# The Effect of Shoulder Pad Design on Head Impact Severity During Shoulder Checks in Ice Hockey

by

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## Ethics Statement

The author, whose name appears on the title page of this work, has obtained, for the research described in this work, either:

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or

b. advance approval of the animal care protocol from the University Animal Care Committee of Simon Fraser University

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# Abstract

Forty-two percent of concussions in ice hockey are caused by hits involving shoulder-to-head contact. The goal of this project was to determine how shoulder pad stiffness affects head impact severity when players delivered checks to an instrumented dummy.

Fifteen participants administered “the hardest shoulder checks they were comfortable delivering” to the head of an instrumented dummy. Trials were conducted with participants wearing two common types of shoulder pads, with and without a 2 cm thick layer of polyurethane foam over the shoulder pad cap.

The study found that a 2 cm thick foam layer overlying the shoulder cap reduced peak linear accelerations to the head by 21.6–27.7%, peak rotational velocities by 10.5-13.8%, while causing no significant increase in shoulder impact velocity. Therefore, integration of foam padding on top of plastic caps warrants further examination as a method for preventing brain injuries in ice hockey.

**Keywords**: Concussion; Ice Hockey; Shoulder Padding; Risk Compensation; Head Injury; Hockey Equipment

# Dedication

This thesis is dedicated to Diana Pricop, whose love and support always motivates me to strive for my best.

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## List of Acronyms

| | |
|---|---|
| CAHA | Canadian Amateur Hockey Association |
| CSA | Canadian Standards Association |
| CTE | Chronic Traumatic Encephalopathy |
| EPP | Expanded Polypropylene |
| GFT | GForce Tracker |
| HITS | Head Impact Telemetry System |
| mTBI | Mild Traumatic Brain Injury |
| NCAA | National Collegiate Athletic Association |
| NHL | National Hockey League |
| PCS | Post-concussion Symptoms |

## Glossary

| | |
|---|---|
| CAHA | The national governing body of amateur ice hockey in Canada from 1914 to 1994. |
| CSA | A non-profit standards organization in Canada that is accredited by the Standards Council of Canada. |
| CTE | A form of encephalopathy found in individuals with a history of repetitive brain trauma. |
| EPP | A closed-cell foam that has strong energy absorption properties. |
| GFT | A commercially sold g-force and rotational velocity monitoring sensor system. |
| HIT | A software system designed to detect concussion type collisions with an array of sensors located in a helmet. |
| mTBI | A traumatically induced brain injury that causes physiological disruption of brain function. |
| NCAA | A non- profit association that regulates the athletic programs of many universities in the United States and Canada. |
| NHL | A professional ice hockey league considered one of the major sport leagues in North America |
| PCS | A complex disorder in which various symptoms can last for an indefinite amount of time following a concussion injury. |

# Chapter 1.

# Introduction

## 1.1. Mild Traumatic Brain Injuries in Sport

Over the last thirty years, there has been a growing recognition of the importance of participation in sport on the improvement of both physical and mental well-being (Taylor, Sallis, & Needle, 1985). Physical inactivity causes a significant economic burden on the Canadian public health system, one that can be drastically reduced by even modest increases in physical activity (Katzmarzyk, Gledhill, & Shephard, 2000). Thus, there has been a growing promotion of sport and physical activity to Canadians over recent years in order to reduce the health care costs attributed to physical inactivity. Unfortunately however, it appears that national sport participation rates have been declining. In 2010, it was estimated that around 7.2 million individuals participate in sport on a regular basis, down from the 8.3 million that was reported in 1998 (Canadian Heritage, 2013). A suspected contributing factor to the decline is the increased exposure and scrutiny that sport related injuries have received. While sports offer numerous health benefits, they have also been shown to be a major cause of injuries to Canadian youth and young adults (Billette & Janz, 2011). In fact, approximately one in four Canadians aged 12 to 19 will suffer a substantial injury each year and two out of three of these injuries are linked to sports. Therefore, management of the injury risk associated with sport is essential in promotion of physical activity to Canadian adolescent and young adults.

There has also been a growing concern regarding the high frequency and long-term consequences that sports-related brain injuries have on young athletes, the most common of which being concussion injuries. Concussion is one of the most common sports injuries incurred by Canadian youth and young adults, and one that is especially common in organized contact sport (Browne, 2006). The word “concussion” is derived from the Latin verb concutere, which means “to shake” (McCrory & Berkovic, 2001), and has historically been used to refer to an injury caused by the brain “shaking” within the cranial walls resulting in observable clinical symptoms. While some controversy remains regarding a commonly accepted definition, the general consensus is that a “concussion is a brain injury and is defined as a complex pathophysiological process affecting the brain, induced by biomechanical forces” (McCrory et al., 2013). Concussion is often defined by clinical symptoms in the absence of detectable abnormalities in brain anatomy as revealed by imaging modalities such as computed tomography (CT) (McCrory et al., 2013), and is indicative of mild (as opposed to severe) traumatic brain injury (although many object to the notion of any brain injury as being “mild”). Common signs and symptoms of a concussion injury include confusion, memory impairment, poor coordination, and slow response times to questions (Aubry et al., 2002). Symptoms are often highly individualistic and athletes may frequently experience a variety of additional symptoms following the concussive event such as headaches, sleep deprivation, nausea, and cognitive difficulties. These symptoms are often referred to as post-concussion syndrome (PCS) and can have a significant impact on how individuals are able to perform daily activities. Perhaps unsurprisingly, these deficits are especially profound in children and youth whose brains have been shown to be more vulnerable to concussion injuries (Kirkwood, Yeates, & Wilson, 2006). While generally the majority of athletes who suffer a concussion are able to recover fully within weeks, a large proportion of concussed athletes experience prolonged symptoms and may even face difficulties throughout the remainder of their lives despite having injury mechanisms that are similar to those who are able to recover in a short time (Comper, Hutchison, Magrys, Mainwaring, & Richards, 2010).

If an athlete is suspected of having incurred a concussion injury, current return-to-play guidelines state that the player is to be removed from the game and is to wait until they are asymptomatic at rest before proceeding to engage in further exercise (McCrory et al., 2013). Unfortunately, return-to-play guidelines are not always used and a significant portion of “mild” concussions that go unreported or undiagnosed still result in long-term complications on cognitive function (Echlin et al., 2012; Kroshus, Garnett, Hawrilenko, Baugh, & Calzo, 2015). Adding to the problem is the pressure athletes face to continue playing through a concussion injury from coaches, teammates, and even themselves. Players often do not want to be removed from the game for fear of putting their team at a disadvantage and thus downplay the extent of their injuries (Kroshus et al., 2015). However, athletes who continue to play through a concussion injury are more likely to experience a second concussion (Zemper, 2003). The risk of future concussions also increases with every additional concussive event. This is due to the fact that after a biomechanical injury to the brain, a cellular energy crisis is triggered (Giza & Hovda, 2001). The resulting energy crisis makes the brain more vulnerable and thus a second concussion can cause already injured neurons to suffer irreversible injury. This is especially problematic as studies have established a link between recurrent concussions and later life cognitive difficulties in a variety of contact sports such as ice hockey, football, and rugby (Rabadi & Jordan, 2001; Guskiewicz et al., 2005; McCrory et al., 2013;). Additionally, research has also emerged indicating that even repeated sub-concussive impacts (impacts below the threshold of incurring a concussion injury) to the head can lead to a form of neurodegeneration known as CTE, or chronic traumatic encephalopathy (McKee et al., 2009). Symptoms of CTE typically take years to present themselves and can present as negative changes in mood, cognition, and behavior (Baugh et al., 2012). Recently, various high-profile athletes have expressed their concerns about CTE and have walked away from professional sport due to the risks CTE present, further adding to the negativity surrounding head-related sport injuries (Edlow & Hinson, 2015). Ultimately, as further evidence continues to emerge highlighting the long-term consequences of head trauma in sport, it is important for researchers to continue to investigate ways to mitigate their negative effects and make participation in sports safer for athletes of all ages.

## 1.2. Head Injuries in Ice Hockey

Ice hockey is the most popular contact sport in Canada and has 1.3 million participants annually (Canadian Heritage, 2013). Historically, the game had been associated with injuries to the face and head attributed to a lack of protection (Biasca, Wirth & Tegner, 2002). Sweden in the 1950s became the first country in which the use of hockey helmets became common, eventually becoming mandatory for all levels of

Swedish hockey in 1963. In North America, helmets remained unpopular with both players and fans. It wasn't until the death of NHL player Bill Masterson, resulting from a brain injury caused by his head hitting the ice in 1968, that perceptions about helmets began to change and their use in North American hockey leagues started to slowly rise. In 1975, the Canadian Amateur Hockey Association (CAHA) implemented the requirement for all amateur hockey players to wear a helmet certified by the Canadian Standards Association (CSA), and in 1979, the National Hockey League (NHL) followed suit by making helmets mandatory for all incoming players (Pashby, 1993). These early helmets were designed with a plastic shell to prevent skull fractures and severe focal brain injuries such as subdural hematoma. While these helmets proved to be successful in reducing the incidence of deaths caused by head injuries in ice hockey, there continued to be a high incidence of ocular, dental, and facial injuries (Odelgard, 1989). Eye injuries and blinding were especially areas of concern, often the result of being struck with a stick or puck. To prevent these catastrophic injuries, facial protectors were developed in 1972 and became available in two forms: full caged facemasks that covered the entirety of the face and jaw, and half plastic visors that covered the eyes but still left the mouth and jaw exposed (Biasca, Wirth & Tegner, 2002). Full facemasks proved to be very effective in preventing eye injuries when worn properly and in 1978 - the CAHA required all amateur hockey players to wear a CSA certified facemask (Asplund, Bettcher & Borchers, 2009). However facial protectors, especially full facemasks, remained unpopular at the professional level due to complaints of obscuring vision as well as perhaps being related to a culture amongst ice hockey players who view players wearing a visor to be less "tough" than those who do not (Chong & Restrepo, 2011). It thus wasn't until 2013 that the NHL finally mandated that all players coming into the league be required to wear at least a half plastic visor.

Despite the advances in helmet technology and the slow change in culture to embrace further facial protection, ice hockey still accounts for the greatest number of sports-related head injuries that occur in youth (Cusimano et al., 2013), and concussions are the most common type of injury suffered by both male and female hockey players (Agel & Harvey, 2010). Additionally, it has been suspected that bigger and faster players, as well as new equipment, have increased the risk of concussion in recent years (Wennberg & Tator, 2003). Unfortunately, it has proved difficult for researchers to determine the exact biomechanical mechanisms and injury threshold that cause concussion injuries in ice hockey. However, in recent years many groups have attempted to gain a better understanding of these mechanisms by outfitting the helmets of hockey players with sensor technology in order to measure impacts that occur in real-life settings (actual games) (Mihalik et al., 2010; Guskiewicz & Mihalik, 2011; Rowson & Duma, 2011; Mihalik et al., 2012; Wilcox et al., 2014). The Head Impact Telemetry (HIT) sensor system commonly used in these studies, developed at Virginia Tech, has been extensively used in football, and has been validated based on correlations from measurements taken from helmets on the head of a Hybrid III dummy, the standard dummy used in impact studies (Rowson, Brolinson, Goforth, Dietter, & Duma, 2009; Rowson & Duma, 2011). These studies have provided a wealth of knowledge and have demonstrated the association between linear and rotational accelerations of the head with concussion injuries. This in turn has led to the development of novel testing procedures for evaluating helmet performance in reducing concussion injury in both football and, as of recently, hockey (Rowson & Duma, 2011; Rowson, Rowson, & Duma, 2015).

Determining the game situations that most often cause head injuries is difficult, especially based on most player recollections or eyewitness accounts, due to the fast nature of the game. Recently studies have shown that archived video footage is a reliable resource for determining the sequences of events leading to an injurious event (Mihalik et al., 2010; Hutchison, Comper, Meeuwisse, & Echemendia, 2013a). Studies using this technique have shown that concussion injuries are most often the result of being struck by another player. Player-to-player contact causes approximately 88% of the total concussions in elite play (Hutchison, Comper, Meeuwisse, & Echemendia, 2013b) and 67% of brain injuries in children and youth players aged 5 to 19 (Cusimano et al., 2013). This insight has informed improvements in helmet design and rules of play. While helmets certainly have a role in reducing the number of head injuries sustained in hockey and have been found to reduce the risk of severe traumatic brain injuries, researchers have been unable to find strong evidence that helmets reduce concussion incidence (McCrory et al., 2013). In addition, recent rule changes in the NHL that made targeting a player’s head illegal were found to be ineffective in reducing the number of concussions sustained by players over the course of the season (Donaldson, Asbridge,

& Cusimano, 2013). Thus in addition to improving helmet design and implementing rule changes, there is also a need to improve player skills/knowledge training, the rink environment, and other types of protective equipment in order to reduce the number of concussions suffered by hockey players.

## 1.3. The Role of Shoulder Padding and Risk Compensation in Concussion Injuries

There has been much debate regarding how the design of shoulder and elbow pads can reduce the risk of concussion injuries in hockey. Studies analyzing video footage have shown that 42% of concussions in the NHL were caused by direct contact to the head by the shoulder (Hutchison, Comper, Meeuwisse, & Echemendia, 2013c). As ice hockey has evolved, so too has the design of the shoulder and elbow pads that players wear. It was in the 1930s that shoulder and elbow protectors were first introduced, and in their infancy, were fashioned out of leather caps filled with horsehair (Stoner, 1993). As players became taller, heavier, and faster (Echlin et al., 2010), equipment manufacturers began to produce more heavily armoured body protection. Shoulder and elbow pads became larger and began incorporating designs with hard plastic shells. These changes have also coincided with the observed increase concussion incidence in various leagues in both North America and Europe from the late 1980s to the early 2000s (Biasca, Wirth & Tegner, 2002). While the aim of these pieces of equipment is to reduce injury to the elbows and shoulders of the players that wear them, a potential consequence is that the increased protection may cause players to hit with an increased velocity and force. This is a phenomenon known as risk compensation, where individuals adjust their behavior in response to the perceived level of risk, becoming more careful when they sense greater risk (of incurring a shoulder injury for example) and less careful if they feel more protected. It has been speculated that these pieces of protective equipment are also being used as offensive weapons to inflict concussion injuries, negating the effects of protective equipment such as helmets (Hagel & Meeuwisse, 2004; Tator, 2012). Therefore, it is a possibility that the evolution of shoulder padding design over the years may have partially contributed to the increased rates of minor traumatic brain injury observed in ice hockey.

Risk compensation and the role it has on aggressiveness has been investigated in football. Generally, football tackles involve a player using his upper body to drive into an opposing player. The introduction of the plastic shelled helmet in football led to a shift in tackling technique, causing players to begin tackling by using the head as the initial point of contact with the opponent. The rise of this technique, known as spearing, was speculated to be caused by players feeling that their head was well protected when tackling with the plastic shelled helmet (Mueller, 1998). The plastic shelled helmet therefore would allow players to hit harder while feeling more protected and less at risk of injuring themselves. Researchers also noted that during the time period where spearing was commonly used, there was a significant increase in tackling fatalities, and after a rule change that made spearing illegal, the amount of head-related fatalities in football declined despite an increase in football participation (Mueller, 1998). In recent years, groups have been working to move players away from this dangerous tackling technique as a way to reduce injuries. A randomized controlled trial was conducted with NCAA Division 1 football players where an intervention group performed tackling drills without the use of helmets and shoulder pads (Swartz et al., 2015). Early results from this study showed that, compared to the control group, the intervention group had 30% fewer head impacts per athlete exposure by the season's end, presuming that the players in the intervention group had learned safer tackling methods by practicing tackling without reliance on their protective equipment.

The sport of rugby has also been a target area for risk compensation research. In rugby, players are only allowed to wear a set of thin and soft shoulder pads no thicker than 1 cm when uncompressed (World Rugby Board, 2015). While rugby shoulder pads have been shown to attenuate force under lower loads, they appear to bottom out under higher impact loads (Harris & Spears, 2010). In other words, once the foam has been compressed through the majority of its thickness, its stiffness increases abruptly (Subic, 2007). Despite an increase in shoulder pad usage, there was a slight increase in the number of rugby related shoulder injuries (Rugby Football Union, 2002). This could perhaps be the result of players overestimating the amount of protection that the shoulder pads provide, leading to players behaving more aggressively. Similarly, risk compensation has also been speculated to play a role in the headgear worn by rugby players. Studies have found that rugby players feel that they have the ability to play harder and feel more protected while wearing headgear compared to when playing without it (Finch, McIntosh, & McCrory, 2001).

Studies on risk compensation in hockey have focused on the role of facial protection on player aggressiveness, and have reported that recreational ice hockey players who wear some sort of facial protection feel they can play more aggressively than those who wear no facial protection (Woods et al., 2006). Also, while players not wearing facial protection were more likely to incur a facial laceration or bone fracture, those that wear facial protection experience more serious injuries over time, perhaps due to their increased aggressiveness. Similar evidence at the professional level has shown that when wearing a visor, players become more aggressive, offsetting some of the visor's protective effect (Chong & Restrepo, 2011). These players were also assessed more penalty minutes per game when wearing the visor.

An integrated approach to sport injury prevention has recently been proposed that takes into consideration not only biomechanical factors, but also behavioural factors (McIntosh, 2005). In biomechanical terms, an injury occurs when there is transfer of energy to a tissue that results in structural failure (Fung, 1993). Each tissue in the human body has different mechanical properties that govern how it responds to a particular load and what the ultimate strength of that tissue is to that particular load. The peak magnitude of the applied mechanical load depends on the change in momentum in the collision, and the stiffness of the impact site. The magnitude and direction of the load in turn influence the peak linear and rotational accelerations of the head, which govern risk for brain injury (McIntosh, 2005). Therefore, faster and stronger players will lead to higher energy collisions, which have a greater potential to cause tissue failure. Biomechanically focused injury prevention strategies attempt to modify the mechanical load applied to the player. For example, helmets attenuate impact energy and therefore reduce acceleration of the head. However, as an unintended consequence, the helmet also modifies the biomechanical properties of the impact that was received by the player getting hit. As the tackling player is now available to deliver a hit with their own helmet at a greater momentum than they were able to previously, the added protection that the helmet provided to the player receiving the check may be negated. Thus, it is important to consider - and to take into account - behavioural factors such as how a player's risk compensation strategy may be modified when designing biomechanical interventions to prevent injury in sport.

Despite the ongoing public discussion on shoulder padding and risk compensation, there has been very little publically available research on the biomechanical performance of shoulder pads in ice hockey, or the effect that padding stiffness and foam have on peak head accelerations and how this may depend on player risk compensation. In 2010, the NHL began work to reduce the size of shoulder pads and mandated that all shoulder cap covers have at least half an inch of foam padding over them (Shoalts, 2011). Foam has been shown to have effective impact force attenuation properties (Parkkari et al., 1995; Kannus, Parkkari & Poutala, 1999). Research conducted on rugby equipment has also shown that the impact-energy attenuation performance of foam can be incorporated into headgear successfully (A. McIntosh, 2004). It thus logically follows that similar design considerations may prove useful and can go into integrating foam into ice hockey shoulder padding. The Hockey Canada rulebook specifies that “shoulder cap protectors must follow the contour of the shoulder cap without becoming a projection/extension beyond or above the shoulder or shoulder cap. This contoured padding must not be more than 2.54 cm (1 in.) in thickness beyond the top ridge of the shoulder and shoulder cap” (Hockey Canada, 2015). While there is a rule stating that elbow pads that do “not have a soft protective covering of sponge, rubber or a similar material at least 1.27 cm (1/2in.) thick shall be considered dangerous equipment”, no such similar rule exists in the rulebook for shoulder pads (Hockey Canada, 2015).

Despite steps undertaken by the NHL to move towards smaller and limited plastic shoulder padding, only one previous study has assessed the effect of shoulder pad design and the presence of foam on head accelerations (Kendall, Post, Rousseau, & Hoshizaki, 2014). This study used a linear impactor that delivered impacts to a Hybrid III headform at two different impact velocities: 6.5 m/s and 7.5 m/s. The linear impactor was fitted with a variety of different shoulder pads in three experimental conditions – a shoulder pad with a plastic cap inlay, a shoulder cap with no plastic inlay, and a shoulder pad with expanded polypropylene (EPP) foam. Three impacts were delivered to the Hybrid III head per shoulder pad condition under both of the aforementioned impact velocities. No statistically significant differences were found in peak linear acceleration resulting from either impact velocity. However, at an impact velocity of 7.5 m/s, the EPP shoulder pad had a lower peak linear acceleration (127 g, SD= 4.6 g) compared to both the plastic inlay condition (141 g, SD = 13.6 g), and the no plastic inlay condition (141 g, SD = 9.7 g). With regard to peak rotational acceleration, the EPP cap shoulder pad had lower peak rotational acceleration under both impact velocities compared to the other shoulder pad conditions. However, the results were only statistically significant ($p<0.05$) under the 6.5 m/s impact velocity where the EPP cap had a peak angular acceleration of 5882 rad/s$^2$ (SD= 270 rad/s$^2$) compared to 8205 rad/s$^2$ (SD= 758.8 rad/s$^2$) for the plastic inlay condition and 9436 rad/s$^2$ (SD= 146.8 rad/s$^2$) for the no plastic inlay condition. From these results, the group concluded that shoulder pad design “can have a significant effect on the dynamic impact response of a headform” and was indeed influential in reducing peak rotational head accelerations.

## 1.4. Objectives

While the Kendall study (Kendall, Post, Rousseau, & Hoshizaki, 2014) provided the first insight into how shoulder pad design may affect peak head accelerations during shoulder-to-head impacts in a controlled lab setting, further investigation is necessary on how actual hockey players in naturalistic settings modify their hitting behavior and risk compensation while wearing differently designed shoulder pads. As literature investigating other contact sports such as football and rugby shows, it is important for external validity to account for these behavioural factors in evaluating the effect of a novel shoulder pad design on head accelerations.

Accordingly, the goal of the present study was to determine how shoulder pad design affects head impact severity in ice hockey shoulder-to-head impacts. This goal was addressed by conducting laboratory experiments where players delivered checks to an instrumented dummy. Two specific aims were addressed:

Aim 1: To determine how peak linear accelerations and peak rotational velocities of the head during shoulder-to-head collisions are affected by shoulder pad design - specifically by padding stiffness, modified by the addition of a 2cm thick foam overlayer.

Aim 2: To determine if shoulder pad design affects the aggressiveness of players in delivering shoulder checks to the head.

# Chapter 2.

# Materials and Methods

## 2.1. Participants

Fifteen males with competitive ice hockey experience in body checking leagues participated in the study. The participants ranged in age from 16-25 years, with the mean age being 22 years (SD = 2.7). The average body mass of the participants was 85 kg (SD=9.8 kg; ranging from 64 to 104 kg), and the average height was 183 cm (SD=5.2 cm; ranging from 173 to 191 cm). All participants were healthy and had no contraindications for participation in contact sports. All participants provided written informed consent, and in the case of subjects under the age of 19, parental consent was obtained in addition to subject assent. The Research Ethics Committee of Simon Fraser University approved the experimental protocol.

## 2.2. Laboratory Instrumentation

For the experiment, a custom-made body-checking dummy was designed to allow participants to deliver shoulder checks in a safe and controlled fashion. The torso of the body-checking dummy was adapted from a Century “BOB XL” kickboxing dummy. The BOB dummy has a high strength plastisol body that is filled with urethane foam and contains no hard elements. It is used extensively in gymnasiums and its design makes it ideal for receiving multiple impacts. The BOB dummy torso was attached to a metal tube and base for support and was mounted to the ground via a ball joint (as shown in Figure 2.1). The metal tube and base was filled with sand to give the dummy a mass of 61 kg and the tube was covered in foam to ensure there was no risk of injury from contact with a hard surface. When stood upright, the dummy had a height of 178 cm; however, the dummy was also designed to have its height and weight easily adjusted so that it could be used for different age groups and playing levels. The dummy was also secured to an overhead spring/winch system via a one-way locking pulley in order to provide the dummy with a uniform rotational stiffness, as well as to prevent the dummy from rebounding back at participants when checked. The spring/winch system was mounted on a force plate (Bertec model 6040H) which ensured that the initial tension in the spring was constant in all trials.

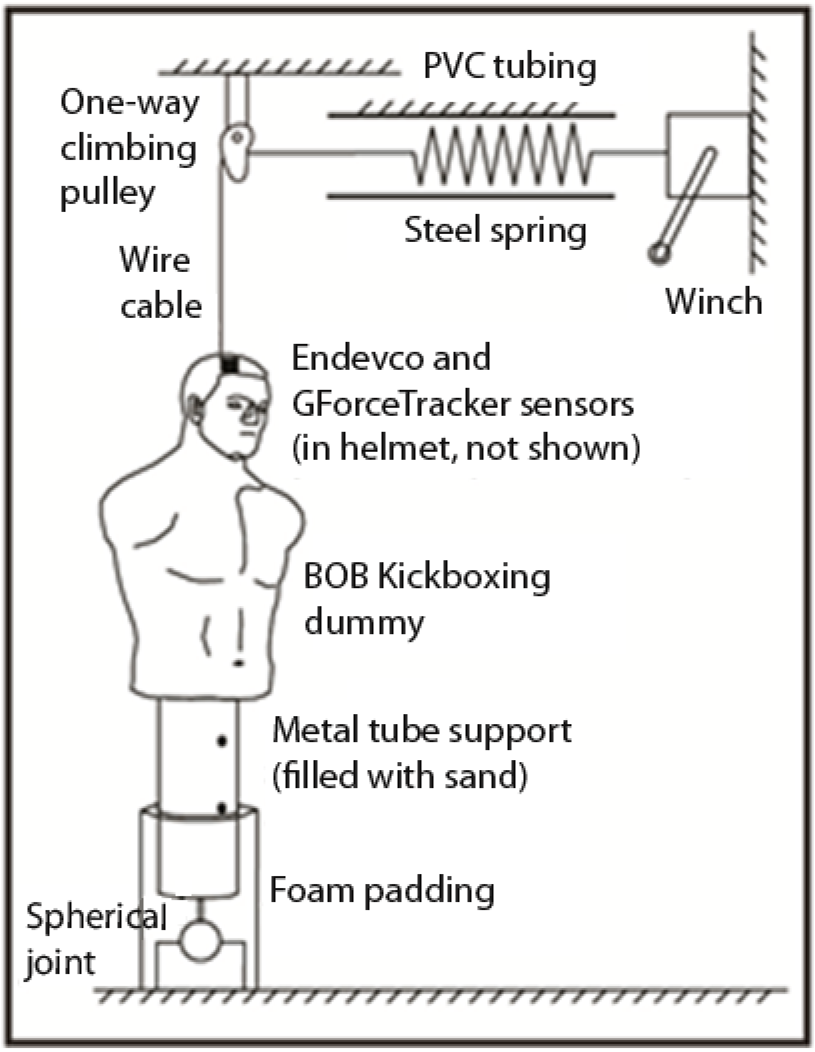
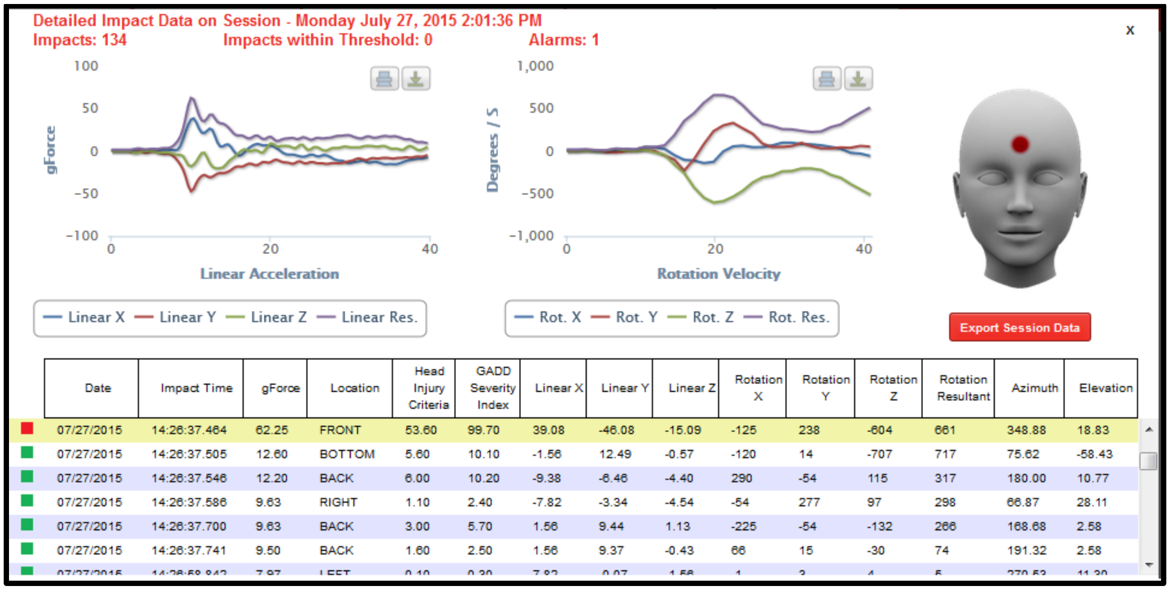



**Figure 2.1. Schematic of the custom made body-checking dummy. The torso of a BOB kickboxing dummy was mounted to a metal tube that was connected to a spherical joint on the ground. The dummy was connected overhead to a spring/winch system allowing its initial stiffness to be controlled and a one-way locking pulley ensured that the dummy was unable to rebound at participants after it was hit. A helmet on the dummy's head contained an Endevco and GForceTracker accelerometer that measured peak linear acceleration and peak rotational velocity respectively.**

Peak linear head accelerations and peak angular velocities were obtained from sensors placed in a CSA certified caged helmet on the dummy's head. Peak linear head accelerations were acquired at 20 kHz via an Endevco 7624C tri-axial linear accelerometer setup and filtered using a 300 Hz low-pass filter compatible with the SAE J211 protocol (Society of Automotive Engineers International, 2007). Angular velocities were acquired at 800 Hz via a G-Force Tracker (GFT) sensor that stored all impacts detected above a set threshold of 8g onboard where it was also filtered with a low pass filter of 100 Hz (Firmware version: GFT2, Artaflex Inc., Markham, ON, Canada). Following the experiment, the GFT was connected to a laptop via micro USB and data was uploaded to a cloud storage. From the cloud storage, the raw data was extracted for analysis as shown in Figure 2.2. The GFT has been used routinely in hockey, football, and lacrosse to quantify the magnitude and frequency of head accelerations experienced by players over the course of a season, as well as in lab based studies (Allison, Kang, Maltese, Bolte, & Arbogast, 2014; Campbell et al., 2015). The G-Force Tracker has also been shown to be comparable with standard approaches, such as the standard HIT system, which have been commonly used to obtain rotational head velocity measures in sport (Campbell et al., 2015). Both the Endevco accelerometer and the G-Force Tracker were adhered using a 3M™ VHB™ adhesive to the left inside crown of the helmet and placed next to each other (Figure 2.3). Lastly, a Nikon S2 high-speed camera (1200 fps) was mounted perpendicular to the dummy in order to record the shoulder position of players coming in to deliver shoulder checks to the dummy.




| Date | Impact Time | gForce | Location | Head Injury Criteria | GADD Severity Index | Linear X | Linear Y | Linear Z | Rotation X | Rotation Y | Rotation Z | Rotation Resultant | Azimuth | Elevation |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| 07/27/2015 | 14:26:37.464 | 62.25 | FRONT | 53.60 | 99.70 | 39.08 | -46.08 | -15.09 | -125 | 238 | -604 | 661 | 348.88 | 18.83 |
| 07/27/2015 | 14:26:37.505 | 12.60 | BOTTOM | 5.60 | 10.10 | -1.56 | 12.49 | -0.57 | -120 | 14 | -707 | 717 | 75.62 | -58.43 |
| 07/27/2015 | 14:26:37.546 | 12.20 | BACK | 6.00 | 10.20 | -9.38 | -6.46 | -4.40 | 290 | -54 | 115 | 317 | 180.00 | 10.77 |
| 07/27/2015 | 14:26:37.586 | 9.63 | RIGHT | 1.10 | 2.40 | -7.82 | -3.34 | -4.54 | -54 | 277 | 97 | 298 | 66.87 | 28.11 |
| 07/27/2015 | 14:26:37.700 | 9.63 | BACK | 3.00 | 5.70 | 1.56 | 9.44 | 1.13 | -225 | -54 | -132 | 266 | 168.68 | 2.58 |
| 07/27/2015 | 14:26:37.741 | 9.50 | BACK | 1.60 | 2.50 | 1.56 | 9.37 | -0.43 | 66 | 15 | -30 | 74 | 191.32 | 2.58 |

**Figure 2.2.** **Screenshot of the cloud based GFT software system that summarizes all impacts that occurred in a session. Impacts are recorded and coded in separate 40 ms segments. Raw data from the system is obtained by clicking the export session data button which produces an excel file with linear acceleration and rotational velocity measures for each sample.**

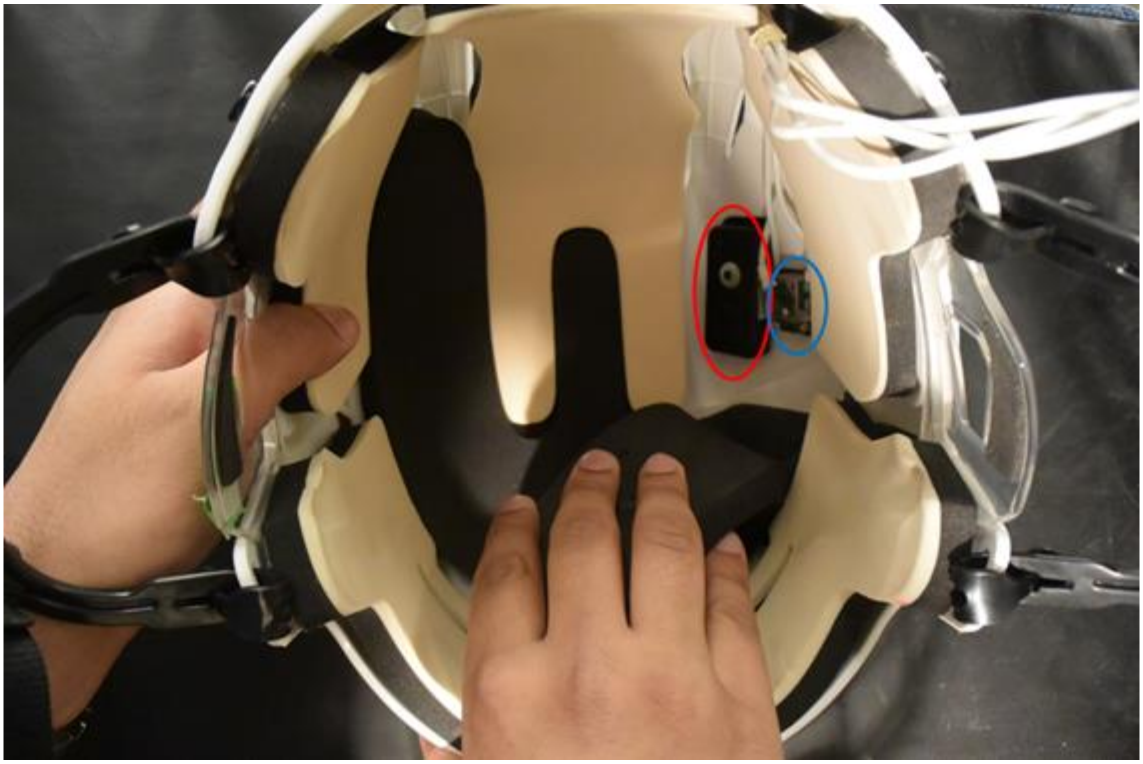


**Figure 2.3. Photograph of the helmet worn by the body-checking dummy. The helmet contains an Endevco 7624C accelerometer setup (circled in blue) and a GFT sensor (circled in red) mounted on the left inside crown.**

Two commercially available and commonly used shoulder pads were obtained for players to deliver their checks. Pad A (the Sherwood 5030 Traditional shoulder pad) is marketed as a lightweight shoulder pad that features a synthetic leather and low-profile foam covering over its hard polyethylene plastic shoulder caps. The rest of the shoulder pad also contains low-profile segmented single-density foam throughout and has lacing to adjust the fitting of the pad. Pad B (the Bauer Supreme One.6 shoulder pad) is more anatomically designed with its shoulder cap fitting closer to the curvature of the shoulder. The plastic cap of Pad B is also covered in layers of medium density foam and light density foam. The rest of Pad B features a multi-panel design that contains plastic inserts covered with a variety of medium-and high-density foams making it bigger and heavier than Pad A (886 grams versus 711 grams). Foam shoulder pad cap overlays were moulded for both Pad A and B. These moulded foam shoulder overlays were 2cm thick for both pads, which is a thinner thickness than the 2.54 cm restriction specified by Hockey Canada for contoured padding above the shoulder cap (Hockey

Canada, 2015). The effect of 1 cm foam thickness versus 2 cm foam thickness was investigated in a series of pilot tests. Results from the pilot testing indicated that a 2 cm foam thickness caused on average, 3.6g (SD = 0.8) and 81.3 deg/s (SD = 55.5) lower head accelerations of the body checking dummy compared to a 1 cm foam thickness. The foam cap overlays were constructed with a moulded layer of polyurethane foam (with density of 220 kg/m$^3$) that matched the existing curvature of the shoulder cap (See Figure 2.4). They were placed over the existing shoulder cap of both pads and were adhered in place using double-sided tape, as well as a mesh net, that tightly held the foam caps in place in order to minimize movement over the plastic cap.


**Figure 2.4. Diagram showing Pad A (on left) and Pad B (on right) with the moulded foam cap adhered on top of the right shoulder cap. Pad A is lighter in construction than Pad B and contains much less foam over its plastic cap. Pad B contains multiple layers of foam over its plastic cap and features a more rounded cap profile. The foam overlays that were moulded for each pad were 2cm thick and designed to match the shape and curvature of the underlying plastic cap.**

## 2.3. Experimental Protocol

Participants were instructed to administer with their right shoulder “the hardest shoulder check they were comfortable delivering” to a marked target located on the helmet cage of the checking dummy with their right shoulder. Players were given a straight, 3m run-up to approach the dummy and were allowed to build up a speed of their choosing before making contact with the target. Players were given a minimum of three practice trials prior to the start of the experiment in order to familiarize them with the procedure, and the mass and stiffness of the checking dummy.

Figure 2.5 illustrates the experimental protocol undertaken by each participant. Checks were delivered to the body-checking dummy in two different impact configurations:

a) Directly in front of the dummy (Straight-On Impact)

b) 40° angle to the dummy (Angled Impact)

A pilot video analysis study showed that these are two common angles at which checking players approach opposing players to deliver head impacts in hockey. The straight-on impacts involved the dummy leaning forward at a 20° angle and were designed to simulate a severe hit where a player would be skating with his head down. In this scenario a player would not be anticipating a hit and therefore more likely to suffer a severe head impact (Mihalik et al., 2010) The angled hits had the dummy standing straight up and were designed to simulate a blind-side hit. Similarly in this hit, a player may not anticipate the incoming check as the trajectory of the incoming player would be outside the field of view of the player being hit. Often this hit occurs when a player is focusing on shooting or passing the puck and is not aware of an opposing player approaching at an angle towards him (Mihalik et al., 2010).

In each of the two impact configurations, trials were conducted with participants using four shoulder pad combinations:

a) Shoulder Pad A unmodified

b) Shoulder Pad A modified with a 2cm foam cap

c) Shoulder Pad B unmodified

d) Shoulder Pad B modified with a 2cm foam cap

Participants delivered five checks with each of the four randomly presented shoulder pad combinations. The order of shoulder pad combinations was randomized via the use of a random number table.

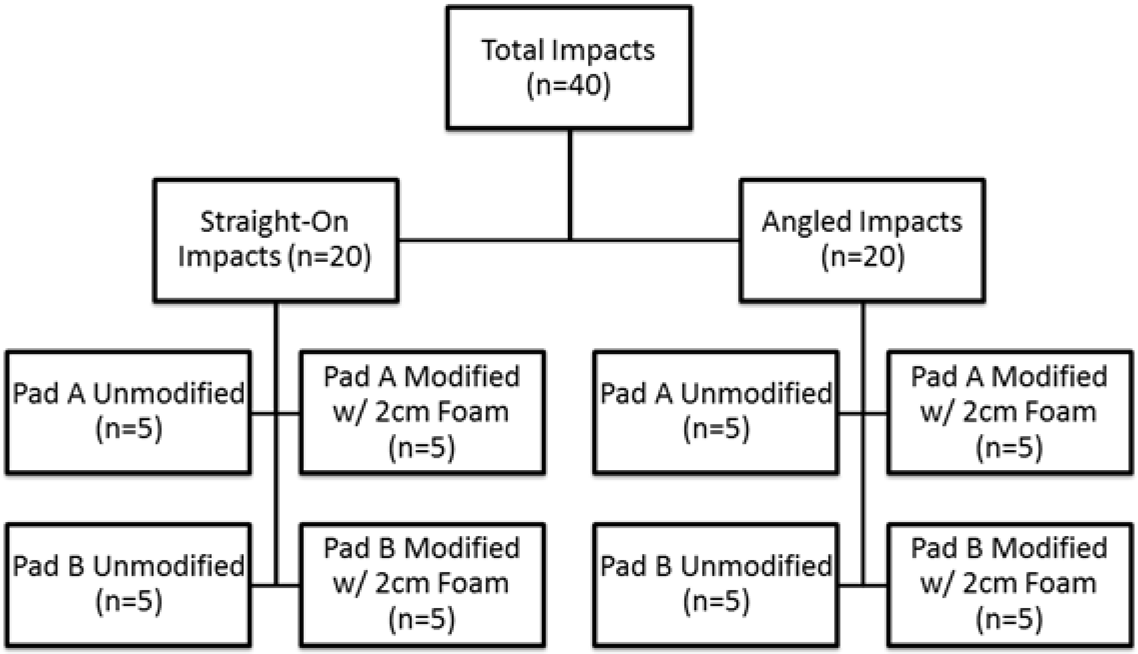



**Figure 2.5. Organizational chart illustrating the experimental protocol. Participants delivered forty impacts to the body-checking dummy. Twenty of these impacts were conducted with the participant approaching the dummy straight-on, and the other twenty were delivered with the participant approaching at a forty-degree angle to the dummy. In each of these impact configurations, participants delivered five impacts with each of the four shoulder pad combinations. Participants were randomly assigned to complete either the straight-on impacts or angled impacts first and within each impact type, the order of shoulder pad combinations was randomized.**

After all trials were completed, participants completed a post-study questionnaire which asked on a modified Likert-scale, how comfortable they felt in delivering checks to the dummy while wearing each of the four shoulder pad conditions, from 1 (extreme discomfort in the shoulder) to 5 (extreme comfort in the shoulder). This was done in order to gauge the subject's own perceptions towards each shoulder pad combination as these may have affected the amount of risk they took in delivering checks. Additionally, the questionnaire asked participants to rate how checking the dummy compared to checking an unsuspecting player in a real game on a scale from 1 (completely different) to 10 (completely similar).

## 2.4. Data Analysis

For each trial, the peak linear acceleration at impact was extracted from the raw data of the Endevco accelerometer. The peak angular velocity at impact was extracted from the raw data trace obtained from the G-Force Tracker cloud based software system (as shown previously in Figure 2.2). Typical raw traces of the 3D accelerations (from the Endevco accelerometer) and 3D rotational velocities (from the GFT) for foam padded versus unpadded shoulder pads are shown in Figure 2.6. Peak values were taken from the initial stage of impact (the first 40ms in which the participant's shoulder would first make contact to the cage of the dummy). This was done as in some cases, depending on how participants followed through with their check, a higher peak would occur later on in the trial. This additional peak was observed rarely and always occurred after the initial 40ms of impact. The accuracy of each trial was assessed by reviewing the high-speed video to ensure that participants were making contact with the target on the dummy's cage.

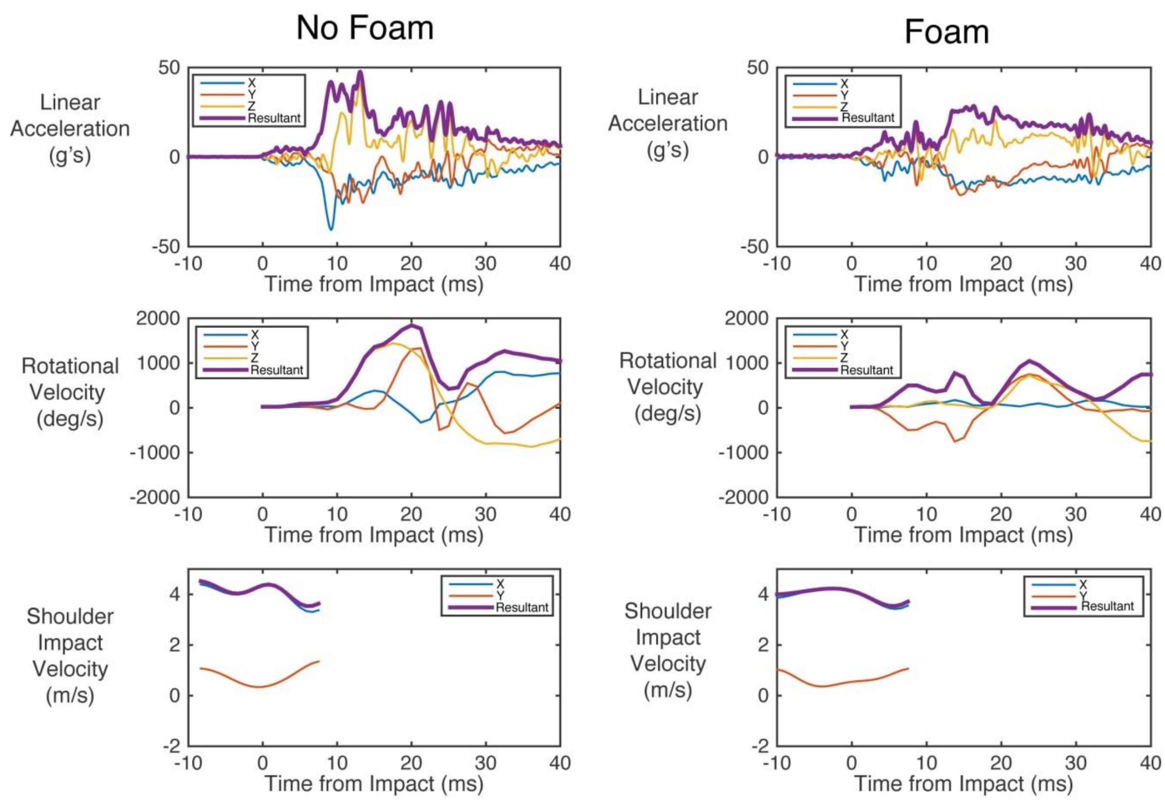



**Figure 2.6.** **Typical raw traces of the 3D accelerations (top), 3D rotational velocities (middle), and 2D shoulder impact velocities (bottom) observed for unpadded (left) versus foam padded (right) shoulder pads. 3D accelerations (x-axis = perpendicular to the coronal plane; y-axis = perpendicular to the sagittal plane; z-axis = perpendicular to the transverse plane) were obtained from the Endevco accelerometer setup for a continuous 5 second recording period and a Matlab script was used to calculate the resulting magnitude. 3D rotational velocities (x-axis = perpendicular to the sagittal plane; y-axis = perpendicular to the coronal plane; z-axis = perpendicular to the transverse plane) were obtained from the GFT. The GFT collects data based upon a user-inputted threshold (8g). The sensor records and saves 8ms of the impact prior to the threshold and 32ms of the impact following the threshold. Peak rotational velocity was obtained from the exported raw GFT data. 2D shoulder impact velocities (x-axis = perpendicular to the coronal plane; y-axis = perpendicular to the transverse plane) were obtained from differentiating the position data obtained from the DLTdv5 toolbox to velocity units. The velocity trace was filtered with a fourth-order 100Hz low-pass Butterworth filter in order to remove the high frequency noise associated with digitizing error. Time "0" denotes the frame in which shoulder impact was determined to occur with the dummy's cage and the resulting shoulder velocity at that point was calculated.**

The high-speed video for each trial was loaded into an open source Matlab toolbox commonly used for digitizing video files (Version DLTdv5; Hedrick, 2008). The DLTdv5 toolbox was used to digitize the position of a marker located on the back of the participant's right scapula into x and y coordinates for each frame (as show in Figure 2.7). The pixel tracks obtained from the DLTdv5 toolbox were then scaled to distance units using a known scale in the video frame located on the helmet of the body-checking dummy. The distance units were then differentiated to velocity units using Matlab and filtered using a fourth-order, 100 Hz Butterworth low-pass filter in order to remove the high frequency noise associated with digitizing error (Mountcastle, Ravi, & Combes, 2015). A typical raw trace of this is illustrated in Figure 2.6. The high-speed video was examined to determine which frame the subject first made initial contact with the cage. This could be seen as the cage would noticeably deflect once contact was made. The shoulder velocity at contact was then taken for the corresponding frame selected.

**Figure 2.7. Screenshot of a frame from the high-speed camera. The video clip was loaded into the DLTdv5 toolbox that was used to digitize the location of a marker located on the back shoulder of each shoulder pad. The red circle denotes the point marked by the toolbox for this particular frame. A minimum of fifteen frames prior to contact were digitized in order to differentiate the resulting position data into velocity units.**

## 2.5. Statistical Analysis

Each primary outcome (peak linear acceleration, peak rotational velocity, and shoulder impact velocity) was analyzed by conducting a two-factor repeated-measures analysis of variance for both impact configurations (straight-on and 40°). This analysis was chosen in order to control for participant differences in skill, hitting technique, and mass, that may influence outcome measures. For each of the three main outcome measures, the main effects of the presence of foam (foam cap vs no foam cap) as well as the type of shoulder pad (Pad A versus Pad B), were investigated. Interactions between foam presence and shoulder pad type were also investigated. A level of 0.05 or less was chosen to indicate statistical significance. Post-hoc paired t-tests were conducted to determine the differences between paired combinations of padding conditions in mean peak linear acceleration, peak rotational velocity, and shoulder impact velocity. Descriptive statistics were used to summarize the results obtained from the post-study questionnaire completed by participants. All analyses were performed using JMP (Version 12.0.1).

# Chapter 3.

# Results

## 3.1. Summary of Statistical Results

A summary of all statistical results from the two-factor repeated-measures ANOVA is presented in Table 3.1. The presence of foam over the shoulder caps was independently associated with peak linear acceleration and rotational velocity, but not horizontal shoulder impact velocity, for both impact configurations. The type of shoulder pad was associated with peak linear acceleration, and for straight-on impacts, with peak rotational velocity. Interactions between foam and shoulder pad type were weak or non-existent.

**Table 3.1: Summary of Statistical Results**

| | Straight On Impacts | | Angled Impacts | |
|---|---|---|---|---|
| | **F- Ratio** | **P- Value** | **F-Ratio** | **P-Value** |
| ***Peak Linear Acceleration*** | | | | |
| Foam | 33.29 | <.0001* | 22.53 | <.0001* |
| Type of Shoulder Pad | 10.15 | .003* | 3.99 | .05* |
| Foam x Type of Shoulder Pad | 4.24 | .05* | 0.05 | .82 |
| ***Peak Rotational Velocity*** | | | | |
| Foam | 5.39 | .03* | 7.40 | .01* |
| Type of Shoulder Pad | 5.01 | .03* | 0.08 | .78 |
| Foam x Type of Shoulder Pad | 0.22 | .64 | 1.90 | .18 |
| ***Shoulder Impact Velocity*** | | | | |
| Foam | 0.01 | .92 | 1.00 | .32 |
| Type of Shoulder Pad | 2.37 | .13 | 1.66 | .20 |
| Foam x Type of Shoulder Pad | 0.50 | .48 | 1.00 | .82 |

[*] denotes statistical significance at a level of 0.05 or less

## 3.2. Peak Linear Acceleration Results

When players delivered checks with foam-modified pads versus unmodified pads, there was a decrease of 27.7% in the average value of peak linear head acceleration of the dummy during straight-on impacts (mean = 31.13 g versus mean = 43.04 g; mean difference = 11.91 g (*SE* = 2.06 g); $p < .0001$). The difference found for the angled impacts was similar, with a 21.6 % decrease in peak linear acceleration when pads were modified with foam compared to when they were unmodified (mean = 26.34 g versus mean = 33.59 g; mean difference = 7.25 g (*SE* = 1.53 g); $p < 0.0001$). Individual participant change in average values of peak linear acceleration between foam and no-foam conditions is shown in Figure 3.1. In all cases, at least 11 of the 15 participants had a lower average value of peak linear acceleration while wearing foam-covered pads compared to non-foam covered pads.

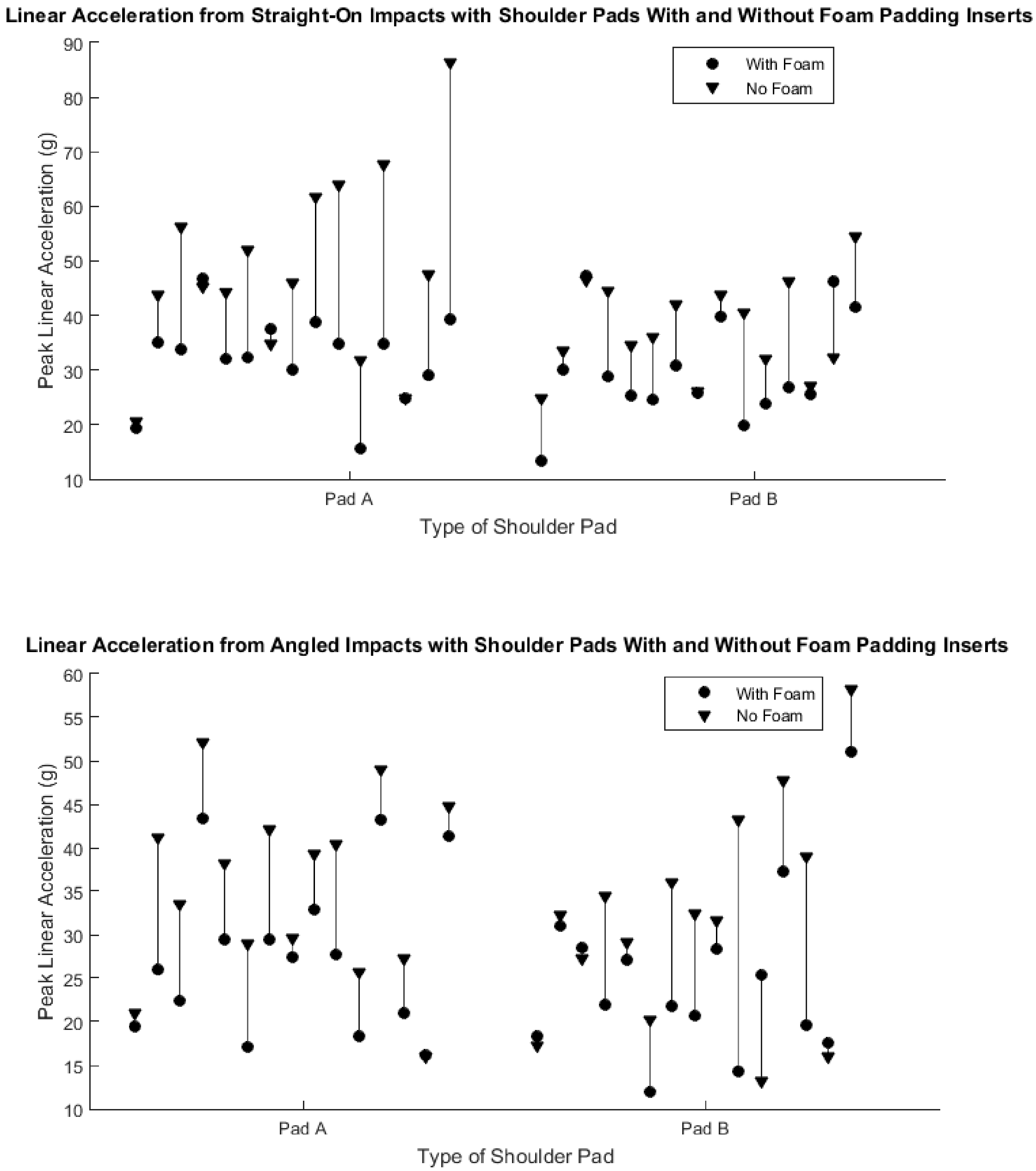
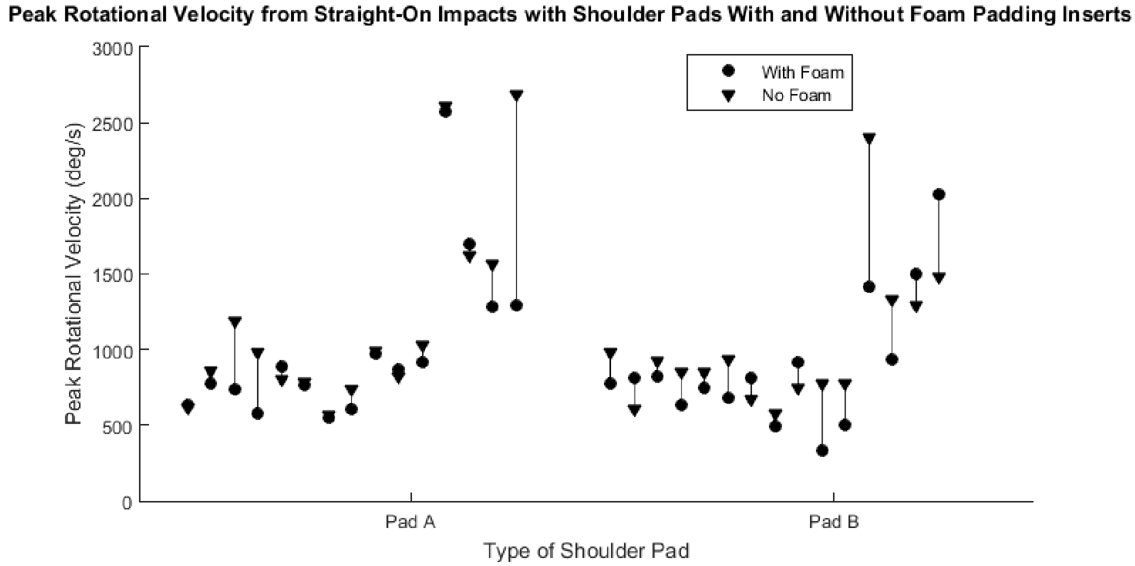



**Figure 3.1.** **Individual participant change in peak linear acceleration for straight-on (top panel) and angled impacts (bottom panel) between foam covered pads (circles) and non-foam covered pads (triangles) for both types of shoulder pads (Pad A and B). Peak linear acceleration was, on average, 11.91 g (*SE* = 2.06), and 7.25 g (*SE* = 1.53 g) less in straight-on and angled impacts respectively, when checks were delivered with foam modified pads versus unmodified pads.**

Evidence was also found ($p$ = 0.003) that peak linear acceleration was, on average, 6.58 g ($SE$ = 2.06) greater when players delivered checks with Pad A (mean = 40.38g) versus when they checked with Pad B (mean = 33.80g) in the straight on impacts. In angled impacts, similar evidence was found ($p$ = 0.05) where, on average, impacts delivered with Pad A (mean = 31.49g) were 3.05 g ($SE$ = 1.53) greater than those delivered with Pad B (mean = 28.44g). There was also evidence of an interaction effect between the presence of foam and type of shoulder pad used in straight on impacts ($p$ = 0.05). Examination of a profile plot indicated that the presence of foam had a more significant reduction in peak linear acceleration when it was applied over Pad A in comparison to Pad B. This interaction effect was not detected in the angled impacts ($p$ = 0.82).

## 3.3. Peak Rotational Velocity Results

The presence of foam also had an effect on the peak rotational velocity experienced by the dummy's head in both straight-on ($p$ = 0.03) and angled impacts ($p$ = 0.01). On average, foam modified pads (mean = 951.56 deg/s) caused impacts that were 13.8% less than that of their unmodified counterparts (mean = 1103.87 deg/s) in straight-on impacts (mean difference = 152.31 deg/s ($SE$ = 65.58)). In angled impacts, this decrease was 10.5% (mean = 724.44 deg/s for modified pads versus mean = 809.43 deg/s for unmodified pads; mean difference = 85.00 deg/s ($SE$ = 31.24)). Individual participant change in peak rotational velocity between foam and no foam conditions is shown in Figure 3.2.



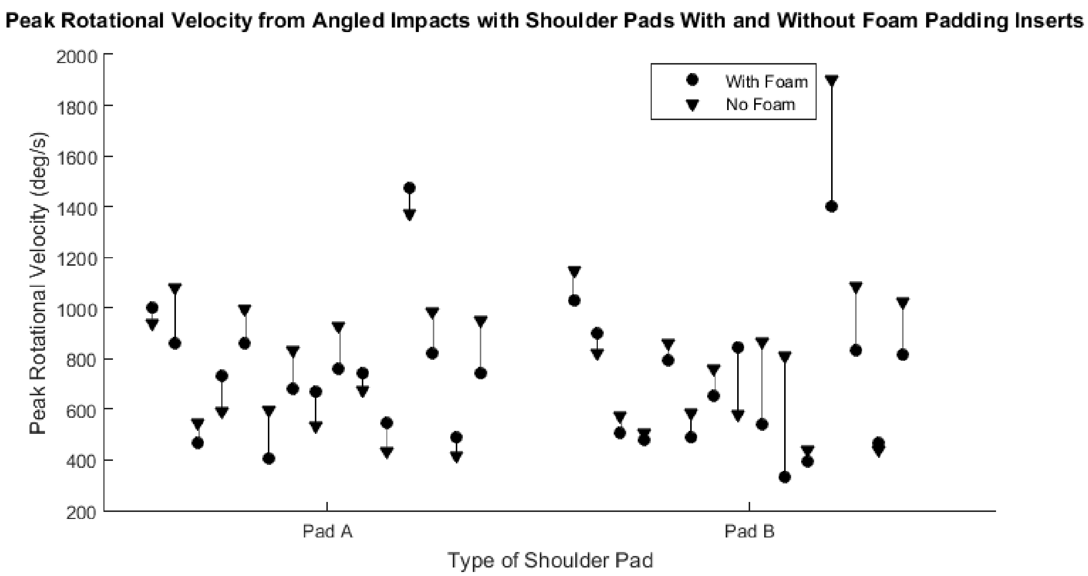




**Figure 3.2. Individual participant change in peak rotational velocity for straight-on (top panel) and angled impacts (bottom panel) between foam covered pads (circles) and non-foam covered pads (triangles) for both types of shoulder pads (Pad A and B). Peak rotational velocity was, on average, 152.31 deg/s (*SE* = 65.58), and 85.00 deg/s (*SE* = 31.24) less in straight-on and angled impacts respectively, when checks were delivered with foam modified pads versus unmodified pads.**

Evidence was found for a main effect of shoulder pad on rotational velocity in straight on impacts (mean = 1101.10 deg/s for Pad A versus mean = 954.33 deg/s for Pad B; mean difference = 146.77 deg/s; $p = 0.03$). This effect was not detected in angled impacts ($p = 0.78$). There were no interaction effects between presence of foam and type of shoulder padding on peak rotational velocity for either straight-on impacts ($p = 0.64$) and angled impacts ($p = 0.18$).

## 3.4. Shoulder Impact Velocity Results

In regards to shoulder impact velocity, no evidence was found that the presence of foam had an effect on player shoulder impact velocity for either straight-on impacts ($p = 0.92$) or angled impacts ($p = 0.32$). Individual participant change in peak shoulder impact velocity between foam and no foam conditions is shown in Figure 3.3.

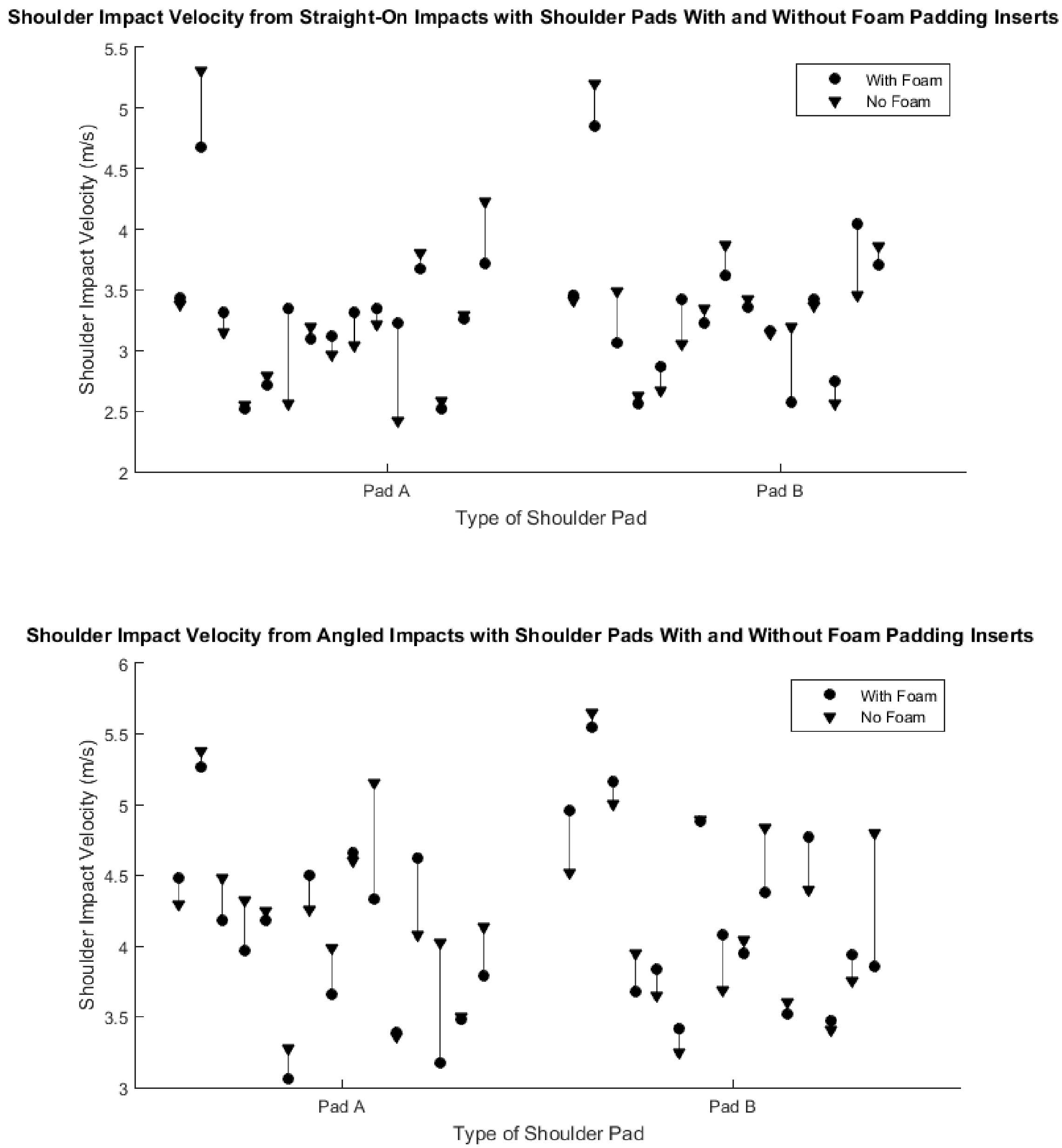



**Figure 3.3.** **Individual participant change in shoulder impact velocity for straight-on (top panel) and angled impacts (bottom panel) between foam covered pads (circles) and non-foam covered pads (triangles) for both types of shoulder pads (Pad A and B). No evidence that the presence of foam had an effect on shoulder impact velocity was found.**

The type of shoulder padding players wore was also shown to have no effect on the mean shoulder impact velocity in both types of hits (straight on: $p = 0.13$; angled: $p = 0.20$). No interaction effects were detected in either straight-on impacts ($p = 0.48$) nor angled impacts ($p = 0.32$).

## 3.5. Post Study Questionnaire Results

Figure 3.4 illustrates how comfortable participant ratings of comfort in delivering checks to the dummy with each of the four shoulder pad conditions.

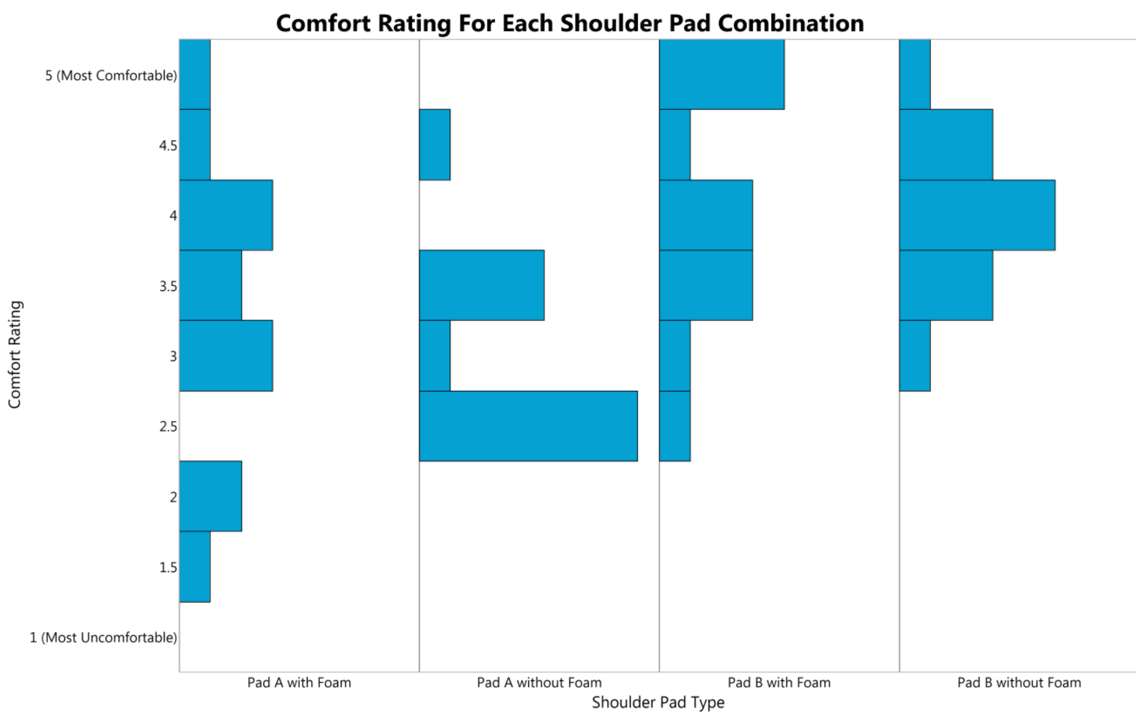



**Figure 3.4. Histogram showing the distribution of participant responses to how comfortable they felt delivering checks (on a scale of 1-5) in each of the four shoulder pad conditions.**

Participants generally felt more comfortable checking with Pad A (*Mdn*: 3.5) when it was covered with a foam cap compared to the standard pad with no foam cap (*Mdn*: 2.5). With regard to Pad B, the median response was the same for both the foam-modified (*Mdn*: 4) and standard unmodified pads (*Mdn*: 4). Participants also generally reported feeling more comfortable delivering checks when wearing Pad B (*Mdn*: 4) compared to when they were wearing Pad A (*Mdn*: 3)

## 3.6. Order Effect Results

The effect of shoulder pad combination order was investigated to examine whether the severity of impacts delivered by the participants changed over the course of the experiment, potentially due to fatigue or becoming more comfortable with the experimental paradigm. No evidence was found that the shoulder pad condition order had an effect on either peak linear acceleration ($p = 0.82$) or peak rotational velocity ($p = 0.95$). Additionally, no evidence was found that shoulder pad condition order had an effect on shoulder impact velocity ($p = 0.99$).

## 3.7. Peak Linear Acceleration and Peak Rotational Velocity Correlation

A Pearson correlation coefficient was computed to assess the relationship between the peak linear acceleration and peak rotational velocity. Only a slight correlation was found between the two variables, $r = 0.20$, $p < 0.0001$. A scatterplot illustrates the results (Figure 3.5).

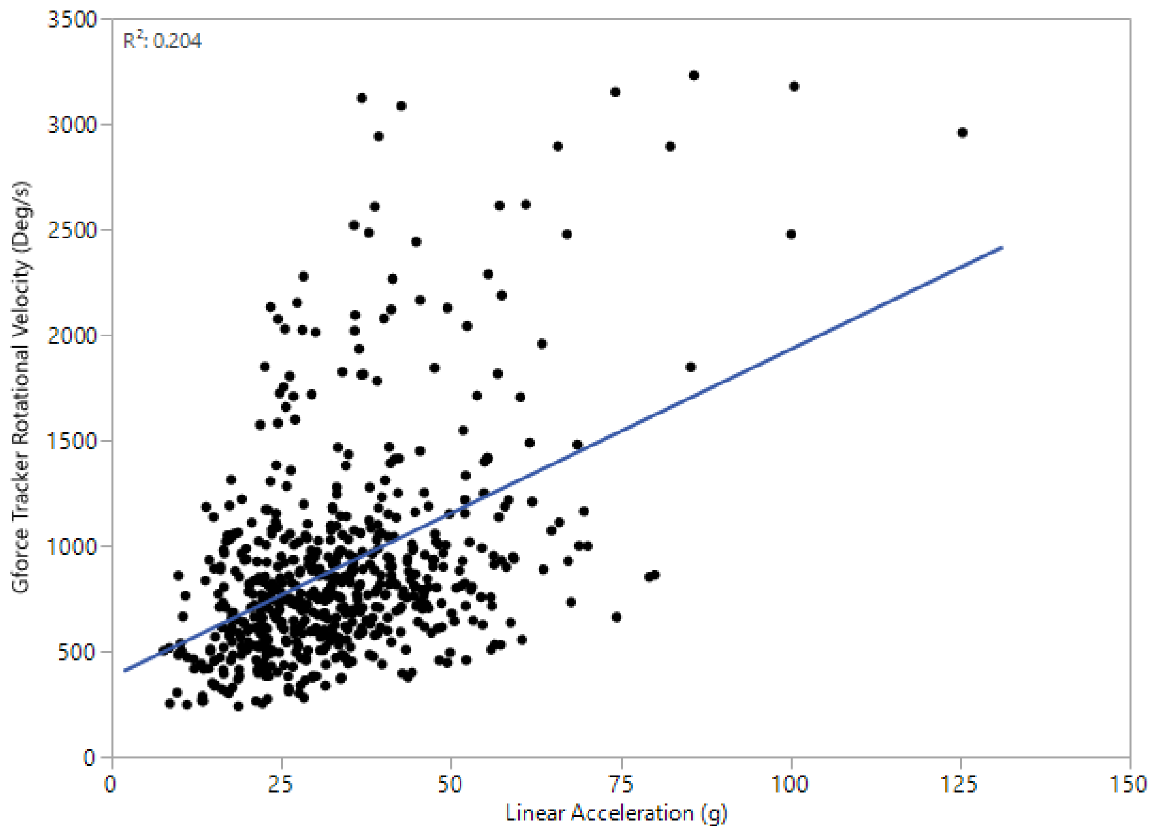



**Figure 3.5.** **Scatterplot showing the correlation between the peak linear acceleration and peak rotational velocity. A moderate correlation was found between the two variables, r = 0.20, p < 0.0001.**

# Chapter 4.

# Discussion

## 4.1. Head Impact Severity due to Shoulder Pad Design

The first aim of the study was to determine how a foam insert covering existing shoulder pads influenced the peak linear accelerations and peak rotational velocities experienced by the head in a shoulder check. Previous literature has shown that concussion injuries tend to occur as a result of a combination of both the linear and rotational components of an impact; thus, it is important to evaluate the effect that shoulder pad design has on both components (Ommaya, 1985). Additionally, our results (as well as previous literature; Rowson, Brolinson, Goforth, Dietter, & Duma, 2009) show that peak linear acceleration and peak rotational velocity are only slighty correlated. Research using sensor technology in contact sport has shown that the mean peak linear acceleration which a player experiences in a head impact is approximately 20 g (Mihalik et al., 2010; Wilcox et al., 2014), with concussive impacts occurring on average at approximately 105 g (Rowson & Duma, 2011). On average, the hits participants delivered in the present study had a peak linear acceleration of 33.5 g (SD= 14.7), and the range of values observed were 7.7 g – 125.4 g (n=600), highlighting that the peak linear acceleration observed in the present experiment is similar to values observed in actual hockey games. With regard to peak rotational velocity, field studies have shown that the average concussive impact occurs at 1277.7 deg/s and the average sub-concussive impact is around 315.1 deg/s (Rowson et al., 2012). The average peak rotational velocity observed in the present experiment was 896.2 deg/s (SD = 504.5 deg/s) and values ranged from 209.25 deg/s to 3228.8 deg/s (n=600), which once again indicates that the hits observed in the laboratory are comparable to those that occur on the ice. Much of this literature comes from athletes at the varsity level. The subjects who participated in this study ranged in age from 16 to 25 and played in various contact leagues including Amateur AAA, Junior hockey, and Varsity hockey. While this study was not specifically powered to look at the effect of age and level of play on impact severity, no correlation was found between player age and head impact severity. This indicates that these various leagues have players capable of delivering similar impacts despite having differences in player age and maturity.

The results from the present study indicate that a 2 cm thick foam layer overlying the cap of shoulder pads can reduce the peak linear acceleration experienced by the head by 21.6 - 27.7%, and can reduce the peak rotational velocity experienced by the head by 10.5 - 13.8%. Studies have successfully created risk contours relating peak linear and rotational head acceleration with the risk of incurring a concussion based on data captured from instrumented football players (Rowson & Duma, 2013). Thus, it is reasonable to assume that the observed reduction in both peak linear head acceleration and peak rotational velocity caused by the presence of foam on shoulder pads would reduce the risk of incurring a concussion injury when a player is hit in a shoulder-to-head impact. Additionally, this reduction in both peak linear head acceleration and peak rotational velocity may also have an effect in sub concussive impacts (which are speculated to play a role in the development of CTE; McKee et al., 2009). Therefore, reducing the magnitude of these frequent impacts can potentially play a very important role in reducing the development of CTE in athletes. Foam padding has been well documented to have effective force attenuation properties, and has been used in a variety of biomechanical prevention strategies in both sport and medical settings (Parkkari et al., 1995; Kannus, Parkkari & Poutala, 1999; A. McIntosh, 2004). While various designs of foam padding are already being implemented in hockey helmets in order to attenuate head impact forces, padding on shoulder pads does provide additional force attenuation in the very common shoulder-to-head collisions that occur in the game. The foam overlying the shoulder pads can undergo considerable deformation thanks to the material's elastic properties and can thus reduce the force transmitted to a player's helmet. Foam may also spread the area of contact, and reduce peak pressure.

Of note is that, twenty-seven out of the six hundred impacts recorded had a peak linear acceleration of 60g or greater, and these twenty-seven impacts were delivered by a combination of eight players. However, only one of these twenty-seven impacts was delivered when a subject was wearing a shoulder pad that had a foam cap covering overlay. It has been previously suggested that concussive impacts can result from a wide range of magnitudes of peak linear acceleration, from 60g and upwards to 170g (Guskiewicz & Mihalik, 2011). Therefore, by putting foam caps over a standard set of shoulder pads, it was possible to almost completely eliminate impacts that occurred in a "concussion risk zone" for this laboratory paradigm. This also lends support towards the NHL's work in mandating foam covering on all shoulder padding and perhaps even increasing the mandated amount. While peak linear accelerations and rotational velocities experienced by the head in shoulder-to-head impacts can be decreased thanks to the force attenuation properties of foam, it should be noted that this alone is not sufficient in preventing all of the brain injuries that occur due to these types of hits (especially during hits of high impact severity).

With regard to rotational results, the findings presented in this study also support those of the recent study conducted by Kendall and colleagues on the effect that ice hockey shoulder padding has on head impact dynamics using a Hybrid III head and neck mechanical system (Kendall et al., 2014) in showing that foam has a role in reducing the angular dynamics of the head in shoulder-to-head impacts in ice hockey. However, the findings of the present study have some differences with regard to peak linear acceleration. While Kendall and colleagues did not find any statistically significant differences between their foam cap and full shoulder pad conditions, they did see a trend towards the EPP cap having a lower peak linear acceleration at a 7.5 m/s impact velocity. In line with this, the present study found a statistically significant decrease in peak linear acceleration when pads were modified with a foam cap. Possible explanations for this discrepancy perhaps lie in the different methodology employed by Kendall et al., as well as the impact velocity at which the checks were delivered. The present study had trained hockey players delivering impacts, as opposed to the linear impactor that was used in the Kendall et al., study. While this resulted in the present protocol having more inherent variability than a protocol utilizing a controlled linear impactor, it also allowed us to assess the effect of foam under conditions where players (instead of a mechanical test system) delivered the impacts. Additionally, while Kendall et al. performed trails at 6.5 m/s and 7.5 m/s, a range of impact velocities from 2.5 m/s to 5.7 m/s was observed in the present study, which allowed us to evaluate participant risk-taking. These lower velocities observed in the experiments may be in a range in which the foam hasn't "bottomed out" yet, compared to the higher velocities in the Kendall study. This bottoming out occurs when the stiffness of the foam increases abruptly once the foam has been compressed throughout the majority of its thickness (Subic, 2007). The differences may also relate to the type of foam used in each study.

Additionally, evidence was found that the design of shoulder pads also has a role on the dynamics of the head impact, as signified by the higher accelerations and rotational velocities experienced when players delivered checks with Pad A compared to Pad B in the straight-on impact configuration. Once again, looking at impacts above 60g, eighteen out of those twenty-seven impacts were delivered with Pad A, and the ten highest impacts were all delivered with Pad A as well. The two pad designs used in the present study have very different surface profiles as well as material properties in the shoulder cap. While Pad A is lighter in construction, its hard polyethylene plastic shoulder cap is covered in synthetic leather and minimal foam. The pad also has a less anatomical fitting shoulder cap and generally had- a larger area of its shoulder cap being contacted when a check was delivered with it. Pad B, on the other hand, is heavier in construction (886 grams versus 711 grams) and has multiple layers of different density foam beneath a plastic outer cap. The multi-density foam already built on to the shoulder cap of Pad B could have provided additional attenuation along with the moulded foam caps making it more protective than Pad A. The pad is also more moulded to the shoulder joint, and players generally had less of the cap in contact with the head when delivering a hit. These properties could explain the differences observed between shoulder pad conditions. Further examination is required to better understand the role of shoulder pad stiffness, geometry, curvature, and the location of the plastic component within the foam and membrane materials, in influencing head accelerations in ice hockey, in order to design an optimal shoulder pad that minimizes impact severity to the head of a player.

## 4.2. Player Aggression due to Shoulder Pad Design

The study's second aim was to investigate whether player aggression (as reflected by shoulder velocity at impact) was affected by shoulder padding, and to determine if players were more likely to hit harder if they felt more protected. Participants generally deemed Pad B to be more comfortable to deliver a check with, as illustrated by the results in the post-study questionnaire. However, shoulder impact velocity was not statistically different between Pad A and B in either impact configuration. This might reflect that players tried to hit as hard as they could regardless of what shoulder pad they were wearing. Following the conclusion of the experiment, several participants commented that they could “feel it” in their shoulder more when delivering a check with Pad A. Several participants noted that they did not feel at risk for incurring any type of shoulder injury despite feeling more discomfort when hitting with Pad A. Players also commented that they wanted to hit as hard as they could regardless of the type of shoulder padding they were wearing. Additionally, the accelerometer and gyroscope results showed that players delivered harder impacts with Pad A, despite no significant difference in contact velocity. This is once again likely due to the design of Pad A, which has less foam and is less formed to the shoulder joint, than Pad B.

It was also found that adding a foam cap onto shoulder padding had no effect on aggressiveness as reflected by shoulder impact velocity. Results from the post-study questionnaire showed that participants varied in their opinions on how the presence of the foam cap affected their comfort level in delivering checks. While some players indicated that they felt more comfortable in delivering checks with the foam cap, others felt less comfortable. This suggests that the perceived effect of foam caps was variable, (perhaps based on how much foam is on the shoulder pad the player typically uses), but ultimately did not cause players to change their impact velocity.

## 4.3. Limitations

While the present research provides valuable insight on how shoulder padding can be used to mitigate head injuries in ice hockey, there are some inherent limitations in this research. The head and neck of the checking dummy was originally designed to be used in kickboxing and is made entirely of foam rubber, with no hard “skeletal” elements. However, the foam was stiff enough to not buckle, and the rotational stiffness was controlled via an overhead spring. Players also commented that the dummy provided a realistic simulation of an opposing player. When asked to rate how checking the dummy compared to checking a player in a real game on a scale from one (completely different) to ten (completely similar), participants gave the experimental paradigm a median rating of seven. A metal cage was also attached on to the helmet of the dummy, which was the site of impact by the shoulder. A cage is required in all minor hockey leagues as well as in the NCAA (Biasca, Wirth & Tegner, 2002).

Another limitation is that measurements of head linear acceleration and rotational velocity were obtained from sensors located inside the crown of the dummy's helmet, and therefore do not reflect the true acceleration and velocity experienced by the center of the dummy's head. This is a common limitation in sensor-based research and groups have shown the importance of keeping sensor location constant, which was ensured for all subjects (Campbell et al., 2015).

Additionally, our method for obtaining shoulder impact velocity only considered movement in the sagittal plane (where the high-speed camera was oriented), in which it was assumed that most of the resultant impact velocity occurred. It was found that players generally moved towards the dummy in the "x" direction to deliver the check, and raised their shoulder in the "y" direction to make contact with the dummy head. Movement in the "z" axis (such as rotation of the shoulder inwards to the video view prior to impact) was not captured.

The experimental conditions were not blinded to either the participants nor the researchers. This could have unconsciously biased participants in how they hit the checking dummy. Additionally, participants had a short time to become familiar with each shoulder pad condition. It is possible that with more time players would acclimatize their behavior to each type of shoulder pad as they became more comfortable with its use. Players were also running, rather than skating when delivering checks, which they would be doing in their naturalistic environment. While there is no reason for believing that similar results would not occur while skating, future studies might validate this assumption.

## 4.4. Future Directions and Public Messages

These findings, coupled with those of Kendall et al. (2014) support the benefit of having 2cm thick foam pad on top of shoulder caps. The foam did not cause players to hit with an increased velocity, and lowered indices of the risk of head injury. Padding the shoulder area of jerseys with lightweight embedded foam may have similar effects and is potentially another avenue of research. Additionally, future studies should investigate how the mass and stiffness of the body-checking dummy affects impact severity. Other common impact types should also be modeled and a more realistic head/neck unit could provide insight into concussion injuries caused by “whiplash”.

The findings of this study support the value of a rule change by Hockey Canada to add a specification for shoulder pads that already exists for elbow padding; declaring that all shoulder padding that does “not have a soft protective covering of sponge, rubber or a similar material at least 1.27 cm (1/2in.) thick shall be considered dangerous equipment”. While NHL Pro-Stock shoulder padding are mandated to have a half inch of padding over its hard components, there are commercially available shoulder pads in retail stores that do not meet these guidelines and are used by amateur players in a variety of different leagues. Thus, an improved effort should be made to provide more knowledge and information to amateur and recreational hockey players regarding the importance of safe shoulder pad selection. Additionally, amateur and recreation leagues should also look to ensure that there is sufficient enforcement of proper equipment use in their leagues. Further work should also focus on further identifying the optimal geometry and material properties of shoulder pads. Specifically, future laboratory studies are required to gain an understanding regarding the ideal construction of padding that incorporates rigid (i.e. plastic) and soft (foam) structures.  Our findings comparing Pad A and Pad B show that despite no significant differences in contact velocity, one pad induced greater measures of head impact severity. It is therefore important going forward to evaluate a greater variety of shoulder pad models with the ultimate goal to create shoulder pad manufacturer guidelines and standards across all levels of play.

When taking into account these design considerations, it is also important to ensure that the amount of protection that the shoulder pad offers to the shoulder itself is not decreased to an amount that would cause an increased risk of incurring injury. Further research should be done testing novel designs that have minimal or no plastic in them before they are deployed to players to ensure that they still provide acceptable protection to the shoulder. Additionally, as illustrated above, it is important for researchers to take an integrated approach when developing sport injury prevention solutions. Not only is it important to test how novel designs perform biomechanically, but it is also important to observe how players change their behavior in response to a change in protective equipment. One solution for preventing concussions would be for players to lower their aggressiveness. However, due to the competitive nature of the game, players often do not adjust their behaviour following rule changes, and aggressive play is often encouraged and enjoyed by fans (Jones, Ferguson, & Stewart, 1993; Paul, 2003; Donaldson et al., 2013). Thus, passive forms of protection are valuable.

## 4.5. Conclusions

With the growing concern regarding the high number of concussion injuries that occur in ice hockey, it is important for researchers to tackle the problem from a variety of different directions in order for change to take place. While many groups attempt to prevent concussion injuries by looking to improve helmets, the role that another piece of equipment has on concussion injuries is often neglected: shoulder padding. The most common mechanism of concussion in elite play is direct contact to the head by the shoulder (Hutchison, Comper, Meeuwisse, & Echemendia, 2013b). The present study found that modifying the properties of shoulder padding (via the use of foam cap overlays), results in lower peak linear accelerations and rotational velocities, two indices of risk for concussion.

By utilizing an experimental paradigm where hockey players delivered shoulder checks to the head of a custom-made body-checking dummy, the present study evaluated both biomechanically and behaviorally, the effect shoulder pad design has on head injuries in a naturalistic setting. The results obtained from this study show evidence that shoulder pad design plays an important role in the dynamic impact response during a shoulder-to-head check in ice hockey. Thus, future research should focus on optimal design of shoulder padding with regard to both geometry and materials in order to design equipment that is both protective to the checking player's shoulder joint, as well as protective to the head of the player receiving the hit. By doing so, shoulder padding can work in tandem with other interventions such as improved helmet design, knowledge training, and rule changes, to reduce the number of concussions suffered by hockey players of all ages.

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