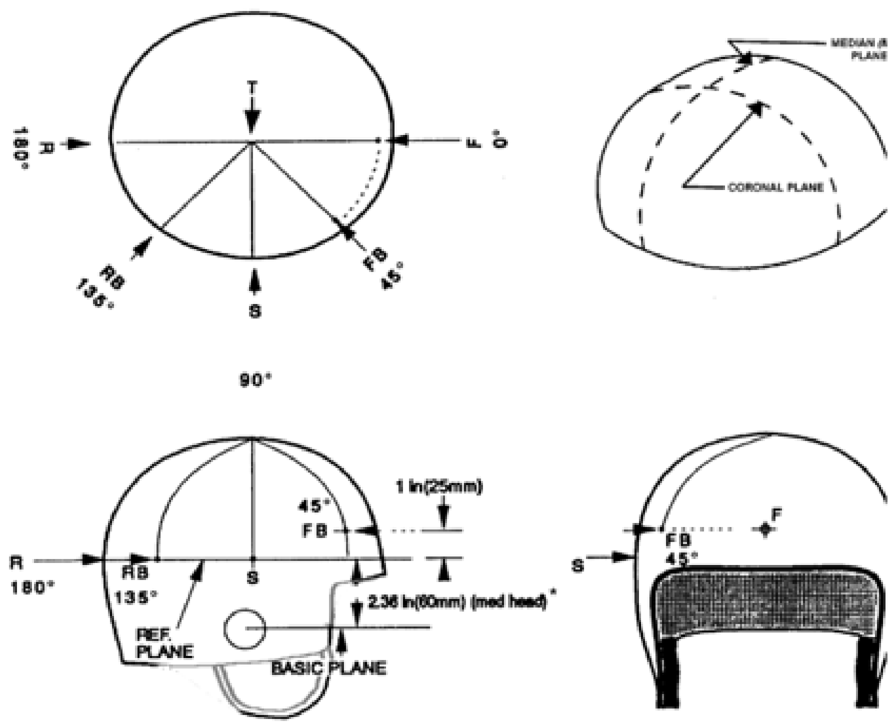

Figure 17. Visual representation of all 5 head impact locations used in the NOCSAE protocol from "Standard Performance Specification for Newly Manufactured Ice Hockey Helmets," by NOCSAE 2016.

Once the NOCSAE headform was correctly aligned with the impact rod, the helmet was fitted to the headform using the manufacturer's fitting instructions. Variability in the data was created by using 18 different impact speeds to represent different participants and/or to simulate different impact scenarios. That is, each combination of three helmet liner type (VN, EPP, and 3D printed TPU), and five impact locations (front, front boss, side, rear, and rear boss) was impacted 18 times at different speeds ranging from 2.01 m/s to 5.13 m/s, for a total of 3 × 5 × 18 = 270 impacts.

Impact velocities can be calculated using Equation 6 to create the velocity data based on the compressed air pressure in the tank of the horizontal impactor as shown in Table 4 (Jeffries, 2017). This range of velocities was used in the current study.

$$Velocity = 0.00005(psi)^3 - 0.0063(psi)^2 + 0.3307(psi) - 2.9423 \quad (6)$$

where:

Velocity = the velocity of the impact rod upon contact with the helmeted headform; and

psi = air pressure in the tank measured in pounds per square inch.

Table 4

Tank pressure and corresponding impact velocity (Jeffries, 2017).

Simulee	Pressure (PSI ± 1%)	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

For each impact, the resultant linear acceleration, measured in g with a sampling rate of 20 kHz, was computed using Equation 4 via Labchart 7 software. HIC was then calculated using Equation 2. To ensure the maximum 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 for HIC. The raw acceleration vs time data for the peak resultant linear acceleration was copied into Matlab® one impact at a time, where the script was used to determine the appropriate t_2-t_1 value to produce the maximum HIC value. The computed HIC values were then transferred into SPSS® for further analysis.

Dependent and Independent Variables

For the static testing, the independent variable was type of helmet liner (TPU, VN, and EEP). The dependent variable was energy percent reduction between the TPU liner and commercial liners (VN and EPP). This analysis was used to address the first research question. The energy percent reduction was computed using Equation 7:

$$Energy\ Percent\ Reduction = \frac{Energy\ Existing\ liner - Energy\ TPU\ liner}{Energy\ Existing\ liner} \tag{7}$$

where:

Energy Existing Liner = the energy absorption of the VN or the EPP liner

Energy TPU Liner = the energy absorption of the TPU liner

For the dynamic testing, the independent variables were impact location (front, side, rear, front boss, and rear boss) and helmet liner type (VN, EPP, and 3D printed TPU). The dependent variables were peak linear acceleration, in units of gravitational acceleration (g) and risk of head injury (HIC) as a compound variable obtained from linear acceleration and time using Equation 2.

Inferential Statistical Analysis

To address the second and third research questions, two way factorial ANOVAs including 5 locations (front, front boss side, rear boss, and rear) x 3 helmet liners (VN, EPP, and 3D printed TPU) with repeated measures on the first and second factor were conducted on measures of peak resultant linear acceleration and risk of head injury (HIC). Descriptive statistics helped describe the mean pair comparisons from the results of the ANOVAs. Although ANOVAs are very robust against violations of normality, a Shapiro-Wilk's Test of Normality on the studentized residuals was conducted to better understand the integrity of the material, possible outliers or any deviations of the material in mitigating acceleration and risk of concussion throughout the dynamic testing.

Results

Static Testing

When examining research question 1, which stated, **what type of helmet liner material (TPU, VN, and EPP) absorbed the most energy during static testing?** The results of the percent energy absorbed by each liner sample over 10 cycles as shown in Figure 17, indicate that all three types of material samples followed a similar pattern. The energy absorbed on the first cycle for each sample was higher than the remaining 9 cycles of the static test. The percent of energy absorption for each sample then stabilized from the 2nd to the 10th cycle of the static testing. When excluding the first load cycle for each material sample, the EPP absorbed the most energy (M=41.8%) from all sample over 9 cycles. The TPU absorbed the second most energy (M=38.6%) and VN absorbed the least amount of energy (M=15.8%).

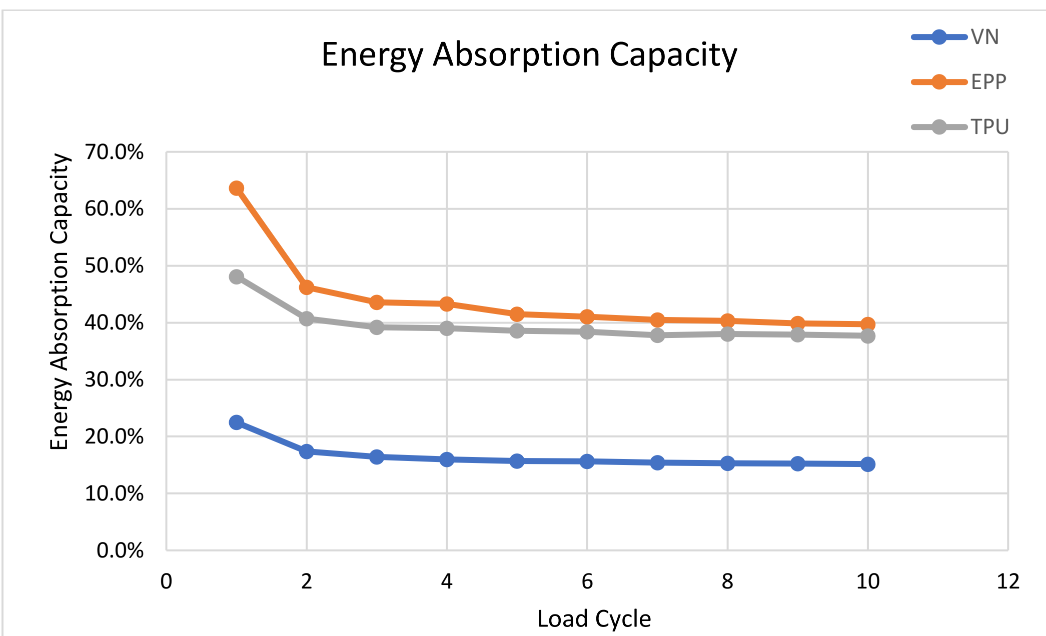

Figure 18. Percent energy absorbed by samples of TPU, VN, and EPP hockey helmet liners during static testing with the Chatillon® TCD1100 force tester.

Dynamic Testing

Peak resultant linear acceleration. When examining research question 2, which stated, **was there a significant difference between TPU, VN, and EPP liners across helmet impact locations on measures of peak linear acceleration during dynamic testing?** The results of the data before examining the two-way repeated measures ANOVA indicated that there were not significant outliers as all the studentized residuals values were less than +/- 3. The Shapiro-Wilk's Test of Normality on the studentized residuals, however, indicated that the data was normally distributed for all, but two (VN liner at the rear location p=.042, and EPP at the rear location p=.039) of the 15 conditions, which resulted from the product of three helmet types and five impact locations. Since 13 out of the 15 conditions met this assumption, and ANOVAs are robust, the researcher considered this violation minor and proceeded with the factorial ANOVA. Mauchly's Test of Sphericity indicated a violation of the assumption of sphericity for the two-way interaction, $X^2(35)=112.833, p<.001$. Since the assumption of sphericity was violated, the Greenhouse-Geisser correction was used when checking for significance. The results of the two-way repeated measures ANOVA indicated a statistically significant interaction effect with a large effect size, $F(2.69, 45.81)=31.14$, $p<.001$, $\eta^2=.647$ between liner materials (TPU, VN, and EPP) and impact locations (front, front boss, side, rear boss, and rear) on measures of peak resultant linear acceleration experienced by the surrogate headform. A graphical representation of the interaction effect can be seen in Figure 18.

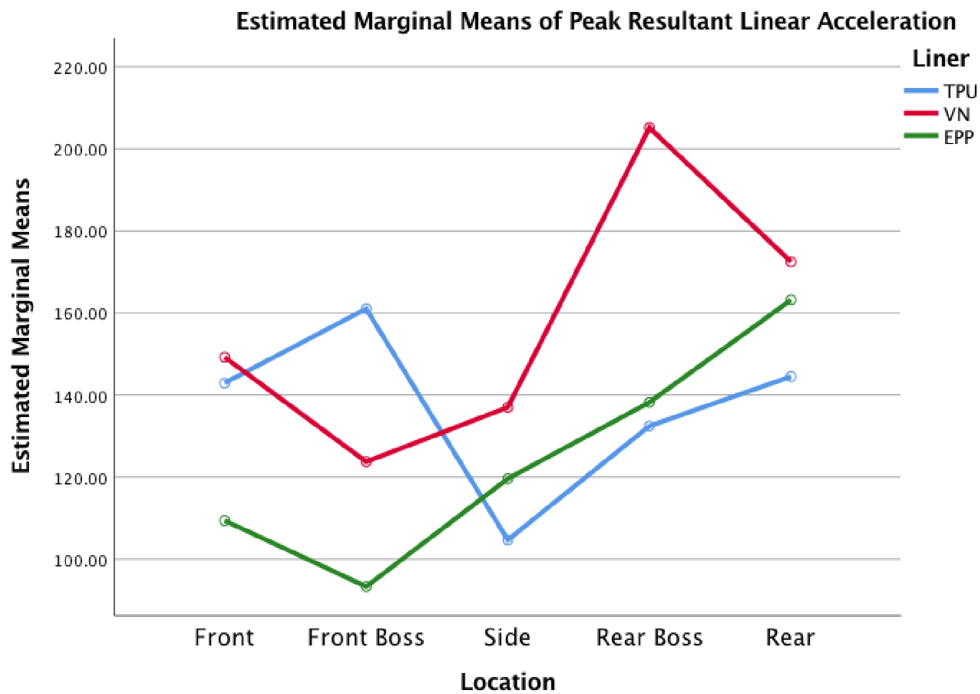

Figure 19. Graphical representation of the interaction effect of liner type and impact location on measures of peak resultant linear acceleration.

The researcher conducted a series of one-way ANOVAs with planned comparisons to examine the simple main effects across liner type for each location separately. Similarly, the researcher conducted one-way ANOVAs to examine the simple main effects across locations for each liner type separately to help explain the interaction in relation to peak linear acceleration measures.

Simple main effects across liner types for each location separately on measures of peak acceleration. At the front location, there was a statistically significant difference with a large effect size between liner types, $F(1.503, 25.558)=34.703$, $p<0.05$, $\eta^2=.750$ on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that the EPP liner (M=109.4, SD=53.3) performed significantly better than both the TPU (M=142.9, SD=59.6) and VN

(M=149.2, SD=75.9) liners (p<.001 for both) and that there was no significant difference between the TPU and VN liners (p=.733).

At the front boss location, there was a statistically significant difference with a large effect size between liner types, $F(2, 34)=123.5, p<0.05, \eta^2=.879$ on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that all three liners were significantly different (p<.001 in all cases) from each other, with the EPP liner (M=93.3, SD=45.9) performing the best, VN (M=123.7, SD=58.4) second, and the TPU liner (M=161.0, SD=59.5) performing the worst.

At the side location, there was a statistically significant difference with a large effect size between liner types,

$F(1.146, 19.479)=12, p<0.05, \eta^2=.413$ on measures of peak linear acceleration. The Bonferroni post hoc analysis indicated that the performance of all three liners was significantly different (p<.05 in all cases) with the TPU liner (M=104.7, SD=43.5) performing best at this location, EPP (M=119.6, SD=58.6) second, and the VN (M=137.0, SD=80.2) performing the worst.

At the rear boss location, there was a statistically significant difference with a large effect size between liner types, $F(2, 34)= 25.4, p<0.05, \eta^2=.599$ on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that there was a significant difference between the VN liner (M=205.2, SD=107.3) and both the TPU (M=144.5, SD=67.4) and EPP (M=138.2, SD=92.0) liners (p <.001 for both) with the VN liner performing significantly worse. There was no significant difference (p>.05) between the TPU and EPP liners at this location.

At the rear location, there was not a statistically significant difference, $F(1.156, 19.652)=4.034, p>0.05$ on measures of peak linear acceleration among the TPU (M=144.5, SD=67.4), VN (M=172.5, SD=119.4) and EPP (M=163.2, SD=108.6) liners.

Simple main effects across location for each liner type separately on measures of peak acceleration. There was a statistically significant difference with a large effect size between impact locations for the TPU lined helmet, F (2.692, 45.756) =43.703, p<0.01, η^2=.720 on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that the order of impact locations from least to most peak resultant linear acceleration is as follows: side (M=104.66, SD=43.49), rear boss (M=132.46, SD=57.21), front (M=142.92, SD=59.58), rear (M=144.50, SD=67.36), and front boss (M=161.02, SD=59.52). The differences between impact locations were all statistically significant (p <.05) with the exception of the pairings of front and rear, and front boss and rear. A graphical representation of the estimated marginal means can be seen in Figure 19.

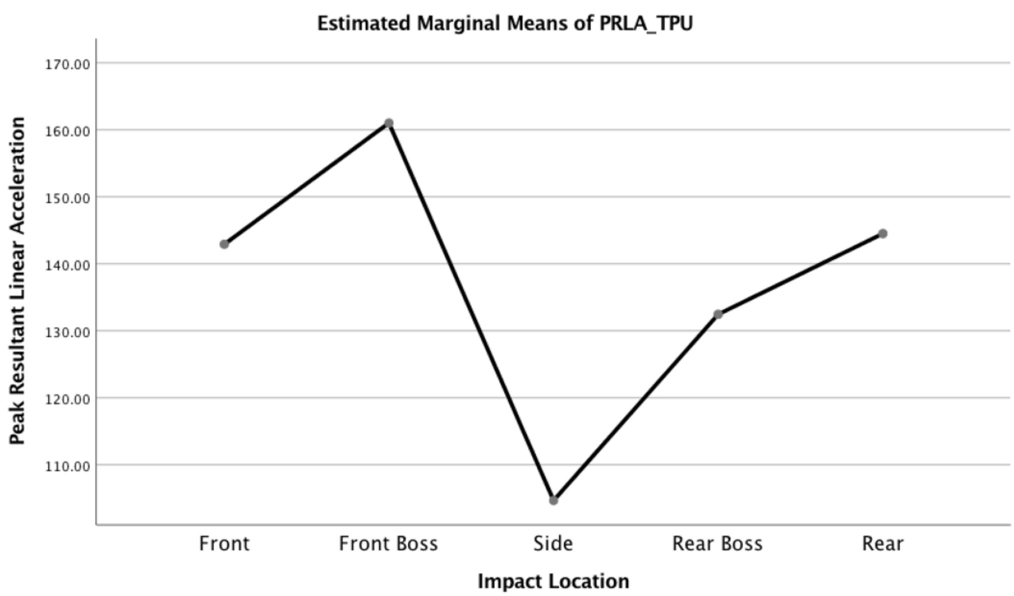

Figure 20. Estimated marginal means of PRLA for the TPU lined helmet.

There was a statistically significant difference with a large effect size between impact locations for the VN lined helmet, F(1.757, 29.866)=21.935, p<0.01, η^2=.563 on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that the order of impact locations from least to most peak resultant linear acceleration is as follows: front boss (M=123.73, SD=58.55), side (M =137.02, SD=80.17), front (M=149.19, SD=75.89), rear (M=172.48, SD=119.43), and rear boss (M=205.15, SD=107.31). The differences between impact locations were all statistically significant (*p*<.05) with the exception of the pairings of front and rear, front boss and side, and rear boss and rear. A graphical representation of the estimated marginal means can be seen in Figure 20.

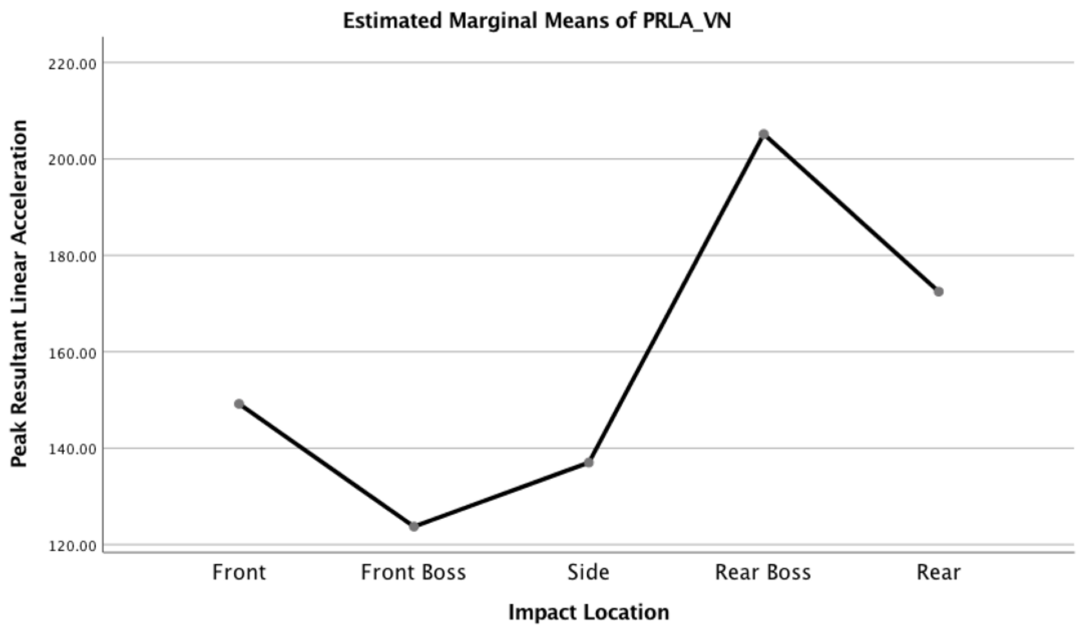

Figure 21. Estimated marginal means of PRLA for the VN lined helmet.

There was a statistically significant difference with a large effect size between impact locations for the EPP lined helmet, $F(1.230, 20.913)=15.253$, $p<0.01$, $\eta^2=.473$ on measures of peak linear acceleration. The Bonferroni post hoc analysis revealed that the order of impact locations from least to most peak resultant linear acceleration is as follows: front boss (M=93.31, SD=53.35), front (M=109.41, SD=53.35), side (M=119.59, SD=58.60), rear boss (M=138.23, SD=91.99), and rear (M=163.18, SD=108.62). The differences between impact locations were all statistically significant (p<.05) with the exception of the pairings of the front and side, front and rear boss, and side and rear boss. A graphical representation of the estimated marginal means can be seen in Figure 21.

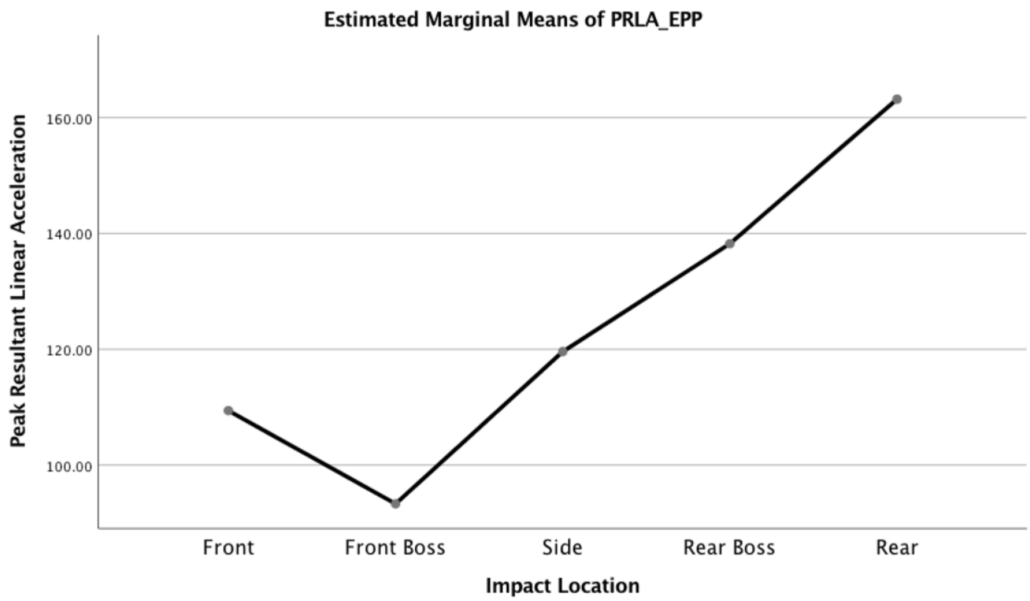

Figure 22. Estimated marginal means of PRLA for the EPP lined helmet.

Head Injury Criteria. When examining research question 3, which stated, **was there a significant difference between TPU, VN, and EPP liners across helmet impact locations on measures of risk of head injury (HIC) during dynamic testing?** The results of the data before examining the two-way repeated measures ANOVA indicated that there were not significant outliers as all the studentized residuals values were less than +/- 3. The Shapiro-Wilk's Test of Normality on the studentized residuals, however, indicated that the data was normally distributed for all but three (VN side location p=.049, VN rear location p=.042, and EPP rear boss location p=.038) of the 15 conditions, which resulted from the product of three helmet types and five impact locations. Since 12 out of 15 conditions met the assumption of normality, and ANOVAs are robust, the researcher considered these violations minor and proceeded with the repeated factorial ANOVA. Mauchly's Test of Sphericity indicated that the assumption of sphericity had been violated for the two-way interaction, $X^2(35)=142.107$, $p<.001$. Since the assumption of sphericity was violated, the Greenhouse-Geisser correction was interpreted for the remainder of the test.

The results of the two-way repeated measures ANOVA, indicated a statistically significant interaction effect with a large effect size, $F(2.156, 36.654)19.846$, $p<.001$, $\eta^2=.539$ between liner type (TPU, VN, and EPP) and impact location, (front, front boss, side, rear boss, and rear) on measures of HIC experienced by the surrogate headform. A graphical representation of the interaction effect can be seen in Figure 22.

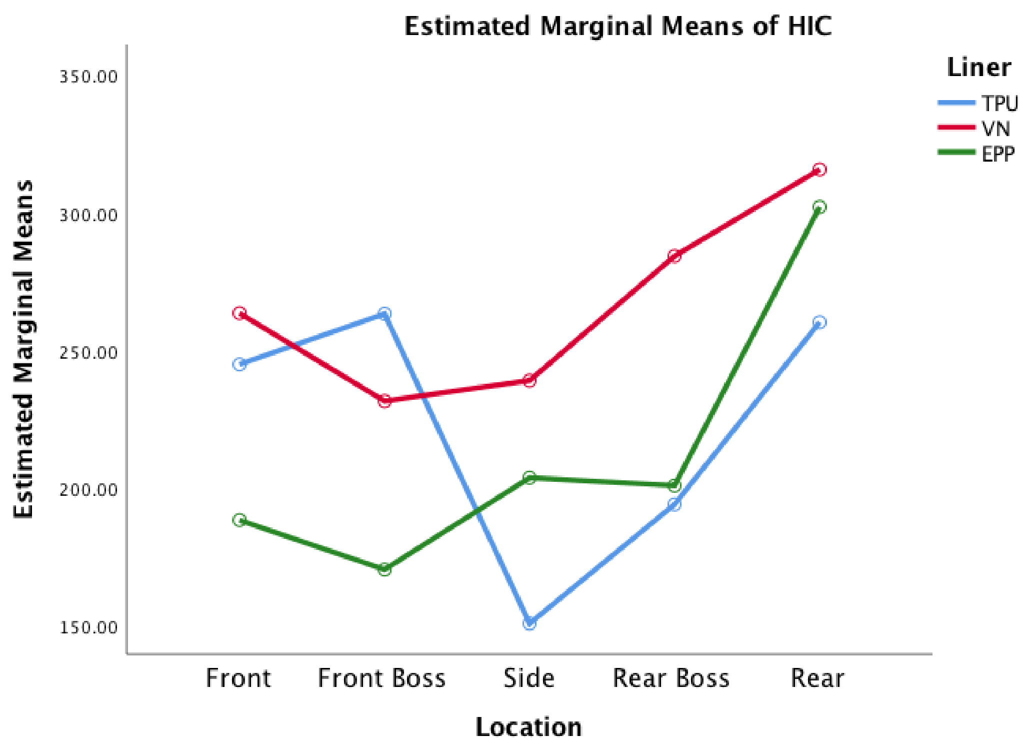

Figure 23. Graphical representation of the interaction effect of liner type and impact location on measures of HIC.

The researcher conducted a series of one-way ANOVAs with planned comparisons to examine the simple main effects across liner type for each location separately. Similarly, the researcher conducted one-way ANOVAs to examine the simple main effects across locations for each liner type separately to help explain the interaction in relation to HIC measures.

Simple main effects across liner types for each location separately on measures of HIC. At the front location, there was a statistically significant difference with a large effect size between liner types, $F(1.110, 18.865)=28.835$, $p<0.05$, $\eta^2=.629$ on HIC measures. The Bonferroni post hoc analysis revealed that the EPP liner (M=188.5, SD=128.9) performed significantly better than the TPU (M=245.0, SD=154.0) and VN (M=263.6, SD=186.3) liners. There was no statistically significant difference between the TPU and VN liners at the front location.

At the front boss location, there was a statistically significant difference between liner types, $F(1.185, 20.150)=61.614$, $p<0.01$, $\eta^2=.784$ on HIC measures. The Bonferroni post hoc analysis indicated that the VN liner (M=231.7, SD=162.7) performed significantly better than the TPU liner (M=263.4, SD=163.4), while the EPP liner (M=170.6, SD=122.4) performed significantly better than both the VN and TPU liners.

At the side location, there was a statistically significant difference with a large effect size between liner types, $F(1.203)=15.023$, $p<0.05$, $\eta^2=.469$ on HIC measures. The Bonferroni post hoc analysis revealed that the EPP liner (M=203.9, SD=138.7) performed significantly better than the VN liner (M=239.2, SD=188.4), while the TPU liner (M=151.0, SD=98.9) performed significantly better than both the EPP and VN liners.

At the rear boss location, there was a statistically significant difference with a large effect size between liner types, $F(2, 34)=17.490$, $p<0.05$, $\eta^2=.507$ on HIC measures. The Bonferroni post hoc analysis revealed that the difference in performance of the TPU (M=194.2, SD=119.4) and EPP (M=201.1, SD=163.2) liners was not significantly different but they both performed significantly better than the VN liner (M=284.3, SD=197.1).

At the rear location, there was a statistically significant difference with a medium effect size between liner types, $F(1.134, 19.273)=4.948$, $p<0.05$, $\eta^2=.225$ on HIC measures; however, none of the pairwise comparisons from the Bonferroni post hoc analysis were statistically significant ($p>.05$).

Simple main effects across location for each liner type separately on measures of HIC. There was a statistically significant difference with a large effect size between impact locations for the TPU lined helmet, $F(1.184, 20.134)=35.184$, $p<0.01$, $\eta^2=.674$ on HIC measures. The Bonferroni post hoc analysis revealed that the order of impact locations from least to most HIC is as follows: side (M=150.99, SD=98.94), rear boss (M=194.23, SD=119.36), front (M=245.03, SD=153.98), rear (M=260.33, SD=175.08), and front boss (M=263.39, SD=163.41). The differences between impact locations were all statistically significant ($p<.05$) with the exception of the pairings of front and rear and front boss and rear. A graphical representation of the estimated marginal means can be seen in Figure 23.

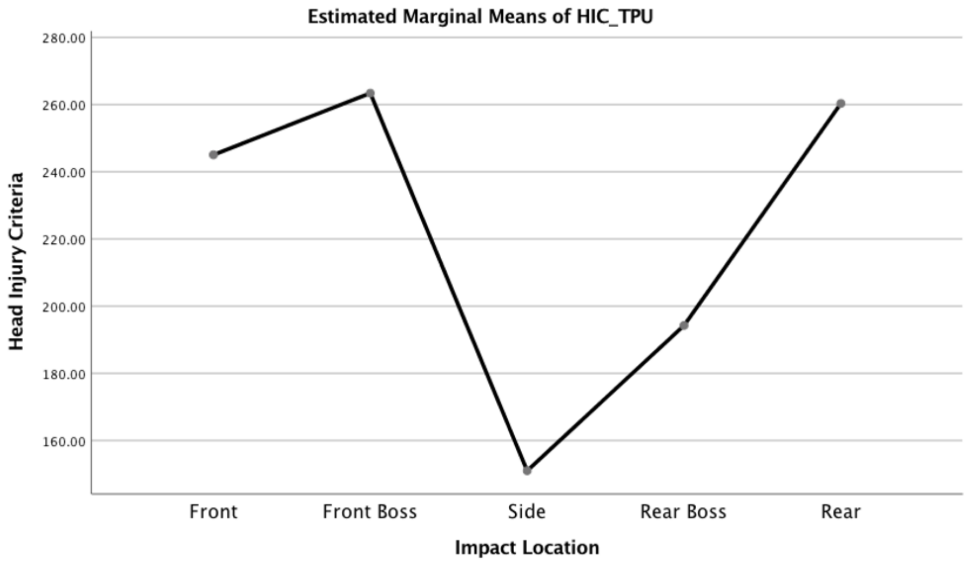

*Figure 24.*Estimated marginal means of HIC for the TPU lined helmet.

There was a statistically significant difference with a large effect size between impact locations for the VN lined helmet, $F(1.503, 25.557)=8.458$, $p<0.05$, $\eta^2=.332$ on HIC measures. The Bonferroni post hoc analysis revealed that the order of impact locations from least to most HIC is as follows: front boss (M=231.69, SD=162.71), side (M=239.16, SD=188.35), front (M=263.59, SD=186.30), rear boss (M=284.32, SD=197.14), and rear (M=315.73, SD=274.23). The differences between impact locations that were statistically significant (p <.05) included the pairings of front and front boss, front and side, front boss and rear boss, side and rear boss, and side and rear. A graphical representation of the estimated marginal means can be seen in Figure 24.

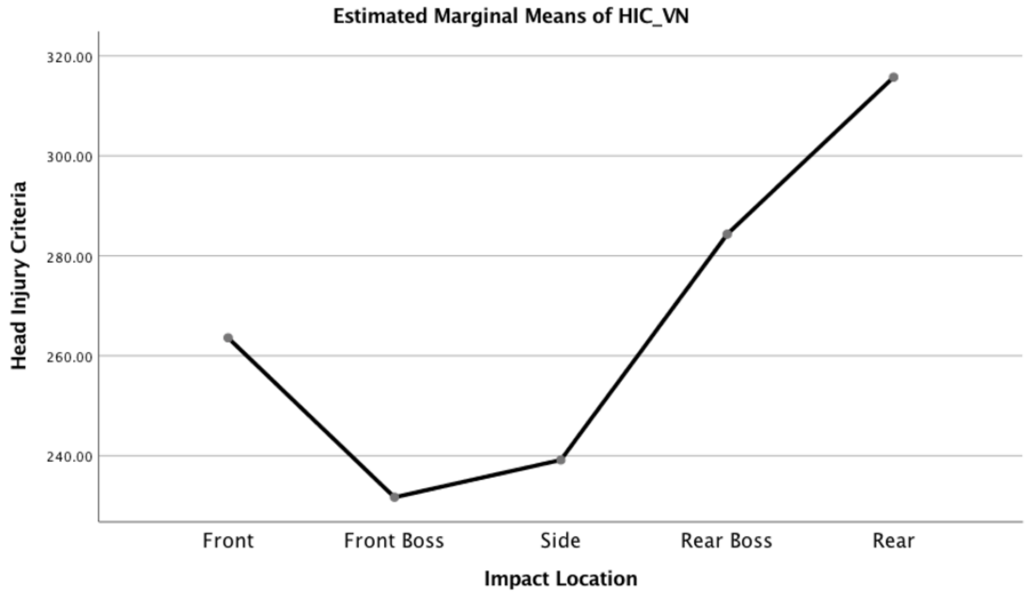

Figure 25. Estimated marginal means of HIC for the VN lined helmet.

There was a statistically significant difference with a large effect size between impact locations for the EPP lined helmet, $F(1.174, 19.965)=15.904$, $p<0.01$, $\eta^2=.483$ on HIC measures. The Bonferroni post hoc analysis, revealed that the order of impact locations from least to most HIC is as follows: front boss (M=170.57, SD=122.43), front (M=188.50, SD=128.89), rear boss (M=201.07, SD=163.17), side (M=203.88, SD=138.74), and rear (M=302.13, SD=248.48). The differences between impact locations were all statistically significant ($p<.05$) with the exception of the pairings of front and side, front and rear boss, front boss and rear boss, and side and rear boss. A graphical representation of the estimated marginal means can be seen in Figure 25.

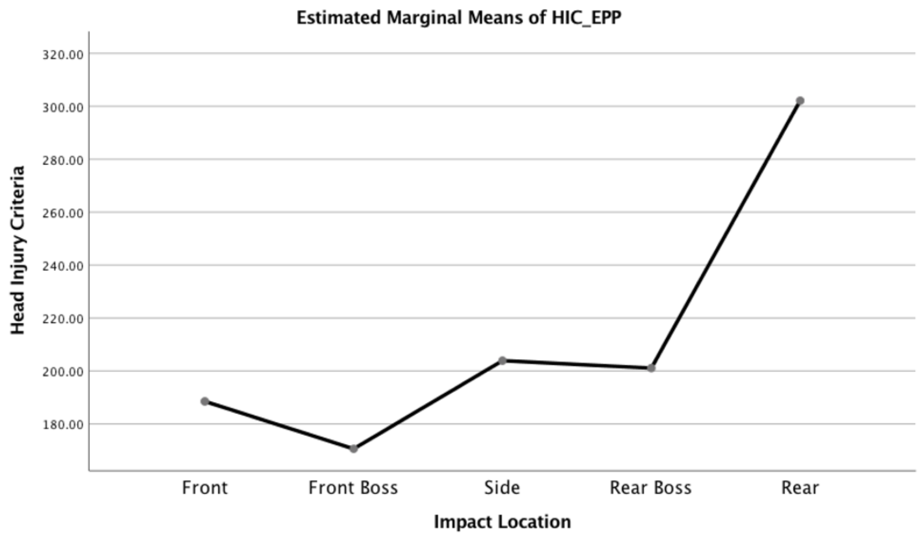

Figure 26. Estimated marginal means of HIC for the EPP lined helmet.

Discussion

Helmets currently represent the best form of head protection against traumatic brain injuries for hockey players (Biasca et al., 2002). Hockey helmet designs vary greatly depending on their lining material properties and outer shell. It is proposed that the inner liner protects the head and brain against rapid acceleration/deceleration causing traumatic brain injuries. The outer shell protects the head and brain against focal injuries such as skull fractures (Graham et al., 2014).

The most commonly commercial lining materials are made of either VN or EPP polymers (Rousseau et al., 2009a). The VN liners are created by cutting out specific shapes from large sheets of material, whereas EPP liners are created by pouring the material into a mould to create the desired shape (Rousseau et al., 2009a). Unfortunately, concussions continue to occur due to the inability of existing helmets to mitigate linear and rotational acceleration across different impact locations on the helmet (Clark et al., 2016).

In the current study, an innovative helmet lining was investigated that was made out of TPU inserts. The inserts were placed inside the outer shell of an existing helmet (Figure 13) identical to those having the EPP and VN lining materials. The TPU inserts were then tested against the commercial EPP and VN linings, for the purpose of comparison. The TPU material was chosen because it possesses the capability to morph from a soft to a hard polymer when heated and then cooled (Huntsman, 2010). The benefit of this particular material lies in its capability to withstand multiple compression forces without losing its structural integrity (Huntsman, 2010). During repeated compression tests, for example, the TPU inserts would be compressed by forces over 1000 N and would return to their initial shapes almost instantly upon being released from the compression force.

The importance of testing the performance of helmet materials is to learn about the behaviour of the material under static and dynamic loadings. Static testing provides information on the elastic properties of the material by measuring: (a) the loading forces required to deform the material, and (b) upon unloading, how much the material would restore its original shape (Baumgart, 2000). Consequently, static testing determines the stiffness of the material but more importantly, the material's capability to absorb energy. Dynamic testing, on the other hand, provides information on the ability of the material to absorb energy by deforming and restoring its original shape within a short period of time based on the velocity of the impact (Jamil, Guan, & Cantwell, 2017). Therefore, in the current study, static and dynamic testing were conducted to better understand the capability of the TPU material in minimizing linear impact acceleration and risk of brain injury when compared to existing commercial lining materials.

Static Testing

The ability to attenuate impact forces is a key characteristic of ice hockey helmets. One way this ability can be quantified is by assessing the stiffness of the helmet material, which describes the force needed to cause a certain amount of deformation of the helmet material (Baumgart, 2010). As the materials of concern are polymers, the stiffness is nonlinear, however, static testing can be used to gain information about the liner materials' capability to absorb impact energy as an avenue to reduce the magnitude of the possible impacts experienced by hockey players during collisions.

Research Question 1 asked: **What type of helmet liner material (TPU, VN, and EPP) absorbed the most energy during static testing?** When the TPU liner material was compared to the two most commonly used ice hockey helmet liner materials (VN and EPP), on percent of energy absorbed during static testing, the result revealed that the TPU sample was able to absorb

38.6% of the loading energy per loading and unloading cycle (Figure 17). The EPP and VN lining materials absorbed 41.8% and 15.8%, respectively. Therefore, the TPU absorbed 7.7% less per cycle when compared with the EPP but absorbed 1.4 times more than the VN. These outcomes can be explained based on the concept of functionally graded materials in which the microstructural details of the material are spatially varied changing the density of the material (Hirai, 1996). Furthermore, the shape of the liners seemed to have an effect on the energy absorption during compression. The shape of the VN liners, for example, were created by cutting out large sheets of material; the shape of the EPP liners were created by pouring the material into a mould to create the desired shape (Rousseau et al., 2009a). Whereas the particular shapes of the TPU liners were created, (a) using thermoplastic material via 3D printing technology; and (b) keeping in mind the needs to absorb linear as well as angular impacts. As Figure 13 shows, the TPU inserts were placed in some locations only. This is in contrast to the placement of EPP and VN liners that cover the entire interior of the helmet. Therefore, the shape and placement of the linings may have an effect on the material's ability to absorb energy under static compression.

The implication of these findings is that the current TPU liner falls well within the range of performance of the commercially available liners during static compression testing. Better performance may be achieved by careful considerations of the sizes and placements of the TPU inserts. The outcome of the static testing provided evidence to proceed with the dynamic testing of the TPU inserts as a potential hockey helmet liner material to mitigate impact forces responsible for causing traumatic brain injuries.

Dynamic Testing

Static testing alone does not completely determine a helmet's ability to mitigate impact forces. The most important characteristic of a hockey helmet is its ability to mitigate impact forces during dynamic loading while being worn by a player. Currently, in helmet testing, this is assessed by dropping a helmeted headform onto a rigid surface or striking it with a rigid impacting rod, and analyzing the linear acceleration experienced by the headform (Clark et al., 2016).

Peak linear acceleration. Linear acceleration is used because of its biomechanical relationship to concussions (Rowson & Duma, 2013). When helmets undergo certification, they are assessed on their ability to mitigate peak linear acceleration from a drop onto a rigid surface (Clark et al., 2016). The idea is to replicate the real-world circumstances where an athlete could be injured by a blow to the head. Since the certification process is based on the mitigation of peak resultant linear acceleration, the current study compared the ability to mitigate this acceleration between the TPU liner and the commercially available VN and EPP.

Research Question 2 asked: **Is there a significant difference between TPU, VN and EPP liners across helmet impact locations on measures of peak linear acceleration during dynamic testing?** A significant interaction effect was observed between helmet liner material and impact location with a large effect size of η^2=.647. This interaction means that at different impact locations, a liner material resulted in different peak resultant linear accelerations revealing a strong relationship between impact location and liner type. The TPU liner performed the best at the side location when compared to the VN and EPP commercial liners (Figure 18). This outcome is meaningful in terms of brain injury prevention in the sport of ice hockey because the side location has been linked to a greater risk of sustaining a mTBI (Elkin, Gabler, Panzer, & Siegmund, 2019; Liao, Lynall, & Mihalik, 2016).

The occurrence of traumatic brain injuries at different impact locations of the head had not only been examined in the sport of ice hockey but also in other sports such as football. Liao et al. (2016) compared the percentage of impacts to the front, sides, and top of the head sustained by football players who suffered a concussion versus those that did not. The researchers found that the group that suffered concussions were significantly more likely to be hit on the sides or top of their head, than the group that was not concussed (Liao et al., 2016). In another study, Elkin et al. (2019) examined the differences in brain tissue strain across locations from constant impact energies on football helmets. The researchers found that the difference in impact location accounted for 33 - 37% of the variance in brain strain for the whole cerebellum (Elkin et al., 2019). These outcomes are important to the findings of the current study because the TPU liner performed the best at the side location, which may offer an avenue to mitigate the risk of sustaining a mTBI by improving helmet liner performance for different helmet brands (Elkin et al., 2019; Liao et al., 2016).

Furthermore, in the current study, significant differences between the liners were found at all locations. The EPP liner outperformed the VN at all five locations (Figure 18). It means that of the two commercially available liner materials, EPP does a significantly better job of mitigating peak resultant linear acceleration. When compared to the TPU liner, the EPP liner performed significantly better at the front and front boss locations. Due to the shape of the shell and the size of the TPU inserts, it was difficult to attach inserts to this area of the shell. Less inserts were used in these locations because of this difficulty, which would explain why the TPU was outperformed by the EPP. A lower number of inserts in the front and front boss locations would mean that there is less material to absorb the impact energy. At the side location, the TPU significantly outperformed the EPP liner. This was the location on the shell of the helmet where it was easiest to attach the TPU liners in a manner that would fully cover the impact zone. Lastly at the rear boss and rear locations, there was no statistically significant difference between the TPU and EPP liners, however, the mean peak resultant linear acceleration at both locations was lower for the helmet with the TPU liner (Figures 19 and 21). This means that the TPU liner performed at least as well as the EPP liner at both locations.

Rousseau, Post, and Hoshizaki (2009b) had similar findings when comparing VN and EPP liners between helmets with the same shell. They determined that the overall peak linear acceleration for EPP liners was less than that of the VN liners, but more importantly, the EPP liner outperformed the VN on measures of peak linear acceleration for all three different helmet models. The results of the current study confirmed that the helmet with the EPP liner significantly outperformed the helmet with the VN liner on measures of peak resultant linear acceleration. It should be noted that such findings have been repeated across helmet brands, by independent research teams, using different equipment.

A plausible explanation of EPP having better performance than VN is as follows. As stated by Hirai (1996), less dense foam liners deform and absorb energy from lower speed impacts and as the foam liner gets progressively denser, it will be more effective at higher speed impacts. Generally speaking, the EPP has a density of 65 kg/m^3 while the VN has 95 kg/m^3 (Spyrou & Hoshizaki, 2001). The maximum impact speed in the current study was 5.13 m/s (Table 3), which meant that all the impact speeds being tested in the current study were considered lower speeds.

Head injury criteria. The HIC measure was also examined in the current study. It was included in the present study because the mechanism of injury depends on the duration of the loading, in addition to the magnitude of the load (Schmitt, Niederer, Mser, & Walz, 2010). Based on Schmitt et al. (2010), static loads are classified as those with a time duration of greater than 200 ms. Static loads tend to cause skull fracture. This type of loading is rare (Schmitt et al., 2010). On the other hand, dynamic loads can cause both focal brain injuries and diffuse brain injuries. Therefore, peak resultant linear acceleration alone does not present the entire picture. In terms of criteria that take into consideration both the magnitude and duration of the loading, the Wayne State University Cerebral Concussion Tolerance Curve (or simply the WSTC) was an important development. The WSTC was proposed in the 1950s to the 1960s (Gurdjian, Lissner, Latimer, Haddad & Webster, 1953; Gurdjian, Roberts & Thomas, 1966; Gurdjian, Webster & Lissner, 1955). Since then, the understanding of the mechanism of brain injuries deepened, due to the advancement of technology and the need for protective devices to mitigate the severity of brain injuries. The GSI and HIC were developed from the WSTC. Both are suitable for use as predictors for TBI as they appear to be less predictive in environments where the impacts involve compliant surfaces (Hoshizaki et al., 2017). Compared with the GSI, the HIC allows for the freedom to choose the appropriate impact duration $t_2 - t_1$ (see Equation 2). There is also a corelation between an HIC score and the probability of severe head injury (Zhang et al., 2004).

Since HIC is based on peak resultant linear acceleration, drastic differences between the results from the two variables were not expected in the current study. It is import to emphasize, however, that HIC not only considers the magnitude of the acceleration, but also the duration of time that the head undergoes acceleration (Rousseau et al., 2009b). The measure of HIC was included in the current study to determine if the TPU liner would perform any differently than the VN and EPP liners in terms of mitigating the peak resultant linear acceleration or spreading that acceleration over a longer period of time, resulting in a less dangerous impact.

Research Question 3 asked **Is there a significant difference between TPU, VN and EPP liners across helmet impact locations on measures of risk of head injury (HIC) during dynamic testing?** It was found that the interaction effect between location and liner type was statistically significant with a medium effect size η^2=.539. This interaction means that the effectiveness of the type of liner material to mitigate risk of concussion depends on the location of the impact, revealing a strong relationship between type of liner and impact location in terms of protection against brain injuries.

The TPU liner was the most effective liner at the side, rear boss, and rear locations as shown in Figure 22. This is an important finding because research has shown that impacts to the side of the head are the most likely to cause a concussion, followed by impacts to the back of the head (Delaney et al., 2014; Liao et al., 2016). A significant increase in the mitigation of the impact force by using the TPU liner at these key locations could lead to an overall decrease in risk of head injury for hockey players. The EPP liner was the most effective at the front and front boss locations and the VN liner was not the most effective liner at any of the five locations tested (Figure 22). These results also indicate that the TPU lined helmet would most likely pass certification standards at all locations except for front boss, since it outperformed one or both commercially available liners that have already passed these tests, at the other four locations.

The HIC results were also found to be consistent with those for peak resultant linear acceleration. Table 5 summarizes the key findings from Peak Resultant Linear Acceleration (PRLA) and HIC in the current study.

Table 5.

Key findings for PRLA and HIC.

Measure	Interaction between Liner Materials and Impact Locations	Most Effective Liner Material at each Impact Location				
		FRONT	FRONT BOSS	SIDE	REAR BOSS	REAR
PRLA	significant	EPP	EPP	TPU	TPU	TPU
HIC	significant	EPP	EPP	TPU	TPU	TPU

It is worth noting that the TPU liners outperformed the EPP at the side location. The study by Liao et al. (2016) found that football players who suffered concussions were significantly more likely to be hit on the sides or top of their heads. Wilcox et al. (2014) also found that male collegiate ice hockey players experienced greater 95th percentile peak linear acceleration from impacts to the back of the head than the front or the side. They also experienced greater 95th percentile peak rotational acceleration from impacts to the side of the head when compared to the front. Since both peak linear and peak angular accelerations are strong predictors of concussion incidence, these locations are important ones to examine. Elkin et al., (2019) examined the differences in brain tissue strain across locations from constant impact energies on football helmets. It was found that the difference in impact location accounted for 33-37% of the variance in brain strain for the whole cerebellum (Elkin et al., 2019). These findings are important to the current study because the TPU liner performed the best at the side location, which was significantly more likely to be hit (Liao et al., 2016).

Therefore, it becomes essential for a liner to have good performances at the side locations. The TPU material may offer an avenue to mitigate the risk of sustaining a mTBI by improving helmet liner performance for different helmet brands (Elkin et al., 2019; Liao et al., 2016).

The current results also mean that no liner did a significantly better job than the others in spreading the impact forces over a longer period of time. The results of the HIC calculations are reflective of the results from the peak resultant linear acceleration from which it is derived. These results indicate that the optimal ice hockey helmet in terms of mitigation of peak resultant linear acceleration and HIC would contain a liner made from multiple materials, choosing the best one for each individual impact location. Implications for helmet manufacturers would include an increase in manufacturing complexity and likely cost, but the potential benefit of decreasing the risk of head injury for hockey players would be a very important step forward in player safety.

In summary, when the interaction effect between location and liner type was examined, it was found that the lining material that resulted in the lowest HIC was dependent on the location of the impact. From the practical perspective, this turned out to be the most significant finding of the present study due to the implications it could have for helmet manufacturers, in using liners made of multiple materials. According to this finding, the optimal helmet would contain a liner made of EPP for the front and front boss locations, and TPU for the side, rear boss and rear locations.

Conclusion

The purpose of this study was to test an innovative, 3D printed TPU hockey helmet liner and compare the results to commercially available VN and EPP lined helmets. The liner materials were tested on their ability to absorb energy during static loading to understand the properties of the materials. They were subsequently tested and compared under dynamic loading conditions to evaluate their ability to mitigate impact forces during simulated head impacts using a surrogate headform. The study aimed to determine the ability of the 3D printed TPU hockey helmet liner to reduce the risk of head injury to hockey players when compared to commercially available helmets.

The most significant finding of the current study is that the effectiveness of hockey helmet liner depends on the location of the impact. This is an important finding because of its practical implication. According to this study, the optimal hockey helmet would contain a liner made of more than one material. This optimal helmet would consist of an EPP liner at the front and front boss locations, and a TPU liner at the side, rear boss, and rear locations. This would require helmet manufacturers to change their methods of production to allow liners of a single helmet to be made of a combination of different materials. According to the results of the current study, this helmet would be optimal in reducing the risk of head injury for ice hockey players and would make the sport safer for players of all ages and skill levels.

Strengths

The primary strength of this study is demonstrating the potential of 3D printed TPU inserts as a viable alternative for ice hockey helmet liner to improve helmet performance in mitigating concussion risk. Static testing shows that the current TPU liner falls well within the range of performance of the commercially available liners during static compression testing.

Better performance may be achieved by careful considerations of the sizes and placements of the TPU inserts. Dynamic testing indicates that the TPU liner is effective at the side, rear boss, and rear locations in reducing the peak linear acceleration and HIC. Combined with EPP at the front and front boss locations, the helmet may give rise to the optimal performance.

Production of TPU liners aside, 3D printing offers researchers the ability to quickly modify and create new designs, allowing for many different prototypes to be tested within a reasonable amount of time. Another benefit is the cost of the material. A 1-kg spool of TPU 3D printer filament currently sells for less than $40, such that each individual insert only cost a few cents to print. Using a material that is also economically viable is beneficial at all stages of research and development. The 3D printing also allows for the creation of variable density liners that are made from a single material (such as TPU) well suited to withstand many impacts while still returning to its original shape.

Limitations

One limitation of this study was that the rotational accelerations were not measured or analyzed. Recent concussion research often includes rotational acceleration as a variable since it is now generally accepted that impacts to the head of an ice hockey player will result in a combination of linear and angular accelerations (Post, Hoshizaki, Gilchrist, & Cusimano, 2017). Equipment and instrumentation limitations, however, made it not feasible to measure rotational accelerations in the current study. For rotational accelerations to be added, three extra accelerometers would need to be added, away from the centre of mass to be able to determine the rotational accelerations about all three axes.

The cylindrical shape of the inserts used in this study also made it difficult to attach them to the helmet shell effectively. The flat tops and bottoms of the inserts do not line up well with the curved shell. In order to better secure the liner to the shell, the design of the inserts should have a curved top or bottom matching the curve of the shell where needed. Having inserts with varying diameters would also allow for a greater portion of the inside of the shell to be covered, such that the impact forces would be distributed over more inserts.

The 3D printer used in this study is Creality CR-10s. It does not have the capability to use two materials to print the same object. Ideally, one material (say, TPU) is for the actual object, and a second is a water-soluble material, called "support material" that is necessary to retain the shape of the object during the printing, and is then dissolved in water once the printing is complete. Since the support material for the inserts used in this study could not be dissolved, it was left as part of the insert. This resulted in the TPU inserts being stronger than intended.

Future Directions

The current TPU insert design was chosen after preliminary static testing narrowed nearly 20 different designs down to the most promising one. Only then was this promising design put through the more comprehensive static and dynamic tests that were reported in this thesis. With the valuable data obtained from static testing by the Chatillon® TCD1100 force tester, and the peak linear acceleration/HIC data obtained from dynamic testing, the current design can be improved to perform better under the conditions where it was outperformed by the EPP and VN liners. Soe and colleagues (2015) examined the effect that the density of the cellular structure had on pulse duration and its ability to mitigate peak translational acceleration. The 3D printed thermoplastic elastomer structures were compared at various densities ranging from solid to 15% filled. They found that the density of the cellular structure had a significant effect on the kinematics of the head during impact. The implications of these results are that despite using the same material, a change in the density is enough to change the kinematic outcomes from an impact to the head. The design of the TPU liner used in this study will serve as a good foundation for further developements. Future studies should keep in mind that alterations to the design's density, as well as overall shape, could have a drastic effect on its ability to mitigate impact forces and limit both linear and angular accelerations to the head.

Future studies should include the variable of rotational acceleration. This would give the researchers a more complete picture of how well different liner materials are able to mitigate all impact forces and moments. With both linear and rotational accelerations contributing to the risk of head injury, both should be examined when testing new materials. Another recent method of studying helmets is through the use of finite element modelling. The benefit of using finite element modelling is that the strain on the brain tissue can be simulated, not merely the after-impact kinematics of the head (Soe et al., 2015). Understanding how different liner materials affect the brain tissues, rather than making inferences based on how the head, as a whole, reacts to the impact, would allow the researchers to gain deeper insight to helmet performance. This would add another dimension to the understanding of the performance of TPU as an ice hockey helmet liner.

The interaction effect between liner type and impact location is also worthy of further investigation. More brands/designs of helmet shells should be tested to ensure the differences in performance between the liner materials at different locations is due solely to the effect of the liner material, and not a combination of the liner material and any particular shell. Due to the fact that the TPU liner is 3D printed and can be easily modified to a desired density, stiffness, shape, and so on, it could be designed to more closely mimic the physical properties of the EPP liner, and those pieces used at the front and front boss locations where it was found to be effective. Understanding why a particular material performs well under certain conditions will help future studies create more effective liners, made from whichever material or combination of materials deemed the best suited for each individual impact location.

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