A Kinematic Rating System for Evaluating Helmet Performance

by

Adrian Wikarna

B.A.Sc., Simon Fraser University, 2017

Thesis Submitted in Partial Fulfillment of the
Requirements for the Degree of
Master of Applied Science

in the
School of Mechatronic Systems Engineering
Faculty of Applied Sciences

SIMON FRASER UNIVERSITY
Fall 2019

Approval

Name:	**Adrian Wikarna**
Degree:	**Master of Applied Science**
Title:	**A Kinematic Rating System for Evaluating Helmet Performance**
Examining Committee:	**Chair:** **Siamak Arzanpour** Associate Professor
	Farid Golnaraghi Senior Supervisor Professor
	Gary Wang Co-Supervisor Professor
	Daniel Abram Supervisor Adjunct Professor
	Stephen N. Robinovitch Internal Examiner Professor Biomedical Physiology and Kinesiology
Date Defended/Approved:	December 5, 2019

Abstract

Adopting a helmet has been very helpful in reducing the risk of head injury in activities with a high risk of impact to the head. However, the main focus of most helmet standards is protecting the head against skull fracture through a pass or fail criterion that only measures the linear acceleration of the head during impact. Yet, it is known that most impacts result in both linear and rotational acceleration to the head.

A pass-or-fail criterion does not inform the consumers how well a helmet performs. In recent years, Virginia Tech Summation of Tests for the Analysis of Risk (STAR) rating system was introduced to provide more insight into a helmet performance. The STAR rating system quantifies the risk of concussion based on the linear and rotational performance of a helmet. However, the science behind concussion is not fully understood, and in addition to helmet performance, the risk of concussion is closely related to other factors such as age, sex, genetic, the direction of an impact, and previous head trauma. The STAR rating also does not include all crucial factors in assessing a helmet performance, and therefore, it may not provide an accurate performance or risk of injury assessment for a given helmet.

In this work, a Kinematic Rating System (KRS) was developed to evaluate helmet performance based on how well a helmet reduces crucial factors such as linear acceleration, rotational acceleration, and rotational velocity. KRS is an effective tool that provides an accurate assessment of the performance of a helmet compared to when the head is not protected by a helmet. KRS requires the helmet of interest to be tested against a 45º anvil at 6.5 m/s impact speed. Various football, hockey, and cycling helmets were tested according to the KRS, and the results were compared with the STAR rating system. In some cases, the performance reported by the STAR rating system were found to have significant discrepancies with the results obtained by the KRS. This is because the STAR rating system does not consider all the crucial factors while evaluating a helmet, such as the magnitude and duration of the acceleration pulse.

Keywords: helmet, testing, evaluation, performance, head injury, concussion

Dedication

For my beloved *papa*, I dedicate this work.

For you have sent me here in the first place.

I made it.

For without your hard-work, love, and sacrifice this would never happen.

Thanks for everything.

Acknowledgements

First and foremost, I would like to express my deepest gratitude to my **God** for by His Grace, I am strengthened and able to do the work required to complete my thesis. Secondly, I would like to thank my **family**: *mama, papa, mami, papi*, for their never-ending support and love.

I would like to express my special gratitude to **Dr. Daniel Abram** for introducing me to this research field in late 2015 and providing me with opportunities. His passion, dedication, and hard work have inspired me. His constant support, guidance, expertise, has brought me into this research and helped me to prepare this thesis. I am thankful to have such a mentor and friend.

I am indebted to my respected academic supervisors, **Dr. Farid Golnaraghi** and **Dr. Gary Wang**, for their invaluable help in preparing this thesis and willingness to provide any necessary support.

My work would not have been finished on time without the help of the **co-op students** at the Head Injury Prevention Lab. I would like to thank *Firda Wijaya, Mikaila Hamilton, Po-Han Huang, Yizhi Qian, Amber Lu, Julien Gibbons, Terry Zhu, Joseph Zhu, Andrew Chan, Argyle Rivera, Ho-Jin Chang, Edmond Chow, Taqdeer Cheema, Gurdas Hothi, Austen Ware, Trence Barbarona,* and *Shawn Liu* for their hard work.

Last but not least, to all my **friends and relatives**, including but not limited to, my beloved grandma, Alvina Amelia, Erin Adela, Sheryl Yatasi, Gerry Karnadi, Dan Maringka and Ben Sundah for their constant encouragement, love, and support.

I humbly extend my gratitude to all persons who support me in their own ways in this work.

Table of Contents

List of Tables

List of Figures

List of Acronyms

ANSI	American National Standards Institute
ASTM	American Society for Testing and Materials
ATD	Anthropomorphic Test Dummy
BrIC	Brain Injury Criterion
BS	British Standard
CEN	the European Committee for Standardization
CPSC	Consumer Product Safety Commision
CSA	Canadian Standards Association
CTE	chronic traumatic encephalopathy
DAQ	Data Acquisition
DOT	Department of Transportation
ECE	Economic Commission for Europe
GAMBIT	Generalized Model for Brain Injury Threshold
HIC	Head Injury Criterion
HIP	Head Impact Power
ISO	International Organization for Standardization
KLC	Kleiven Linear Combination
KRS	Kinematic Rating System
MEP	Modular Elastomer Programmer
MIPS	Milti-directional Impact Protection System
mTBI	Mild Traumatic Brain Injury
NFL	National Football League
NFLPA	National Football League Player Association

NHL	National Hockey League
NOCSAE	National Operating Committee on Standards for Athletic Equipment
PRHIC	Power Rotational Head Injury Criterion
PVC	polyvinyl chloride
RIC	Rotational Injury Criterion
SFU	Simon Fraser University
SHARP	Safety Helmet Assessment and Rating Program
SI	Severity Index
STAR	Summation of Tests for the Analysis of Risk
TBI	Traumatic Brain injury
WSTC	Wayne State Tolerance Curve

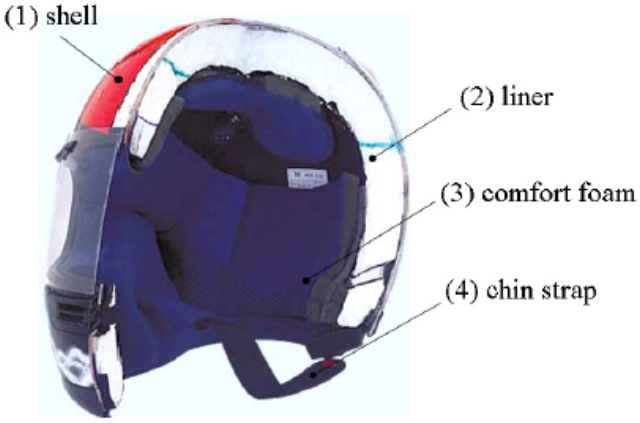
Chapter 1.

Introduction

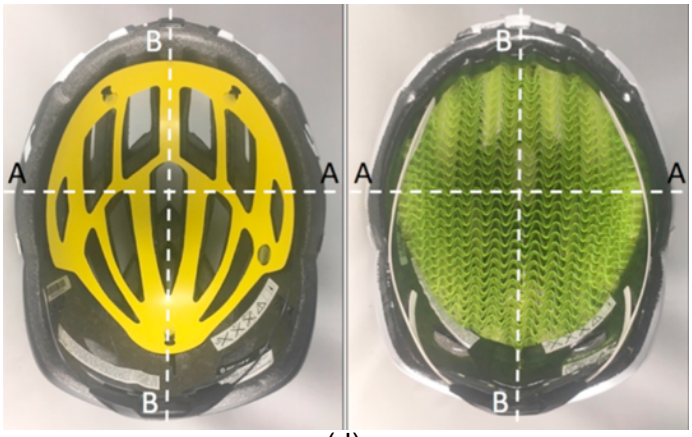
Humans are used to observing what is evident at first. As time goes by, humans become more curious as they start to see and learn beyond what is visible. After receiving an impact to the head, some injuries such as cuts and bleeding, swelling, skull fractures, and loss of consciousness may occur. All the injuries mentioned above were thought to be caused by a direct blow to the head (compression force). Ancient helmets were used to protect the user against the compression force. Early helmets were first invented and used for military purposes (combat) by Sumerians in Mesopotamia around 4500 years ago (Sone, et al., 2016). Ancient helmets were typically made of leather, bronze, or iron (Figure 1.1a). Until the 20th century, helmet design and its purposes had not changed much in terms of structure and the way it was designed. Advancements were made in terms of safety and comfort to better protect the user against modern military technology such as penetration and blast (Figure 1.1b) (Blackman, et al., 2007). Helmet manufacturers started making helmets from solid steel shell, Kevlar, and a combination of synthetic and non-synthetic polymers and fibers. Modern civilian helmets (Figure 1.1c and 1.1d) such as cycling and sports started to be developed when the focus shifted from protection against lethal impacts or penetration toward the more frequent but milder impacts. Modern civilian helmets are typically made from synthetic polymer foam and fibers.

(a)

(b)



(c)



(d)

Figure 1.1: **(a) Ancient helmets (Sone, et al., 2016), (b) Modern military helmet (Blackman, et al., 2007), (c) Modern motorcycle helmet (Petersen, 2018), (d) Modern cycling helmets with additional technologies(Bliven, et al., 2019)**

As humans gain more knowledge about the head and brain anatomy, they started to observe the less visible damage caused by impacts to the head, such as damage to the brain, which is the underlying factor of Traumatic Brain Injury (TBI).

In sports, 80% of all TBI is in the form of mild Traumatic Brain Injury (mTBI) and concussion (Noble & Hesdorffer, 2013). Sports-related concussion made up to 20% of the total number of brain injuries that occur each year in the United States (Brolinson, et al., 2006). The word concussion is derived from a Latin word *concussus* meaning "to shake violently" (Cantu, 1996) and has been known to humanity since the early ages. Concussion's symptoms may include, but not limited to, alteration of consciousness, disturbance of vision, physical impairment, and imbalance (Ommaya & Gennarelli, 1974). In the past, it was thought that a direct blow to the head caused a concussion. Recent studies have shown that the sharp rotation of the head is more likely to cause a concussion (rotational force) (Kleiven, 2013; Gennarelli, et al., 1972; Holburn, et al., 1943). However, even until the present day, the complexity and mechanism involved in a concussion are yet to be fully understood.

Helmets are effective in protecting the user against head injuries (McIntosh, et al., 2011; Cripton, et al., 2014; Breedlove, et al., 2016; Whyte, et al., 2019). However, some research studies suggested that improvements were needed in order for helmets to be effective in protecting against mild Traumatic Brain Injury (mTBI) or concussion as helmets are mainly designed to prevent skull fracture (Whyte, et al., 2019; Sone, et al., 2016). Modern helmets are designed, tested, and certified according to specific standards. For contact sports such as American football (subsequently referred to as football), helmets are tested and certified according to a standard imposed by National Operating Committee on Standards for Athletic Equipment (NOCSAE) (NOCSAE, DOC, 2019). Most ice hockey (subsequently referred to as hockey) helmets are tested and certified according to standards imposed by Canadian Standards Association (CSA), American Society for Testing and Materials (ASTM), and NOCSAE (NOCSAE, DOC, 2016; CSA, 2015; ASTM International, 2015). The most common standard certification for cycling helmets is the Consumer Product Safety Commission (CPSC) standard (Consumer Product Safety Commission, 1998).

From the time that cycling, football, and hockey helmet standards and certifications were introduced (circa 1961-1973), the main impact test scenario

performed on a helmet has been based on falls using a guided drop test rig (Whyte, et al., 2019). The standards require the linear acceleration of the humanoid headform to be measured and compared to a certain threshold. For example, NOCSAE standard performance specification for newly manufactured football helmets passing criteria requires the Severity Index (calculated by integrating the linear acceleration over a period) not to exceed 1200 (NOCSAE, DOC, 2019). The threshold was experimentally determined based on the occurrence of skull fractures in human cadaver (Whyte, et al., 2019). Nowadays, people are more concerned about mTBI rather than skull fractures. Also, most of the head impacts in real-life events are oblique or angled (Chinn, et al., 2001). The impact (contact) force has two components: normal and tangential, therefore causing linear and rotational acceleration of the head (Adanty, et al., 2019). Skull fractures are known to be caused by linear acceleration while concussion, diffuse axonal injury (DAI), and subdural hematoma (SDH) are known to be caused mainly by the rotational acceleration of the head (Kleiven, 2013; Holburn, et al., 1943). As a result, researchers have been testing helmets for rotational acceleration (Chinn, et al., 2001; Adanty, et al., 2019; Rowson, et al., 2015; Bland, et al., 2018; Ebrahimi, et al., 2015; Halldin, 2015). Halldin et al. have also proposed an oblique impact testing standard for cycling helmets (Halldin, 2015) as an improvement of the widely used CPSC standard. Furthermore, NOCSAE will start adopting a new rotational-based helmet test method in late 2019 (NOCSAE, DOC, 2019).

Another limitation of most helmet standards is the use a pass/fail criteria, which cannot provide information about how good a helmet performs under specific impact conditions. Since helmet performance is directly related to its design and can be different from one to another (Breedlove, et al., 2016), consumers are faced with a wide range of selections without knowing how well one helmet is compared to another. A tool that can give information regarding a helmet's performance was needed. This tool can be used to help customers in making an informed decision when purchasing a helmet.

In 2011, Virginia Tech has come up with a Summation of Tests for the Analysis of Risk (STAR) rating system where helmets were rated based on its performance under their specific testing methodology (Rowson, et al., 2015; Bland, et al., 2018; Rowson & Duma, 2011). Although this is a good step forward, the STAR rating system comes with some limitations. One of them is being reliant on a concussion risk curve. According to some research studies, concussion risk varies from person-to-person depending on age,

sex, history of previous head trauma, and genetics. (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004).

Virginia Tech has come up with a STAR rating system where they rate helmets based on its performance under their specific testing methodology (Rowson & Duma, 2011; Rowson, et al., 2015; Bland, et al., 2018). However, the STAR rating suffers from several shortcomings and limitations, such as, but not limited to, not taking into account the tangential component of the impact force, not measuring all crucial variables during the characterization of impact severity, and relying on an injury predicting tool. Therefore, in this work, a new rating system is proposed to provide a tool for evaluating and comparing the performance of a helmet with other helmets in its category.

Helmet performance is directly related to its design and can vary from one helmet to another (Breedlove, et al., 2016). Therefore, consumers are faced with a wide range of selections without knowing how well one helmet is compared to another. The objective of this work is to develop an evaluation method for helmets. The method evaluates helmets based purely on its kinematic response since any tool for predicting the risk of concussion may not be accurate. Evaluating the kinematic response of a helmet can provide a transparent metric on how a helmet reduces the forces that can result in head injury. The Kinematic Rating System (KRS) requires performing multiple impact tests on different areas of a helmet while considering the impact exposure risk based on the available statistics. Preliminary testing and validation of the KRS were performed on different models of football, hockey, and cycling helmets available in the market.

1.1. Thesis Structure

The thesis is composed of five Chapters. The first chapter is the introduction that decribes the motivation and objective of this work. The second chapter is the literature review, which provides the background information needed in this work. Chapter 3 describes the development of the KRS for football, hockey, and cycling helmets. Chapter 4 talks about the validation of KRS by means of testing football, hockey, and cycling helmets according to the defined methodology, and the helmet impact test result will be described and discussed. Lastly, Chapter 5 concludes the study and provides future works and recommendations.

Chapter 2.

Literature Review / Background

2.1. Epidemiology of Head Injuries and Concussion in Sports

A concussion is one of the well-known injuries in sports and has been predominantly underreported in the past decades (Parizek & Ferraro, 2015). A study reported that the concussion rate in collegial sports increased by 7% each year over 16 years period between 1988 and 2004 (Hootman, et al., 2007). Hockey and football are popular helmet-required sports while cycling is used as a form of recreational sports and transportation. There are 12 million recreational cyclists in Canada alone (Ramage-Morin, 2017). Also, there are over 1 million registered hockey players and over 5 million football players in North America (Gough, 2018; Gough, 2018; Lock, 2019).

In North America, head injuries due to sports are problematic due to the economic and social impact it may bring to society. Statistics showed that an estimated of 1.8 million people in North America sustain TBI each year (Brain Injury Canada, n.d.; Conte Jaswal, 2017; Faul, et al., 2010) and each TBI can cost up to $400,000 per case for the medical system of the country (Brain Injury Society of Toronto, 2018). A 5% reduction in the number of TBI can be translated to a reduction of 8,250 cases, which is a significant saving of the taxpayer money and reduction of social and emotional burden in our society. Therefore, it is paramount to provide a better head and brain protection system. Any possible advancement needs to be considered and executed.

2.1.1. Football

A study in 2007 showed that football-related concussions made up to 55% of all concussions reported in collegiate sports (Hootman, et al., 2007). Between 1996-2007, the average concussion rate per football game was 0.40 (Yengo-Kahn, et al., 2015). In 2016, a study reported 0.61 concussions per game in the NFL (Nathanson, et al., 2016), a rise from the previous years. Besides, the latest data collected by the National Football League (NFL) showed that there was a fluctuation in concussion incidence between 2012 to 2018, with a high of 281 incidences in 2017 (Sprecher, 2019).

2.1.2. Hockey

Concussions account for 2-14% of all (professional and non-professional) hockey injuries and 15-30% of all hockey head injuries (Izraelski, 2014). The rate of concussion incidence in hockey is one of the highest in contact sports and team sports (Cantu, 1996; Hootman, et al., 2007; Cusimano, et al., 2013). The number of concussion incidence in the National Hockey League (NHL) is on the rise (Parizek & Ferraro, 2015; Wennberg & Tator, 2003; Benson, et al., 2011; Izraelski, 2014). A study using data from published media reports found that the number of concussion incidence in 2002 is more than triple of the previous decades (Wennberg & Tator, 2003). More recent studies also showed the same increasing trend in the average number of concussions per season (Benson, et al., 2011; Hutchison, et al., 2015; Izraelski, 2014).

2.1.3. Cycling

In the UK, the number of cyclists has increased since 2005 (Forbes, et al., 2017). In Canada, between 1994 and 2014, the number of cyclists has increased while the number of cycling fatalities fluctuates with an average of 74 person per year (Ramage-Morin, 2017). The increase in the number of cycling fatalities can be caused by the advancement of the diagnosing tool. Studies have shown that head and facial trauma is one of the most common types of injuries amongst cyclists (Melo, et al., 2014). A research study has shown that concussion accounts for 1.3% to 9.1% of all cycling-related injuries (Elliott, et al., 2019). Head injury has also been shown to be a key factor of mortality and morbidity in cycling-related accidents (Forbes, et al., 2017; Martin, et al., 2018). Between 2007 and 2012, out of 586 cyclists who died while cycling, 46% of death was attributed to head injury (Martin, et al., 2018).

2.2. Mechanism of Head Injuries and Concussion

Translational head kinematics or motion can induce injuries such as skull fracture, epidural hematoma, and contusions (Kleiven, 2013). Translational motion of the head is caused by impact forces that are in-line with the head's center of gravity. In other words, the head will be accelerated in a straight line (McLean & Anderson, 1997). In real-world conditions, this type of impact is rare (Kleiven, 2013; Chinn, et al., 2001; Willinger, et al., 2015; McIntosh, et al., 2011). On the other hand, rotational head motion

of the head can induce concussion, diffuse axonal injury (DAI), contusions, subdural hematoma, and intracerebral hematomas (Kleiven, 2013; Holburn, et al., 1943). Rotational motion of the head is caused by impact forces that are off-centered (oblique) with the head's center of gravity. Due to the obliqueness of the impact force, the impact force can be broken down into two orthogonal components. As illustrated in Figure 2.1, the two orthogonal components are the normal force (*N*) and tangential force (*f*). Both forces, with their corresponding moment arms d and r, may induce rotational motion of the head. In most cases, reducing both forces can be advantageous.

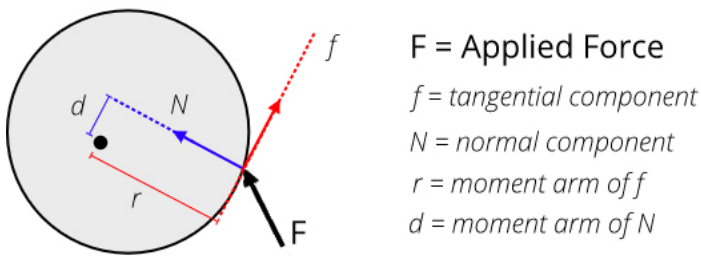


Figure 2.1: Simplified drawing of an oblique impact.

Kinematic parameters such as acceleration and velocity of the head during impact are closely linked to brain injuries (Holburn, et al., 1943; Hodgson & Thomas, 1971; Rowson & Duma, 2013; Gennarelli, et al., 1972; Gurdjian, et al., 1966; Ommaya, et al., 1967; Versace, 1971; Willinger & Baumgartner, 2003; Post & Hoshizaki, 2012; Broglio, et al., 2010). With an increase in acceleration, velocity, or duration of an impact, the risk of injury also rises (Hoshizaki, et al., 2017; Hodgson & Thomas, 1971; Holburn, et al., 1943). Studies have shown that any single kinematic parameters of the head may not be enough to predict mTBI (Whyte, et al., 2019; Chinn, et al., 2001; Greenwald, et al., 2008).

Peak linear acceleration was thought to be a good parameter to describe the severity of a traumatic brain injury (TBI) (Rowson & Duma, 2013; Post & Hoshizaki, 2015; Pellman, et al., 2003; Zhang, et al., 2004; Broglio, et al., 2010). However, it was found that the human brain is more sensitive to rotational acceleration than linear acceleration (Ommaya & Gennarelli, 1974; Gennarelli, et al., 1972; Holburn, et al., 1943; Post & Hoshizaki, 2015; Gennarelli, et al., 1987; Moritz, 1943; Post, et al., 2013). The human brain has minimal resistance to changes in shape compared with the resistance it offers to changes in size (McElhaney, et al., 1976). Therefore, the brain tissue is more

susceptible to strain-induced by shearing or rotational motion than translational motion (McElhaney, et al., 1976; Meaney & Smith, 2011).

According to some research studies, brain tissue response, such as strain, can be a good indicator of TBI such as concussion (Post & Hoshizaki, 2015; Meaney & Smith, 2011; Whyte, et al., 2019; King, et al., 2003). Other research studies suggest that rotational acceleration and velocity have a significant correlation to brain strain response (Kleiven, 2007; Rowson, et al., 2012; Ji, et al., 2014; Patton, et al., 2012). Therefore, the rotational kinematics of the head can be a helpful tool in diagnosing a concussion.

A head injury caused by the rotational motion of the head is directional dependant (Gennarelli, et al., 1987; Takhounts, et al., 2008). The brain strain patterns within the brain are considerably different depending on whether the rotation of the head is applied in the coronal, horizontal, or sagittal plane (Gennarelli, et al., 1987; Meaney & Smith, 2011). In other words, the injurious effect of an impact may be different depending on the location and the direction of the applied force.

2.2.1. Human tolerance or threshold to concussion

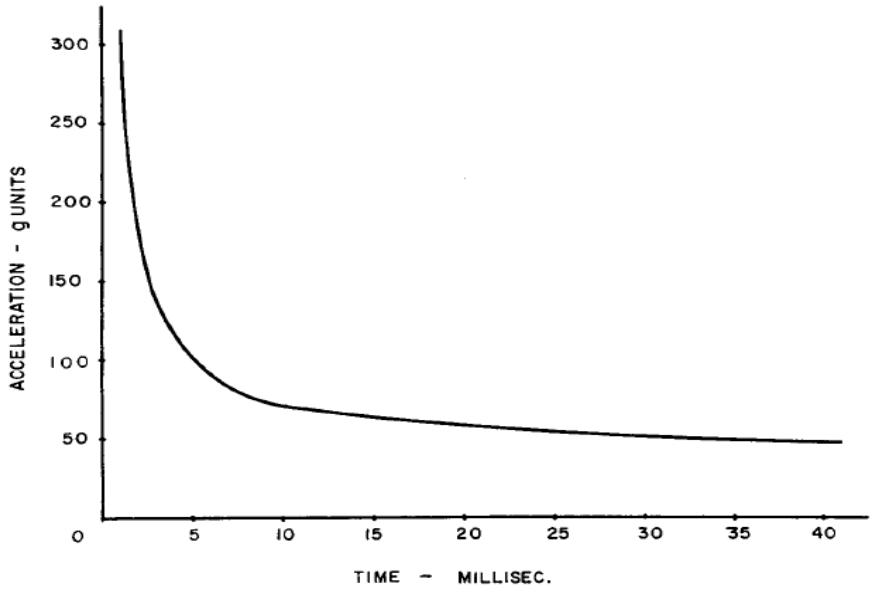
Several research studies have been done to determine the kinematic threshold a human can withstand before suffering a concussion. Early research was conducted on animal subjects (Ommaya & Gennarelli, 1974; Ommaya, et al., 1967; Denny-Brown & Russell, 1941; Gennarelli, et al., 1982). Different levels of accelerations were applied to the animal subject, and the response was studied. The animal study results were then scaled to obtain the human response to the levels of accelerations (Ommaya, et al., 1967). Some researchers resorted to in-situ measurement where they equipped athletes with measurement device on their head/helmets to measure head kinematics during concussive and non-concussive impacts (Rowson, et al., 2009; Funk, et al., 2007; Guskiewicz, et al., 2007; Greenwald, et al., 2008; Broglio, et al., 2010; Crisco, et al., 2012; Beckwith, et al., 2013) while others have conducted numerical simulations and laboratory reconstructions of impacts created by analyzing video footages (Pellman, et al., 2003; Zhang, et al., 2004; Takhounts, et al., 2008; Patton, et al., 2012; Chinn, et al., 2001). Table 2.1 summarizes some research studies that tried to determine the human concussion threshold.

Table 2.1: Summary of major research studies on human concussion threshold

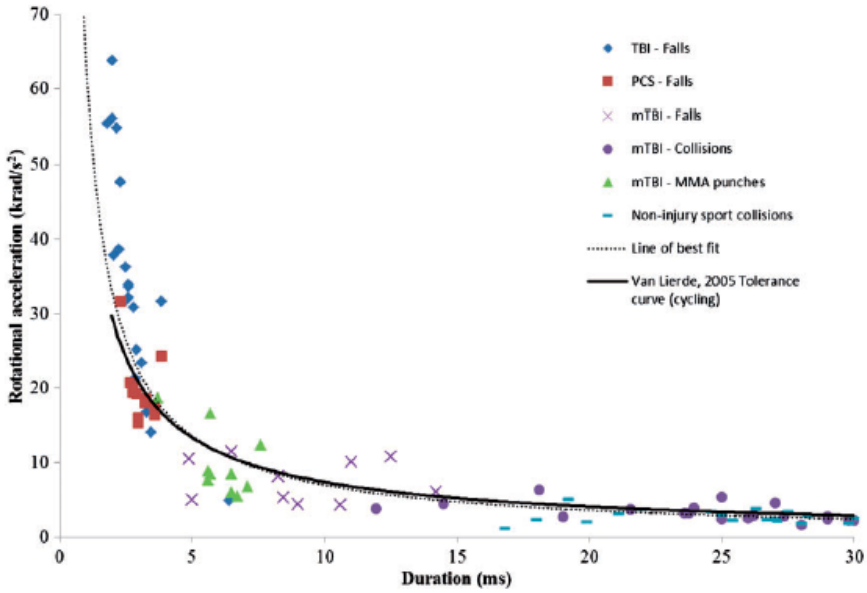
Researcher	Concussion risk	Lin. Acc. (g)	Rot. Acc. (rad/s^2)	Sample Size	Note
Pellman (Pellman, et al., 2003)	0%	20	2,000	31 cases of impacts; 25 concussion cases	Laboratory reconstruction from video analysis.
	80%	98 ± 28	-		
Zhang (Zhang, et al., 2004)	25%	66	4,600	24 concussion cases	FEA based on real-world accidents.
	50%	82	5,900		
	80%	106	7,900		
Funk (Funk, et al., 2007)	10%	165	9,386	27,319 impacts; 4 concussion cases	Football in-vivo measurement.
Guskiewicz (Guskiewicz, et al., 2007)	100%	102.8	-	104,714 impacts; 11 concussion cases	Football in-vivo measurement.
	Suggested threshold	60-80	-		
Greenwald (Greenwald, et al., 2008)	75%	96	7,235	289,916 impacts; 17 concussion cases	Football in-vivo measurement.
Broglio (Broglio, et al., 2010)	Suggested threshold	96.1	5,582.3	54,247 impacts; 13 concussion cases	Football in-vivo measurement.
Crisco (Crisco, et al., 2012)	0%	20.3	1,392	184,358 impacts;	Football in-vivo measurement.

Rowson (Rowson, et al., 2012)	100%	-	5,022	300,977 head impacts; 57 concussion cases	Football in-vivo measurement.
Beckwith (Beckwith, et al., 2013)	100%	112.1 ± 35.4	4,253 ± 2,287	161,732 impacts; 105 concussion cases	Football in-vivo measurement.
	0%	20.7	848		

Other research studies have shown that the human brain can tolerate a different level of accelerations (peak rotational and linear acceleration) depending on the impact duration (Gurdjian, et al., 1966; Ommaya, et al., 1967). For example, human brain can withstand a short burst (less than 5 ms duration) of relatively high magnitude acceleration pulse (Gurdjian, et al., 1966; Ommaya, et al., 1967; Hoshizaki, et al., 2017; Hitosugi, et al., 2014; O'Riodain, et al., 2003). Figure 2.2 shows the proposed injury tolerance curve for linear (Figure 2.2a) and rotational acceleration (Figure 2.2b). Therefore, it is crucial to have a sense of the impact duration in determining the severity of an impact.



(a)



(b)

Figure 2.2: Injury Tolerance Curve for (a) linear (Gurdjian, et al., 1966) and (b) rotational acceleration (Hoshizaki, et al., 2017).

Also, studies have shown difficulties in establishing a concussion threshold for humans (Guskiewicz, et al., 2007). According to research studies, concussion risk varies from person-to-person depending on age, sex, previous head trauma, genetics, location and direction of impact, whether or not it was anticipated (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004; Gennarelli, et al., 1987; Gennarelli, et al., 1982).

2.2.2. Repeated Concussion

In contact sports, such as football and hockey, it is more common for athletes to suffer from multiple impacts to the head over the course of the years that they are involved in the sport. A research study on collegiate football players showed that people with a history of three or more concussions are at higher risk of suffering another concussion in future activities or games (Guskiewicz, et al., 2003; Giza & Hovda, 2001). Also, if a second concussion happens while the first one is still healing, the victim may develop a chronic symptomps or otherwise known as the Second Impact Syndrome (SIS) (Bey & Ostick, 2009). Facts about repeated concussion have been used as the basis for the return-to-play guidelines in contact sports (McLeod, et al., 2017).

Furthermore, athletes, such as football and hockey players, who receive repetitive injury may develop chronic traumatic encephalopathy (CTE) (McKee, et al., 2009). An autopsy of 202 football players showed a high proportion of subjects had developed CTE (Mez, et al., 2017). Another study showed a link between the history of repeated concussion and depression in later age (Guskiewicz, et al., 2007). A study in 2018 found that athletes who began playing football before the age of 12 developed symptoms of CTE thirteen years earlier compared to those who started playing later (Alosco, et al., 2018).

2.3. Head Injury Assessment Criteria/Tools

Studies have shown that any single kinematic parameters of the head may not be enough to predict the occurrence of mTBI (Whyte, et al., 2019; Chinn, et al., 2001; Greenwald, et al., 2008). Some studies came up with a function involving single or multiple kinematic parameters to the head to better estimate the occurrence of mTBI. In this section, some of the most commonly used head injury assessment tools, and their limitations will be described.

Gadd Severity Index (SI)

In 1966, Gadd came up with a Severity Index (SI), a method of assessing a pulse waveform in its entirety (Gadd, 1966). SI considers both the intensity (peak) and the time duration of the acceleration pulse. The SI can be seen in (2.1).

$$\boldsymbol{SI = \int_0^T (a(t))^n dt} \tag{2.1}$$

Where T is the period of the pulse, a is the magnitude of the linear acceleration, and n is the weighing factor greater than one. There are many possibilities for choosing the weighing factor. The most commonly used weighing factor is $n = 2.5$ and was chosen by taking into account a probabilistic injury curve developed by Wayne State University (Gadd, 1966; Versace, 1971). NOCSAE standards for football and hockey helmet utilize SI value as one of the pass/fail criteria. A study in 1971 (Versace, 1971) detailed some limitations of SI. One of the limitations was that the same value of SI was

obtained for different acceleration pulse of the same characteristics (time duration and average acceleration).

Head Injury Criterion (HIC)

In 1972, the National Highway Traffic Safety Administration (NHTSA) introduced the HIC (McHenry, 2004) in response to a study detailing the limitations of SI by Versace (Versace, 1971). HIC was initially developed for assessing the potential of head injury in the automotive crash test dummy (McHenry, 2004). The HIC equation is shown in (2.2).

$$\boldsymbol{HIC} = \max_{t_1,t_2}\left\{\left[\frac{1}{t_2 - t_1}\int_{t_1}^{t_2} a(t)dt\right]^{2.5}\right\} \tag{2.2}$$

HIC takes into account the resultant translational or linear acceleration of the dummy. The power value of 2.5 was determined based on the tolerance curve containing injurious and non-injurious data sets. The time interval, $t_2 - t_1$, was initially selected to be 36 ms. However, studies in the past decades suggested that the time interval should be 15 ms (McHenry, 2004; Prasad & Mertz, 1985).

Generalized Model for Brain Injury Threshold (GAMBIT)

GAMBIT was first introduced in 1985 by a Canadian researcher, James Newman (Newman, 1966). GAMBIT combines the translational and rotational kinematics of the head. The GAMBIT equation can be seen in (2.3).

$$G(t) = \left[\left(\frac{a(t)}{a_{critical}}\right)^n + \left(\frac{\alpha(t)}{\alpha_{critical}}\right)^m\right]^{1/S} \tag{2.3}$$

where $\boldsymbol{a}$ and $\boldsymbol{\alpha}$ are the instantaneous values of linear and rotational acceleration and $\boldsymbol{n}$, $\boldsymbol{m}$, and $\boldsymbol{S}$ are empirical constants. The critical values of the linear and rotational acceleration represent the critical values of brain injury. GAMBIT provides a boundary between injurious and non-injurious motions. Depending on the choice of empirical constant, the boundary shape can be changed depending on available injury data. However, one limitation of GAMBIT is the use of critical values that represents the critical value of brain injury. One study showed difficulty in determining injury threshold for human (Guskiewicz, et al., 2007).

Head Impact Power (HIP)

In 2000, the same researcher that introduced GAMBIT proposed a new head injury assessment function called HIP (Newman, et al., 2000). The HIP linearly combines linear and rotational accelerations and linear and rotational velocities. HIP also takes into account the mass and inertia of the head, as seen in (2.4).

$$\boldsymbol{HIP = \sum m\, a \cdot v + \sum I\, \alpha \cdot \omega} \tag{2.4}$$

The HIP function was developed based on a set of mild traumatic brain injury (concussion) data. The function correlates better with the concussion risk data than other existing head injury assessment functions at the time (Newman, et al., 2000). However, studies have shown difficulty in accurately diagnosing and reporting concussion incidence (Rowson & Duma, 2013; McCrea, et al., 2004; Delaney, et al., 2002). Therefore, the reliability of the concussion data may not be adequate.

Kleiven's Linear Combination (KLC)

In 2007, KLC was introduced. KLC linearly combines both HIC and rotational velocity, as seen in (2.5). The combination showed a good correlation with the distortion strain of the brain and has been validated by computer simulation of NFL concussion data (Kleiven, 2007).

$$\boldsymbol{KLC = 0.004718 \cdot \Delta\omega + 0.000224 \cdot HIC} \tag{2.5}$$

Where $\Delta\omega$ is the peak change in rotational velocity and HIC is the Head Injury Criterion. It has been found by computer simulation that HIC can be used to predict the strain level in the brain for purely translational impulses of short duration. Also, the peak change in angular velocity showed the best correlation with the strain levels for purely rotational impulses (Kleiven, 2006). However, simulating a complex human brain is no easy task and comes with limitations as the human brain may vary from person-to-person (McHenry, 2004). In the KLC study, one limitation was the relatively low mesh density of the head model, which limits the possibility of geometrical detail of the brain (Kleiven, 2007).

Brain Injury Criterion (BrIC)

BrIC consists only of rotational velocity. The study using FE simulation found that rotational velocity correlates well with cumulative strain damage measure (CSDM) and maximum principal strain (MPS) in the brain (Takhounts, et al., 2011). Both parameters were used to determine injury severity. The same study found that rotational velocity is directionally dependent. Hence, BrIC was developed by combining the rotational velocity of the three axes, as seen in (2.6).

$$\boldsymbol{BrIC} = \sqrt{\left(\frac{\boldsymbol{\omega_x}}{\boldsymbol{\omega_{x,Critical}}}\right)^2 + \left(\frac{\boldsymbol{\omega_y}}{\boldsymbol{\omega_{y,Critical}}}\right)^2 + \left(\frac{\boldsymbol{\omega_z}}{\boldsymbol{\omega_{z,Critical}}}\right)^2} \tag{2.6}$$

The critical values of each rotational velocity were chosen to be 66.25 rad/s, 56.45 rad/s, and 42.87 rad/s, respectively, for x, y, and z-direction.

There are some limitations for the BrIC function. First, rotational velocity was chosen as the sole input of the BrIC function based on facts found by the FE study. The inputs are compared to a critical value based on human tolerance to injury. It is known that any FE simulations and injury predicting tool may not be accurate (Guskiewicz, et al., 2007). Another limitation is that the BrIC function does not take into account rotational acceleration. It is known that rotational acceleration is one of the critical factors in diagnosing a head injury (Holburn, et al., 1943; Hodgson & Thomas, 1971; Rowson & Duma, 2013; Gennarelli, et al., 1972; Gurdjian, et al., 1966; Ommaya, et al., 1967; Versace, 1971; Willinger & Baumgartner, 2003; Post & Hoshizaki, 2012; Broglio, et al., 2010).

Rotational Injury Criterion (RIC)

In 2012, a group of researchers came up with a prediction tool for concussion based on the rotational acceleration of the head during impact (Kimpara & Iwamoto, 2012). The RIC equation is similar to HIC, except the linear acceleration is substituted with rotational acceleration, as seen in (2.7).

$$\boldsymbol{RIC} = \max_{t_1,t_2}\left\{\left[\frac{\mathbf{1}}{\boldsymbol{t_2 - t_1}}\int_{t_1}^{t_2}\boldsymbol{\alpha(t)dt}\right]^{2.5}\right\} \tag{2.7}$$

The time duration, $\boldsymbol{t_2 - t_1}$, was chosen by the researchers to be 36 ms. However, the time duration is not strict, one can change it based on their own applications. The RIC was developed and validated with real-life injury data and FE simulations. Therefore, RIC exhibit the same limitations as other tools that were validated the same way.

Helmet assessment criteria were developed to better understand the severity of an impact with respect to head injury. Some of the critera, such as SI and HIC, have been used as a pass/fail criteria in helmet testing standards. However, research studies have shown that no single kinematic criterion can predict the occurene of mTBI (Whyte, et al., 2019; Chinn, et al., 2001; Greenwald, et al., 2008).

2.4. Helmet Design and Testing Standards

2.4.1. Review of Helmet Design and Efficacy

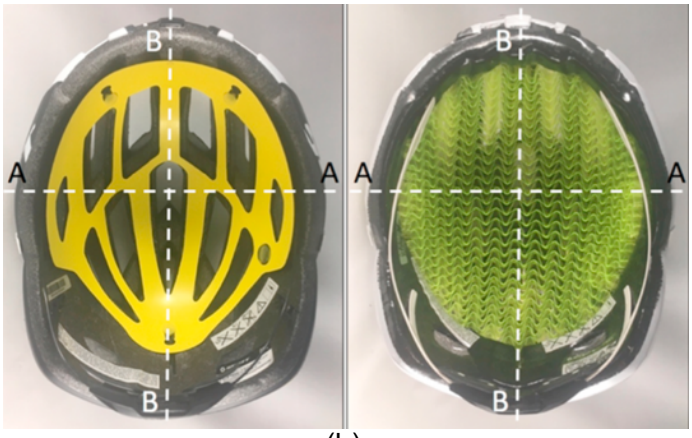
Helmet performance is directly related to its design and can be different from one to another (Breedlove, et al., 2016). Current helmet certification standards fail to provide consumers with information on how well one helmet performs compared to another. There have been lots of efforts in the past decades on evaluating helmet performance. Over the years, researchers have evaluated helmet performance both in-situ and in a lab or simulation. The results showed that by wearing a helmet, injury risk was reduced significantly (Bambach, et al., 2013; Benson, et al., 2009; Cripton, et al., 2014; Olivier & Creighton, 2017; McIntosh, et al., 2011; Bandte, et al., 2017; Attewell, et al., 2001; Yu, et al., 2011).

However, a study suggested that helmet's ability to reduce concussion risk is inconclusive (Benson, et al., 2009) while others suggested helmet is not effective in protecting against concussion (Sone, et al., 2016). One probable reason is the inadequate protection of a helmet against tangential force or rotational acceleration (King, et al., 2003), which has been known to be closely related to concussion (Kleiven, 2013; Holburn, et al., 1943). Moreover, the helmet's impact attenuating capability can also vary depending on the location and the direction of the applied force. There is still room for helmet protection level to be improved (Mills & Gilchrist, 2006; Mills & Gilchrist,

1991). A study found that wearing a bicycle helmet in one specific impact condition increases the rotational acceleration of the head during impact (McIntosh, et al., 2013).

In recent years, researchers have come up with new radical helmet design by adding another layer of material or fully or partially replacing the foams (Bliven, et al., 2019; VICIS Inc., 2019). Figure 2.3 shows some of the latest technologies in football and cycling helmets. VICIS ZERO1 helmet, in Figure 2.3a, uses a non-conventional structure, which consists of flexible columns. Multi-directional Impact Protection System (MIPS) technology, as seen in Figure 2.3b, is a thin plastic shell attached on the inner face of a helmet. MIPS facilitates relative movement of the head and the helmet during impact. Both technologies were developed to improve helmet linear and rotational acceleration performance.

(a)



(b)

Figure 2.3: (a) Modern football helmet (VICIS Inc., 2019) and (b) Modern cycling helmets with additional technologies (Bliven, et al., 2019)

2.5. Review of Available Helmet Impact Testing Methods in Helmet Certification Standards

Most helmets require to be certified by a standard organization. The standards differ based on the type of helmet and jurisdiction. Some of the governing bodies that are responsible for developing the available are the Canadian Standards Association (CSA), the U.S. Consumer Product Safety Commission (CPSC), the American National Standards Institute (ANSI), the American Society for Testing and Materials (ASTM), the National Operating Committee on Standards for Athletic Equipment (NOCSAE), the European Committee for Standardization (CEN), the International Organization for Standardization (ISO), and the British Standard (BS). For example, in North America, football helmets are tested and certified based on a standard developed by NOCSAE. Whyte summarized various testing certification standards in his research study (Whyte, et al., 2019).

The standards may contain various types of testing, such as impact attenuation tests and chinstrap strength tests. Most of the impact attenuation tests use a pass-fail criterion. A standard dummy headform (one or multiple sizes) may be used for testing. For example, as shown in Figure 2.4a, CPSC and CSA standard for bicycle helmet use rigid (ISO/DIS 6220 K1A magnesium) headform, while NOCSAE uses the Urethane NOCSAE headform (Figure 2.4b). In other helmet impact tests performed by researchers, Hybrid III head with and without the neck (can be NOCSAE neck or Hybrid III neck) were used (Whyte, et al., 2019). The kinematic parameters of the headform, such as linear and rotational acceleration, may be measured with the appropriate sensors and method. In almost all tests, the linear acceleration is measured. These measured values may be used directly or indirectly (to calculate another head injury assessment tool, for example) to determine the pass/fail of the helmet.

(a) (b)

Figure 2.4: (a) Rigid K1A Magnesium Headform (Cadex Inc., 2019) and (b) Urethane NOCSAE Headform (MacAlister, 2013)

Each headform has its advantages and disadvantages and, therefore, not interchangeable (Whyte, et al., 2019). For example, NOCSAE headform physical features are more anatomically accurate (MacAlister, 2013; Cobb, et al., 2015). On the other hand, the Hybrid III head impact response is more biofidelic. In addition, Hybrid III is commonly used in biomechanical and impact research (Chinn, et al., 2001; MacAlister, 2013). A direct comparison between Hybrid III and NOCSAE headform (tested without neck using a linear drop tower) showed that the NOCSAE headform linear response is similar to the Hybrid III headform while the rotational response of Hybrid III was higher (Bland, et al., 2018).

2.5.1. Football Helmet Test Methods

In North America, football helmets are tested and certified based on the NOCSAE standard. It is the only available standard after the ASTM F17 was withdrawn in 2017 (Whyte, et al., 2019). NOCSAE standard requires the impact attenuation test to be performed on a guided drop impact test rig against a Modular Elastomer Programmer (MEP) pad at up to 5.5 m/s impact speed (NOCSAE, DOC, 2019).

The impact attenuation test of the NOCSAE standard calls for a helmet to be tested at seven impact locations with three impact speeds (NOCSAE, DOC, 2019). The headform linear acceleration is measured, and the Severity Index (SI) is computed using ((2.8). To pass the test, the peak SI shall not exceed 1200 SI (equivalent to 215 g) on

any impacts and 300 SI on low-speed impacts (Whyte, et al., 2019; NOCSAE, DOC, 2019).

$$\boldsymbol{SI} = \int_{0}^{T} (\boldsymbol{a}(\boldsymbol{t}))^{2.5} \boldsymbol{dt} \tag{2.8}$$

In November 2019, NOCSAE will start requiring a rotational-based pneumatic ram test. The test uses the Urethane NOCSAE headform coupled with a Hybrid III neck. The headform-neck assembly is mounted on a linear bearing table, allowing two-axis rotation and single-axis translation. The test calls for a helmet to be tested at six different impact locations at one impact speed. The peak rotational acceleration shall not exceed 1200 SI and 6000 rad/s^2 (NOCSAE, DOC, 2019).

The implementation of the NOCSAE standard has been helpful to gauge helmet performance and provide a minimum requirement for helmets available in the market and consequently reduce the number of head injuries sustained by helmet users (Levy, et al., 2004). The certified helmets can provide a range of performance, which is directly related to their design and material (Breedlove, et al., 2016; Post, et al., 2013).

On the other hand, the NOCSAE standard has some limitations. One limitation is that the average impact speed for concussive cases in football is higher than the 5.5 m/s impact speed defined in the NOCSAE impact attenuation test (Pellman, et al., 2003; Viano, et al., 2006; Viano, 2005). Another limitation is that the NOCSAE standard is based on a pass/fail criterion and does not provide quantifiable information about the level of performance of the certified helmets. In addition, the threshold for the pass/fail criterion is similar to the critical value of the skull fracture and may not represent the threshold for human concussions (Hodgson & Thomas, 1971; Pellman, et al., 2003; Zhang, et al., 2004; Guskiewicz, et al., 2007; Blackman, et al., 2007).

Linear impactor tests such as pneumatic ram and pendulum test produce little tangential force (Willinger, et al., 2015). On the other hand, research has shown that tangential force is a crucial contributor to the rotational acceleration of the head (Finan, et al., 2008). Furthermore, the linear impactor test simulates helmet-to-helmet impact (Pellman, et al., 2016). It is known that injuries also can occur due to impact against playing surface, other body parts, and jersey (Naunheim, et al., 2002; Withnall, et al.,

2005; Rossi, et al., 2016) which are not simulated by the linear impactor test. Falls to the ground induced some of the most severe head responses (Pellman, et al., 2003).

The pneumatic ram test in NOCSAE standard uses NOCSAE headform coupled with Hybrid III neck. In some impact scenarios, the use of a neck can reverse the direction of the head rotational acceleration (Beusenberg, et al., 2001). Studies have shown that the Hybrid III neck is too stiff compared to human neck and the Hybrid III neck is designed only for flexion and extension and its behavior for other types of motion such as lateral is not known (Aare & Halldin, 2003; Bartsch, et al., 2012; Myers, et al., 1989; Svensson & Lovsund, 1992; Herbst, et al., 1998; Gwin, et al., 2009).). In some of the locations of impact proposed in the NOCSAE pneumatic ram test and STAR rating system (Tyson & Rowson, 2018; NOCSAE, DOC, 2019), the neck responds in the direction other than flexion or extension. Therefore, the results obtained from such impact scenarios may not be reliable.

A typical peak acceleration in football impact lasts for approximately 15 ms (Pellman, et al., 2003; Zhang, et al., 2004; Deck & Willinger, 2008). A study on helmeted headform suggested that the effect of the neck is small for the first 10 ms of the impact (Willinger, et al., 2015). An experimental study on human cadaver suggested a mechanical separation between head and neck for the first 9 ms of impact, resulting in the minimum influence of the neck on the head kinematics (Nightingale, et al., 1996). Another study on human subjects has shown that there is minimal activity of the neck's bilateral sternocleidomastoid (SCM) muscles for the first 50 ms or 80 ms depending on the impact being anticipated or not (Ono, et al., 2003). Similar study has shown that the SCM requires 13-14 ms to respond to mild impact, regardless of the impact was anticipated or not (Kuramochi, et al., 2004). In addition, there is an atlanto-occipital neutral zone where the neck joint can have motion in the range of 10 deg without inducing any force that can affect the kinematics of the head (Ivancic, 2014; Camacho, et al., 1997). Since the Hybrid III neck does not provide a human-like response (Bartsch, et al., 2012; Myers, et al., 1989; Svensson & Lovsund, 1992; Herbst, et al., 1998; Gwin, et al., 2009) and there is an atlanto-occipital neutral zone, therefore eliminating the neck can result in more accurate response for the first 10-15 ms of an impact.

2.5.2. Hockey Helmet Test Methods

There are four ice hockey certification standards available: CSA Z262.1 (2015), NOCSAE ND030 (2016), ASTM F1045 (2015), and ISO 10256-2 (2016) (Whyte, et al., 2019). The ASTM (1045-99) and CSA (Z262.1-M90) standards are most widely used to test and certify hockey helmets in North America. Available hockey helmet test methods do not evaluate the helmet performance in reducing rotational kinematics of the head. The testing standards call for a helmet to be tested on a vertical drop test rig (similar to the NOCSAE standard for football helmet). ASTM and CSA recommend 4.5 m/s impact speed while NOCSAE standard for hockey helmet recommends 5.5 m/s impact speed (Whyte, et al., 2019; NOCSAE, DOC, 2016), both against MEP pad on up to seven impact locations. In the CSA standard, the peak linear acceleration should not exceed 275 g, lower than the 300 g threshold set by the ASTM standard (Whyte, et al., 2019).

Similar to that of the NOCSAE standard for football helmets, a pass/fail criterion does not provide a metric about how well a helmet performs compared to other available helmets. In addition, the threshold for the pass/fail criterion is for the critical value of the skull fracture and does not represent the threshold for concussions (Hodgson & Thomas, 1971; Pellman, et al., 2003; Zhang, et al., 2004; Guskiewicz, et al., 2007; Blackman, et al., 2007).

2.5.3. Cycling Helmet Test Methods

There are multiple cycling helmet certification standards such as CPSC 16 CFT Part 1203 (1998), CAN/CSA D113.2-M89 (2009), and EN 1078 (2014) (Whyte, et al., 2019; Connor, et al., 2016). The CPSC standard is one of the most widely used to test cycling helmets in the market. The test standard call for a helmet to be tested on a vertical drop test rig (similar to the NOCSAE standard for football helmet) using the ISO/DIS 6220 K1A Magnesium Headform. CPSC recommends for up to 6.2 m/s impact speed (Consumer Product Safety Commission, 1998), both against different kinds of steel anvil (flat, curbstone, hemispherical) on different impact locations at different speeds (4.8 and 6.2 m/s). The passing criterion is that the peak linear acceleration should not exceed 300 g (Consumer Product Safety Commission, 1998)

Similar to that of the NOCSAE standard for football helmets, a pass/fail criterion does not provide a metric about how well a helmet performs compared to other available helmets. In addition, the threshold for the pass/fail criterion is for the critical value of the skull fracture and does not represent the threshold for concussions (Hodgson & Thomas, 1971; Pellman, et al., 2003; Zhang, et al., 2004; Guskiewicz, et al., 2007; Blackman, et al., 2007).

Recently, the European Committee for Standardization (CEN) proposed a new certification standard EN13087-11 (Halldin, 2015). The newly proposed method requires a helmet to be tested at three different impact locations (Figure 2.5). The impact speed proposed is 6.5 m/s, slightly higher than the CPSC standard, to account for more severe impacts. The impact opponent is a hard surface angled at 45°. However, the proposed standard has not decided on the pass/fail criterion or which kinematic parameters or head injury assessment tool to measure.

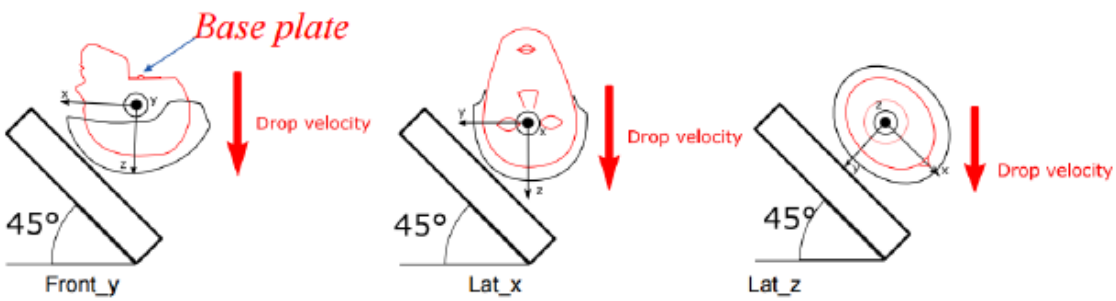


Figure 2.5: Impact locations of EN13087-11 standard

Furthermore, to address the issue of the pass/fail criteria possesses, some researchers came up with rating systems. The purpose of the rating systems was to help customers distinguish helmet performances. Some of the rating systems also test the helmet based on a more realistic impact conditions, taking into account helmet protection against milder impacts that can lead to concussion.

2.6. Review of Available Helmet Rating Methods

2.6.1. SHARP Rating

In 2008, the UK's Department of Transport introduced a Safety Helmet Assessment and Rating Program (SHARP). The SHARP safety rating is intended for assessing motorcycle helmets. Motorcycle helmets are certified to either BS 6658 or UN ECE Regulation 22/05 (BS 6658:1985, 1985; UN ECE Regulation 22/05, 2002). The SHARP testing protocol requires the subject helmet to be tested under a series of linear and oblique impact tests. The oblique impact test measures the friction between the helmet and the anvil, not the head kinematics. The head kinematics are found through a simplified mathematical model of an impact. Mills has highly criticized this process due to its mechanistic model (Mills, 2010).

2.6.2. Folksam

Folksam, an insurance company from Sweden, has been actively testing cycling, ski, and equestrian helmets and rate them based on the performance compared with each other. The test methods consist of shock absorption tests at two locations and oblique impact tests at three locations (Folksam, 2019). For the shock absorption test, the helmet is dropped from a 1.5 m height onto a horizontal anvil (as per European standard EN1078 2012). The linear acceleration of the head cannot exceed 250 g. On the other hand, for the oblique impact test, the Hybrid III dummy head without the neck was used, and the helmeted headform is dropped onto a 45° inclined anvil with the impact speed of 6.3 m/s.

For each test, the head kinematic response was measured and taken as an input to a computer simulation to determine the corresponding concussion risk. It is known that establishing a concussion threshold for the human is a difficult task (Guskiewicz, et al., 2007). According to some studies, concussion risk varies from person-to-person depending on age, sex, previous head trauma, genetics, location and direction of impact, whether or not it was anticipated (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004; Gennarelli, et al., 1987; Gennarelli, et al., 1982). Therefore, any tool for predicting the risk of concussion may not be accurate (Willinger, et al., 2015).

The helmets were rated by comparing the test results relative to each other. The results of each test are then further statistically processed using a weighted average scheme, as seen in (2.9) where T1 and T2 are the relative results in shock absorption tests, and T3 - T5 are the relative results in the oblique impact tests.

$$\boldsymbol{Rating}\ (\%) = \frac{\frac{\boldsymbol{T_1 + T_2}}{\boldsymbol{2}} + \frac{\boldsymbol{2 \times (T_3 + T_4 + T_5)}}{\boldsymbol{3}}}{\boldsymbol{3}} \tag{2.9}$$

Looking at (2.9), the oblique (angled anvil) impact test results were assumed to be twice as important as horizontal (flat anvil) impacts. However, there is no scientific evidence that can support this assumption.

2.6.3. NFLPA Rating

National Football League Player Association (NFLPA) have been performing impact test on football helmets at Biokinetics Lab (Ottawa, Canada) (NFLPA, 2019). All tested helmets have been certified by NOCSAE and were ranked based on its impact performance. The tests were performed using a pneumatic ram test rig (NFLPA, 2019) and Hybrid III headform. The test methodology has the same limitations as the NOCSAE standard (as described in the previous section.)

2.6.4. Virginia Tech STAR

In 2011, Virginia Tech introduced the Summation of Tests for the Analysis of Risk (STAR) rating system for football helmet which is based on predicting the risk of concussion (Rowson, et al., 2015; Bland, et al., 2018; Rowson & Duma, 2011). The STAR rating system uses a pendulum impactor test rig and the Urethane NOCSAE headform coupled with a Hybrid III neck. Similar to the pneumatic ram test used in the NOCSAE standard, the headform-neck assembly is mounted on a linear bearing table, allowing two-axis rotation. Helmets are tested on four impact locations (as shown in Figure 2.6) at three different speeds (3.0 m/s, 4.6 m/s, and 6.1 m/s). The pendulum impactor test method has similar limitations to the NOCSAE pneumatic ram test.

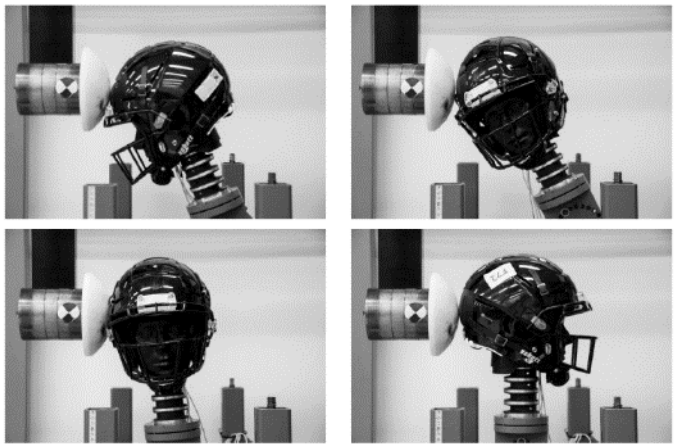


Figure 2.6: Impact locations in STAR test methodology for football helmet (Rowson & Duma, 2011)

The STAR rating system uses the STAR equation (2.10) that relates the impact exposure and the concussion risk estimation at every impact scenario (Tyson & Rowson, 2018).

$$\boldsymbol{STAR} = \sum_{L=1}^{4}\sum_{V=1}^{3} \boldsymbol{E(L,V) * R(a,\alpha)} \tag{2.10}$$

Where $L, V, a,$ and α are impact location (front, front boss, side, and back), impact speed (3.0, 4.6, and 6.1 m/s), peak linear acceleration, and peak rotational acceleration. The impact exposure weighting (E) is a function of impact location and impact speed. The concussion risk (R) is a function of linear and rotational acceleration, as seen in (2.11).

$$\boldsymbol{R(a,\alpha)} = \frac{\mathbf{1}}{\mathbf{1} + \boldsymbol{e}^{-(-10.2+0.0433*a+0.000873*\alpha-0.00000092*a\alpha)}} \tag{2.11}$$

In subsequent years, the Hockey STAR evaluation system was developed and is similar to the adult football STAR evaluation method. The impact location slightly differs from the ones from football STAR (Figure 2.7). However, the impact exposure (E) function has different values as the function utilizes data collected from in-situ experiments on hockey players (Rowson, et al., 2015).

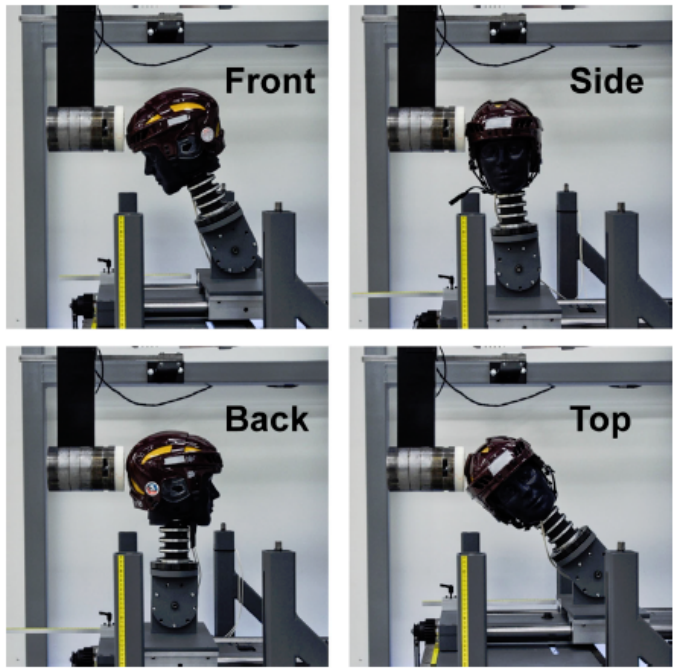


Figure 2.7: Impact locations in STAR test methodology for hockey helmet (Rowson, et al., 2015)

The bicycle STAR (Bland, et al., 2018) evaluation system was also introduced and calls for the cycling helmets to be tested under an oblique impact drop tower at two impact speeds (4.8 and 7.3 m/s) and six impact locations (Figure 2.8). The STAR equation for cycling helmet can be seen in (2.12).

$$\boldsymbol{STAR} = \sum_{L=1}^{6} \sum_{V=1}^{2} \boldsymbol{E(L,V)} * \boldsymbol{R(a,\omega)} \tag{2.12}$$

Where $L, V, a,$ and ω are impact location, impact speed, linear acceleration, and rotational velocity. The impact exposure weighting (E) is a function of impact location and impact speed. The concussion risk (R) is a function of linear acceleration and rotational velocity. The concussion risk can be computed using (2.13).

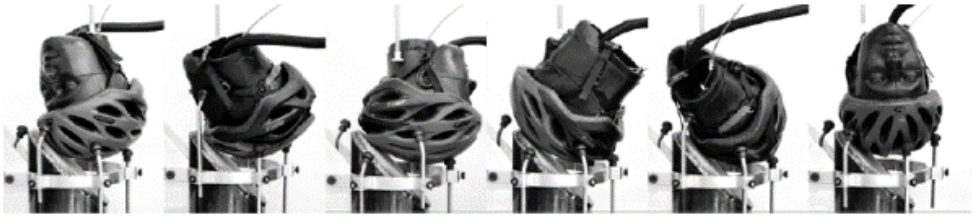


Figure 2.8: Impact locations in STAR test methodology for cycling helmet (Bland, et al., 2018)

$$R(a,\omega) = \frac{1}{1 + e^{-(-10.2+0.0433*a+0.19686*\omega-0.0002075*a\omega)}} \tag{2.13}$$

The STAR value predicts the concussion incidence of one player over a season (Rowson & Duma, 2011). However, the definition of STAR value for bicycle helmet slightly differs. The STAR value for bicycle helmet describes concussion incidence value per helmet model. The lower the STAR value, the better the helmet is. The STAR value is further scaled into five different categories.

Although this is a good step forward, STAR comes with some limitations. The science behind concussion is not clear (Sharp & Jenkins, 2015), and any tool for predicting the risk of concussion may not be accurate (Willinger, et al., 2015). In addition, studies have shown difficulties in establishing a concussion threshold for humans (Guskiewicz, et al., 2007). According to research studies, concussion risk varies from person-to-person depending on age, sex, previous head trauma, genetics, location and direction of impact, whether or not it was anticipated (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004; Gennarelli, et al., 1987; Gennarelli, et al., 1982). The concussion risk functions in (2.11) and (2.13) was developed based on combined head impact data of instrumented high school and collegiate football players over a period of time (37 concussive and 62,974 sub-concussive impacts) (Rowson & Duma, 2013). The head impact data were recorded with the Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH). Such concussion data collection methodology is known to have multiple issues. Firstly, studies have shown difficulty in accurately diagnosing and reporting concussion incidence (Rowson & Duma, 2013; McCrea, et al., 2004; Delaney, et al., 2002). Secondly, the HIT system is known to have poor accuracy in measuring the head impact, which affects any injury risk functions created based on HIT datasets (Jadischke, et al., 2013).

Compared to other suggested concussion threshold, the concussion risk function used in STAR underestimates the concussion risk. As seen in Figure 2.9, an impact with a combination of 3000 rad/s^2 and 100 g kinematics would result in less than 5% chance of injury while other research studies suggested over 55% risk of concussion at 100 g (Zhang, et al., 2004; Blackman, et al., 2007). This comparison highlights the difficulties of establishing concussion risk for human due to high variations (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004; Gennarelli, et al., 1987; Gennarelli, et al., 1982). Also, as suggested by different studies, the injury risk is higher when the linear and rotational acceleration are applied together (King, et al., 2003; Gennarelli, et al., 1972; Ueno & Melvin, 1995).

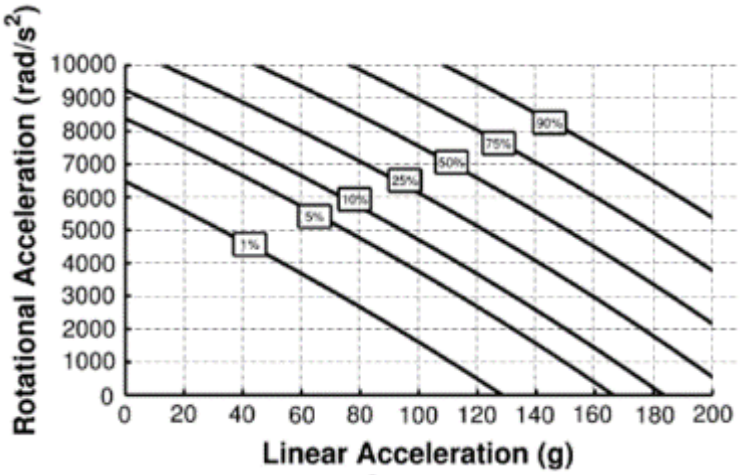


Figure 2.9: Injury risk in terms of a combination of linear and rotational acceleration (Rowson & Duma, 2013)

Table 2.2: Comparison of concussion risk assessment based on linear and rotational acceleration from different research studies (King, et al., 2003; Zhang, et al., 2004; Rowson & Duma, 2013)

	King (King, et al., 2003)			**Zhang** (Zhang, et al., 2004)		
Lin. Acc. (g)	57	79	98	66	82	106
Rot. Acc. (krad/s^2)	4.4	5.8	7.1	4.6	5.9	7.9
Concussion Risk	**25%**	**50%**	**75%**	**25%**	**50%**	**80%**
Concussion Risk (VTech)	**2%**	**12%**	**55%**	**3%**	**12%**	**59%**

Past research studies have shown that the human brain can tolerate a different level of accelerations (peak rotational and linear acceleration) depending on the impact duration (Gurdjian, et al., 1966; Ommaya, et al., 1967; Hoshizaki, et al., 2017; Schuller & Niemeyer, 2005). In other words, both the peak value and the duration of the acceleration pulse determine the severity of the impact. STAR rating system for football and hockey helmet only measures peak linear acceleration and peak rotational acceleration.

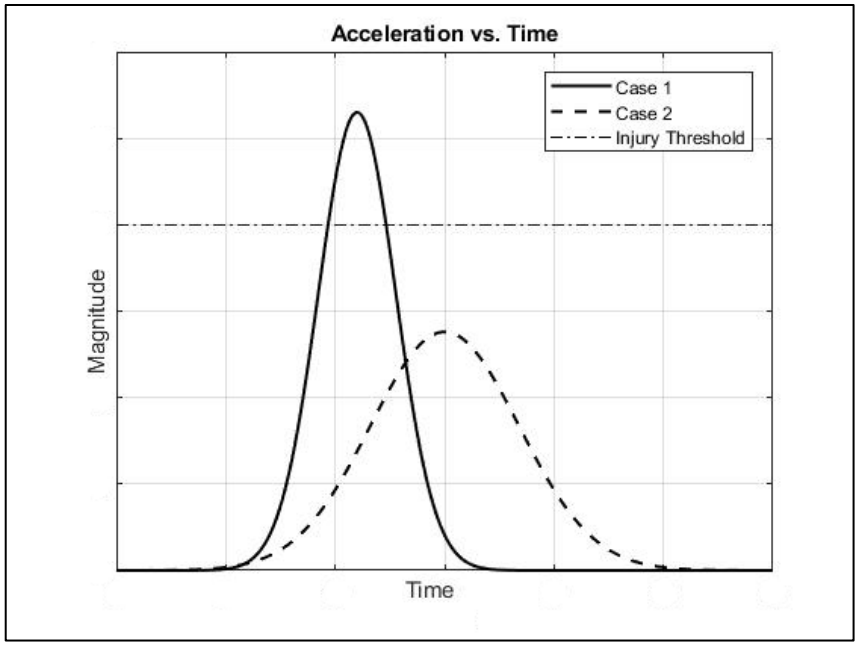


Figure 2.10: Illustration of injury severity information in two acceleration pulses

An illustration was made and can be seen in Figure 2.10. Case 1 and Case 2 are the acceleration graphs of two different impact scenarios. According to football STAR, considering only the peak acceleration value, the impact of Case 1 is more severe than Case 2. However, when considering the pulse duration, this may not be the case.

In contrast, the STAR rating system for cycling helmet measures the peak linear acceleration and peak change in rotational velocity. Measuring the peak change in rotational velocity does not provide clear information regarding the duration and magnitude of the acceleration pulse. For instance, assuming that the area under the curve for Case 1 and Case 2, shown in Figure 2.10, are the same, the rotational velocity reported for both cases would be the same. Yet, the two impact scenarios are inherently different and result in a different amount of risk of injury. Therefore, it is essential to consider both the peak value and the duration of the acceleration in assessing a helmet performance.

2.7. Summary

Most of the current helmet evaluation methods only monitor the linear acceleration of the head. However, the rotational acceleration of the head is known to be closely related to brain injuries such as concussions, diffuse axonal injury (DAI), contusions, subdural hematoma (SDH), and intracerebral hematomas (Kleiven, 2013). Moreover, most available testing methods utilize a pass/fail criterion, which fails to gauge how well a helmet passes the test compared to others.

Researchers have come up with helmet rating methods or systems such as the Virginia Tech STAR rating system (Rowson, et al., 2015; Bland, et al., 2018; Rowson & Duma, 2011). However, the STAR rating system also comes with some limitations. One of the limitations is that the STAR rating system relates the head kinematics with injury risk, which is proven to be difficult (Mollayeva, et al., 2018; Albrecht, et al., 2016; Iverson, et al., 2004; Gennarelli, et al., 1987; Gennarelli, et al., 1982).

The following chapters will describe the development of a Kinematic Rating System (KRS) for football, hockey, and cycling helmets. The KRS, aside from not predicting injury risk, considers all the crucial variables in determining impact severity, such as the magnitude and duration of the acceleration pulse.

Chapter 3.

Development of the Kinematic Rating System

3.1. Objective

The objective is to develop a Kinematic Rating System (KRS) for cycling, football, and hockey helmets. The KRS evaluates helmets based on its impact attenuation capability under different impact scenarios. The evaluation method involves performing multiple impact attenuation tests (with different impact scenarios) on a helmet. Multiple impact scenarios were determined based on the probability of each scenario happening in real life. In each impact scenario, the kinematics of the head with and without wearing a helmet was analyzed. Multiple sets of baseline values of the kinematic parameters (linear and rotational accelerations and rotational velocity) were determined by performing impact attenuation tests on the dummy headform without wearing a helmet. A general equation, namely the grade equation that takes the kinematics of the head as the inputs were developed. After performing impact attenuation tests under multiple impact scenarios, a final grade was calculated by combining multiple grades obtained from different impact scenarios. Preliminary testing and validation of the helmet evaluation method were performed on cycling, football, and hockey helmets.

3.2. Description of Experimental Testbed

Impact attenuation tests were performed on the oblique impact test rig that was developed and built at the Head Injury Prevention (HIP) Lab. The test rig is a patent-pending state-of-the-art apparatus[1]. As shown in Figure 3.1a, the rig consists of a rail, a carriage, and an anvil. A drop assembly, as shown in Figure 3.1b is attached to the carriage and consists of a horizontal suspension arm, a vertical bar, and a release mechanism. The helmeted headform is suspended on the release mechanism via cords. The cords were installed in different locations of the headform, allowing for different suspending configurations. Small hook-and-loop fasteners are placed on different

[1] D. Abram, F. Golnaraghi and G. Wang, "Suspension-based Impact System". Patent Patent Application No. 16/289140, 2019.

locations on the drop assembly as stabilizing points to preserve the suspending configuration. A gate is placed on the rail to trigger the release mechanism and disengage the cords and allow the helmeted headform to start its free-fall motion right before contacting the anvil. The drop assembly allows free-fall from a controlled height and the disconnection of the headform right before impact. Figure 3.1c shows the Suspend-X test rig that was built in the HIP lab.

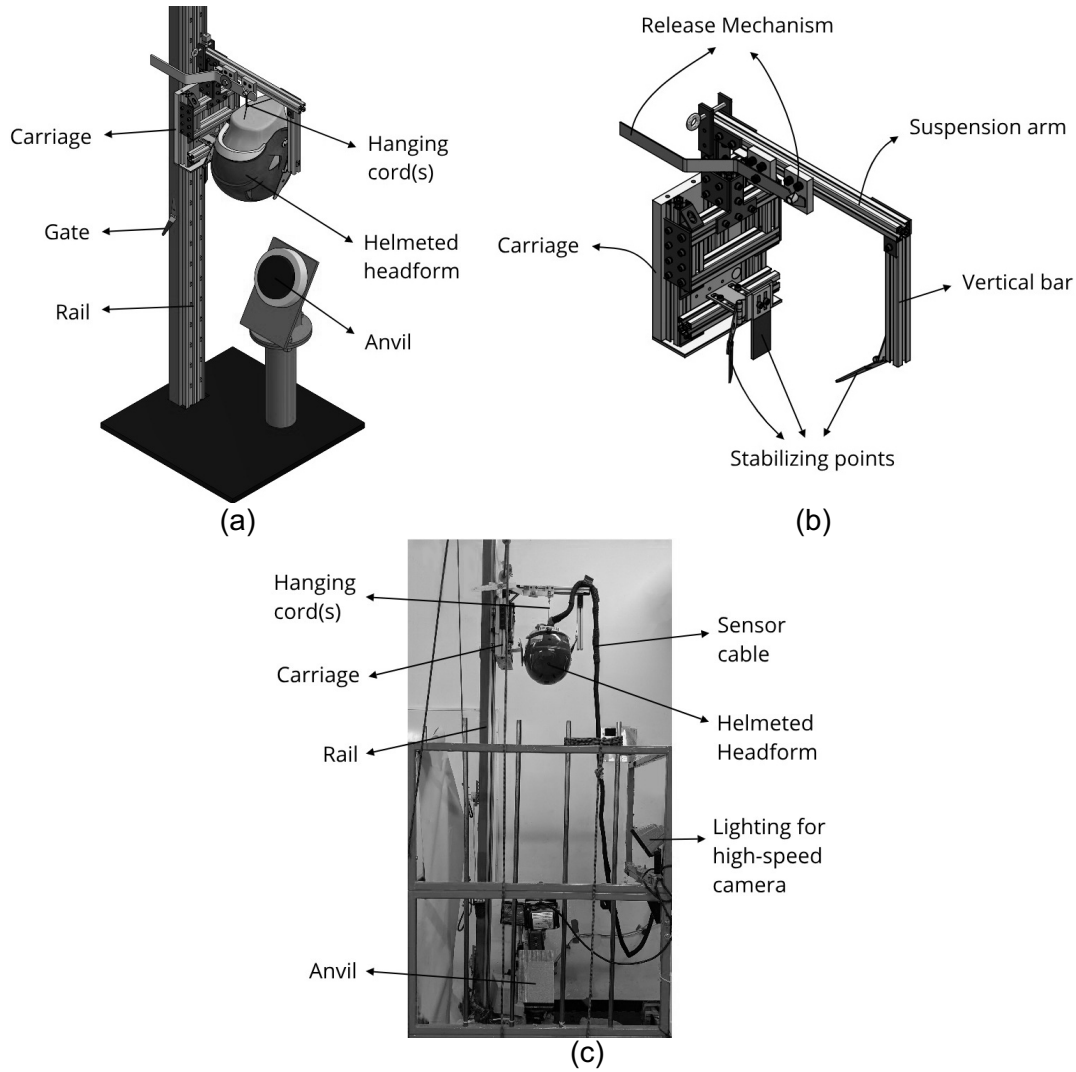


Figure 3.1: (a) CAD model of the Oblique Impact Test Rig at the HIP Lab and (b) the drop assembly, and (c) Oblique Impact Test Rig at the HIP Lab

The test rig is equipped with a standard neckless Hybrid III 50th percentile male dummy headform from Humanetics Innovative Solutions. Hybrid III is the most extensively used ATD in biomechanical research (Whyte, et al., 2019). The dummy headform is covered with a 10 mm thick vinyl skin (Figure 3.2). A study showed that the

surface condition of the dummy headform could affect the impact condition (Ebrahimi, et al., 2015; Trotta, et al., 2018).

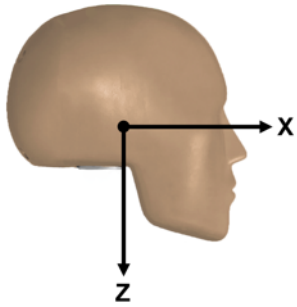


Figure 3.2: Hybrid III dummy headform with the reference measurement axis.

The dummy headform is suspended on the test rig above the anvil. Table 3.1 and Figure 3.3a-c details the default (0, 0, 0) orientation of the dummy headform. Impact scenarios are defined by specifying the dummy headform pre-impact orientation and impact velocity.

Table 3.1: Default orientation of the dummy headform

Orientation	Z	Y	X
Default	0	0	0

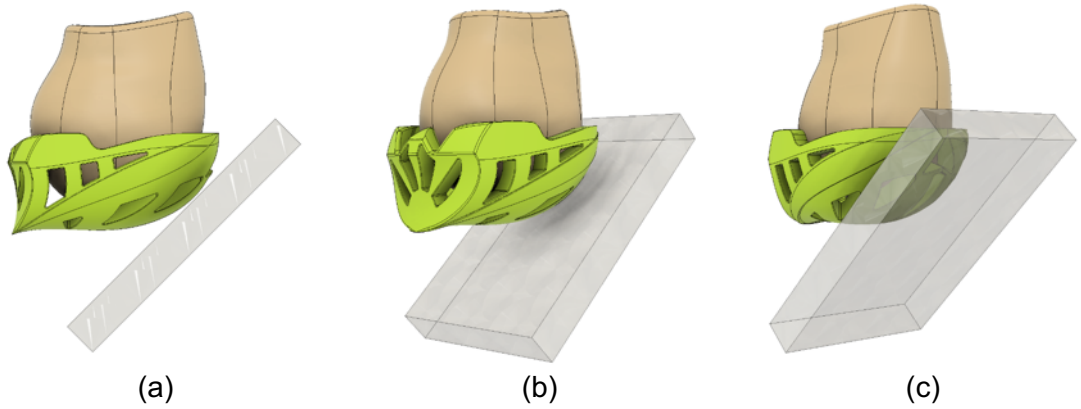
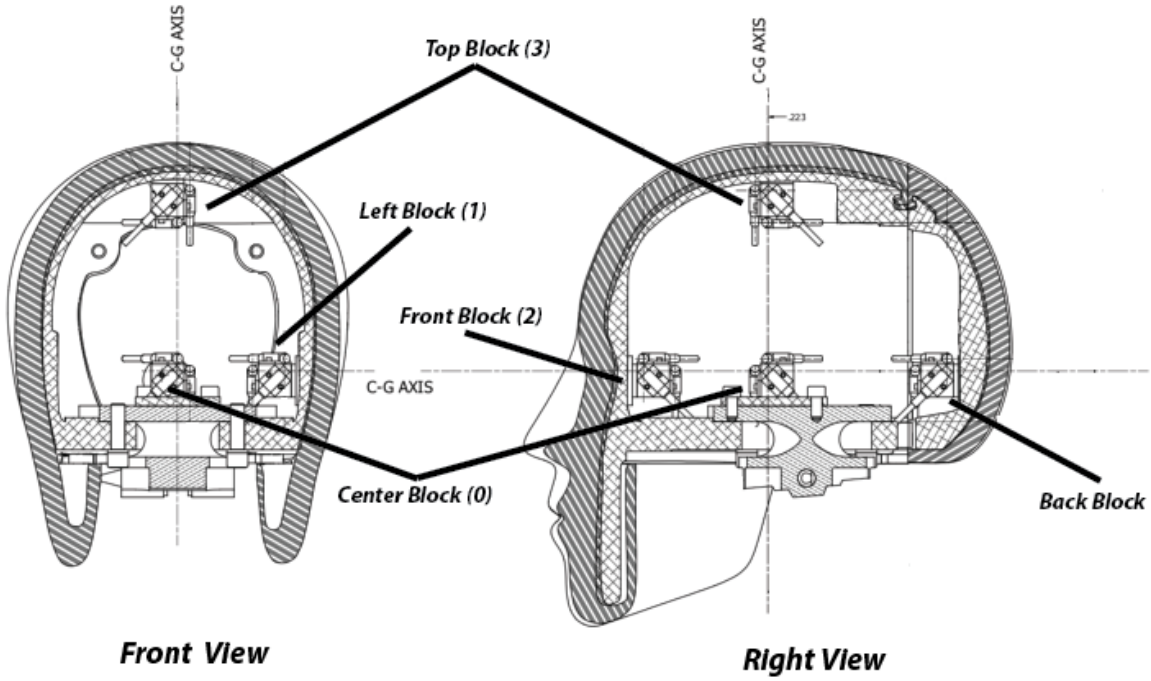
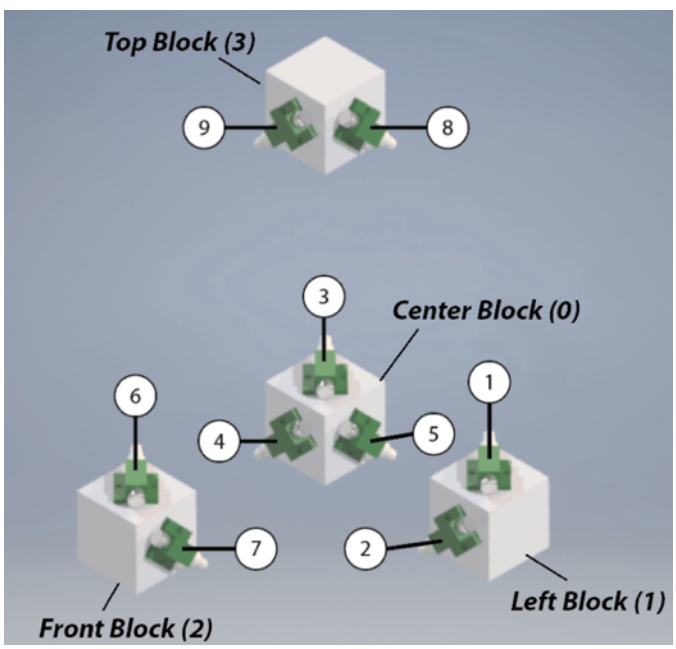


Figure 3.3: Default orientation of the suspended dummy headform.

The headform is equipped with nine ENDEVCO 7264C single-axis linear accelerometers (Meggitt Sensing Systems Irvine, California) arranged on a 3-2-2-2 array (Figure 3.4a-b) to measure linear $(a(t))$ and calculate rotational acceleration $(\alpha(t))$ (Padgaonkar, et al., 1975). Data were collected at a sampling rate of 20 kHz and transferred to a DAQ card controlled by a LabView program. Depending on the type of the helmet, the linear accelerometer data was filtered through a low-pass fourth-order Butterworth filter: 1650 Hz for cycling helmet and 1000 Hz for football and hockey helmet (Consumer Product Safety Commission, 1998; NOCSAE, DOC, 2017). Furthermore, the test rig is equipped with a high-speed camera by Sanstreak Corp, California (Figure 3.5). The camera is capable of recording up to 5000 frames-per-second (fps) in High Definition (HD, 1080p) to ensure the quality and consistency of the impact test.



(a)



(b)

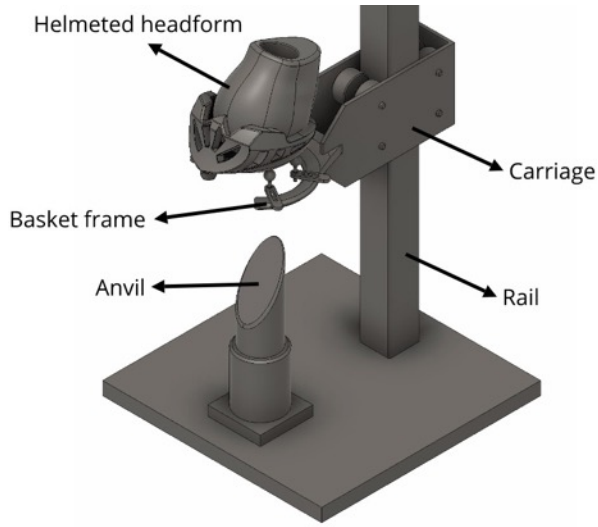
Figure 3.4: (a) Hybrid III drawing (Humanetics Innovative Solutions) with the sensors attached inside and (b) Illustrations of the 3-2-2-2 accelerometer assembly.

Figure 3.5: Edgertronic SC2+ high-speed camera (Sanstreak Corp., 2019).

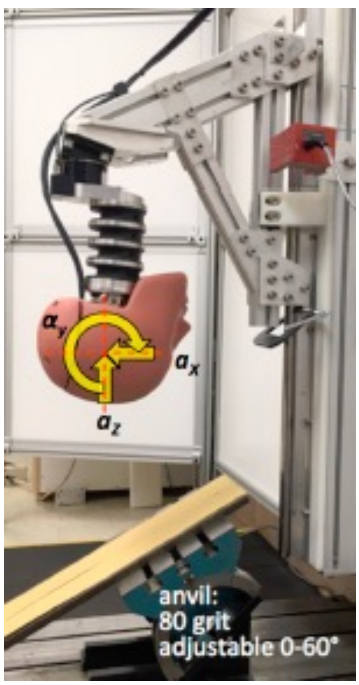
Other research studies have shown that the result of a full-body dummy versus a detached dummy headform impact testing can be different (Chinn, et al., 2001; Ghajari, et al., 2013). In some cases, a detached dummy headform tested at a higher speed can be used as a suitable replacement for a full-body dummy (Chinn, et al., 2001).

3.2.1. Comparison with Other Types of Impact Test Rig

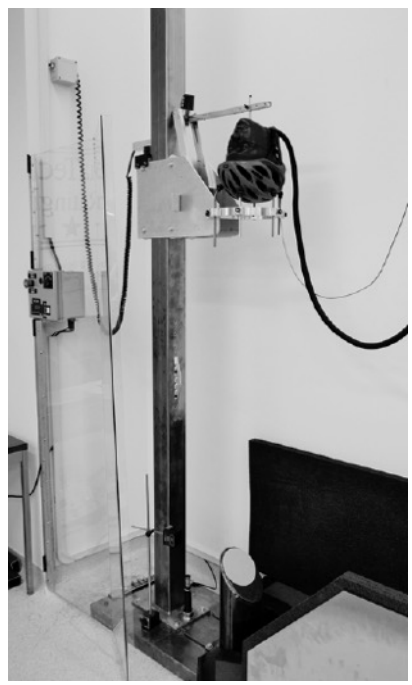
In sports-related accidents or injuries, generally, there are two types of impacts. The first type is fall, which is very common in cycling and hockey (against ice). The second type is a strike, which is also common in contact sports such as football and also hockey. Vertical drop test rig, similar to the one described in the previous section, is commonly used to simulate falls. The headform is fixed to the carriage using either a neckform or a solid attachment, which results in limited or no rotational acceleration of the headform. In cases where the ATD headform separates from the carriage after impact, a support structure called the basket frame is used to maintain a given headform orientation before impacting the anvil, as shown in Figure 3.6a. The helmeted dummy headform (e.g., Hybrid III head or NOCSAE) is dropped from a height on a vertical guide rail on to an impact opponent. The impact opponent can be a flat or angled anvil. Figure 3.6b and 3.6c show an example of another version of a vertical impact rig.



(a)



(b)



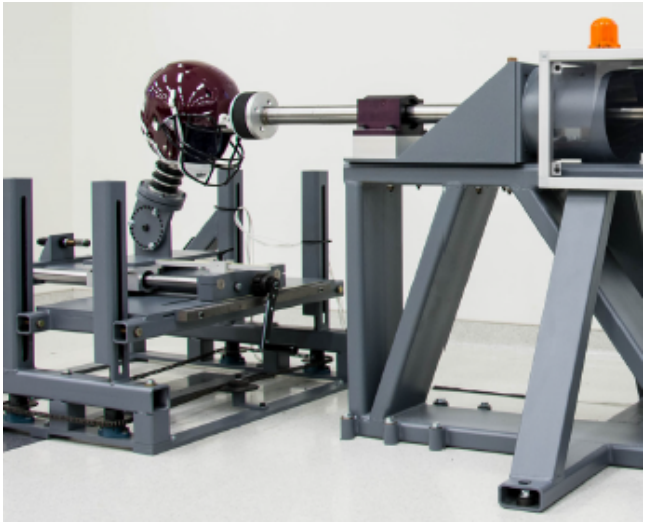
(c)

Figure 3.6: (a) An Oblique Impact Test Rig with a basket frame structure (b) Helmet Impact Testing facility at Biomechanics Laboratory, Legacy Research Institute, Portland (Bliven, et al., 2019) and (c) Oblique impact drop tower at Virginia Tech Helmet Lab, Blacksburg (Bland, et al., 2018).

Ringed basket frame has a disadvantage when used to perform impact tests onto anvils inclined at steep angles such as 15˚ to the vertical, as required in the ECE 22.05 testing standard (UN ECE Regulation 22/05, 2002). During such impacts, the helmeted headform may rebound off the anvil and into the path of the falling basket frame damaging and compromising the integrity of the testing apparatus and headform.

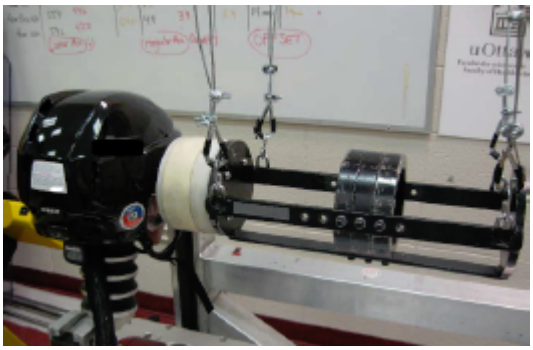
Another method used for evaluating headgear performance with ATD headforms is by generating speeds for both colliding objects (Pang, et al., 2011; Aare & Halldin, 2003; McIntosh, et al., 2013). At the moment of impact, friction at the impacting surfaces causes a tangential force on the helmeted headform, inducing rotational motion. Some studies used a combination of a pendulum and a rail where the headform was fixed to the pendulum arm. The headform and pendulum assembly strike an oblique anvil dropped from a height (Siegkas, et al., 2019; Roseveare, et al., 2016). As a result, the plate and pendulum methods can achieve high resultant impact speed (Siegkas, et al., 2019; Aare & Halldin, 2003). Testing with a moving anvil can present different impact conditions compared with a drop test onto static anvils. The horizontal striking plate could produce dissimilar head accelerations due to the different normal forces observed between the two scenarios (Willinger, et al., 2015). Although this method may be valuable for reproducing real-life impacts for automotive or sports accidents, having more moving components adds to the complexity of the testing and may cause repeatability and maintenance issues (Aare & Halldin, 2003; McIntosh, et al., 2013).

Aside from the drop rigs, there are other types of linear impact test rig: pneumatic linear impactor and pendulum impactor, as shown in Figure 3.7. During football and hockey helmet impact test, a stationary headform assembly is struck by a moving impactor mass (Jadischke, et al., 2016; Clark, et al., 2016; Pellman, et al., 2006; Post, et al., 2013; Oeur, et al., 2014). Pellman et al. (2006) and Jadischke et al. (2016) performed helmet impact tests with a pendulum arm, and Hybrid III head-neck assemblies mounted on a sliding base. The base allowed the assembly to have up to five degrees of freedom motion after impact. A similar rig was used in hockey impact study by Oeur et al. (2014), where a pneumatic linear ram rapidly pushed the impacting mass towards a head-neck assembly.



(a)

(b)



(c)

Figure 3.7: (a) Pneumatic linear impactor test rig at Virginia Tech Helmet Lab (Rowson, et al., 2018), (b) Pendulum impactor at Virginia Tech Helmet Lab, Blacksburg (Rowson, et al., 2015), and (c) Pendulum impactor at the School of Human Kinetics, University of Ottawa. (Oeur, et al., 2014)

Linear impactor test rigs such as pneumatic linear ram and pendulum arm simulate helmet-to-helmet impacts (Pellman, et al., 2016) and produce small tangential force (Willinger, et al., 2015). A study by Finan et al. showed that tangential force is a crucial contributor to the rotational acceleration of the head (Finan, et al., 2008). Furthermore, it is known that injuries are also commonly occurred due to impact against playing surface, body parts, and jersey (Naunheim, et al., 2002; Withnall, et al., 2005; Rossi, et al., 2016), which may not be accurately represented by linear impactor test rigs.

The purpose of the sliding base in linear impactor test rigs is to allow realistic motion of an object after being struck. A study showed that rigidly attaching a head-neck assembly to the base resulted in different impact response as in an accident reconstruction due to excessive loading (Pellman, et al., 2006). Another study suggested that the effect of changing torso mass was minimal on the head kinematics (Beusenberg, et al., 2001). In addition, when the translating table was utilized to simulate torso mass and found no significant difference in linear and rotational response between rigidly affixing the lower neck of the head-neck assembly versus mounting the lower neck to a weighted (12.78 kg) translating table (Walsh, et al., 2014).

Some researchers recommend to use the dummy headform coupled with a dummy neck (e.g. Hybrid III neck) (Rowson, et al., 2015; Rowson, et al., 2018; Oeur, et al., 2014; Bland, et al., 2018; Tyson & Rowson, 2018; Walsh, et al., 2011). However, Hybrid III dummy neck is known to be too stiff when used in a test-setting that measures rotational kinematics (Aare & Halldin, 2003). The Hybrid III neck is designed only for flexion and extension and the behavior for other types of motion is not studied (Aare & Halldin, 2003; Bartsch, et al., 2012; Myers, et al., 1989; Svensson & Lovsund, 1992; Herbst, et al., 1998; Gwin, et al., 2009). In some impact scenarios, the use of a neck can reverse the direction of the head rotational acceleration (Beusenberg, et al., 2001). Therefore, the results obtained from such impact scenarios may not be reliable.

A study suggested that the effect of the neck on the helmeted headform response is small for the first 10 ms of the impact (Willinger, et al., 2015). A typical peak acceleration in helmeted head impact lasts for approximately 15 ms (Pellman, et al., 2003; Zhang, et al., 2004; Deck & Willinger, 2008). From a study on human cadavers, a mechanical separation happened between head and neck for the first 9 ms of impact,

which results in minimal to no influence of the neck on the overall head kinematics (Nightingale, et al., 1996). Another study on human subjects showed that there is a minimal activity of the neck's muscles for the first 50 ms to 80 ms depending on whether or not the impact was being anticipated (Ono, et al., 2003). A study on human neck showed that the neck's muscles require 13 to 14 ms to react to mild impact, regardless if the impact was anticipated or not (Kuramochi, et al., 2004). In addition, there is an atlanto-occipital neutral zone where the neck joint can have a 10° motion without affecting the kinematics of the head (Ivancic, 2014; Camacho, et al., 1997). As a result, using a detached headform and a free-fall test rig is reasonable.

3.3. Description of Impact Testing Methodology

During an oblique impact to the head, the brain strain response (pattern) can be different based on the direction of the rotational motion of the head (Gennarelli, et al., 1987; Meaney & Smith, 2011). In other words, the severity of an impact differs based on the magnitude, direction, and location of the applied force(s). It is paramount for the KRS to be able to cover the most common impact location and to evaluate the helmet's performance in attenuating forces from all, if not most, scenarios possible.

The goal of the KRS is performing oblique impact tests with different impact scenarios on a helmet. The oblique impact tests were performed using the drop rig described in Section 3.2. The use of the free-fall drop test rig allows the helmeted headform to have an unrestricted motion during and after impact. The test methodology differs based on the type of helmet: cycling, football, and hockey.

3.3.1. Football Helmet Test Methodology

Taking into account the NOCSAE standard, the impact opponent (anvil) for football helmet test methodology is chosen to be a Modular Elastomer Programmer (MEP) pad. The anvil is inclined at 45° to provide even distribution of normal and tangential impact forces and was agreed among different research studies (Halldin, 2015; Halldin & Kleiven, 2013; Deck, et al., 2012).

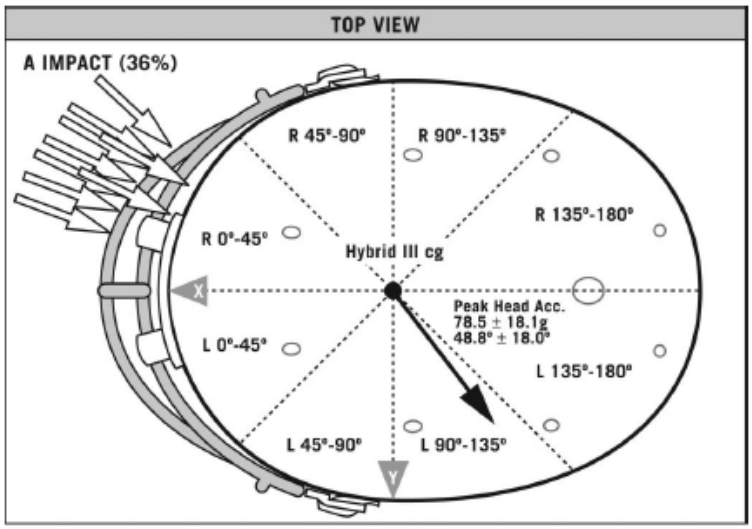
The impact speed is chosen to be 6.5 m/s (±3%), higher than the recommended 5.5 m/s from the NOCSAE standard. Some research studies suggest that football impact

happens at speeds higher than the speed defined in the standard (Pellman, et al., 2003; Tyson & Rowson, 2018; Sproule, et al., 2017). Other research studies have shown that a dummy headform without a neck tested at a higher speed can be used as a suitable replacement for a full-body dummy (Chinn, et al., 2001).

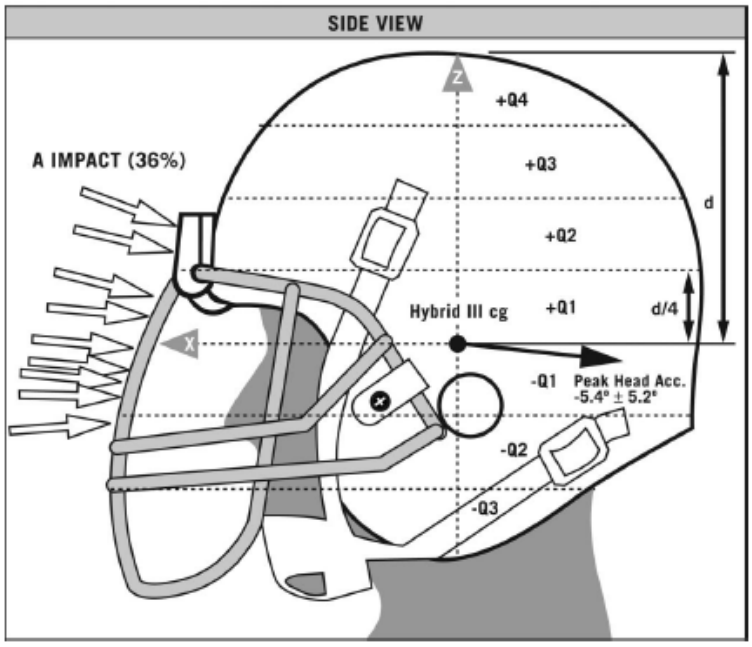
Helmets are tested without the visors, facemasks, or other attachments and are fitted according to the manufacturer's specifications. The 50th percentile Hybrid III headform has 59.7 cm head circumference. We found that the dummy headform works with most medium to large size helmets.

Determination of the Locations of Impact

Daniel conducted studies on youth (Daniel, et al., 2012), Rowson and Crisco on collegiate (Rowson & Duma, 2011; Crisco, et al., 2012), and Pellman on professional football players (Pellman, et al., 2003) to determine the frequency and location of head impacts. In each study, the recorded impacts were grouped into categories (Crisco, et al., 2010; Daniel, et al., 2012). However, as seen in Figure 3.8, Pellman partitioned the helmeted head differently (Pellman, et al., 2003). For this study, the categories defined by Crisco, Rowson, and Daniel (subsequently referred to as CRD) are adopted. An approach was taken to re-categorize the impacts recorded in Pellman's study into the groups defined by CRD. Figure 3.9 shows the partitioning of the head based on the defined impact groups by CRD. The location of an impact on the helmeted head can be described with a spherical coordinate system by specifying its azimuth (θ) and elevation (α) angle. Impacts were grouped into four different groups defined as front, back, lateral, and top impacts. Impacts with <65° and >115° elevations and 45° to -45° azimuths were grouped into front impacts, as shown in Figure 3.9. Impacts with <65° elevations and 135° to -135° azimuths were grouped into back impacts. Any impacts with above 65° elevations were grouped into top impacts. Other impacts were grouped into lateral impacts due to the head being symmetrical about the sagittal plane.



(a)



(b)

Figure 3.8: Partitioning of the helmet according to Pellman (Pellman, et al., 2003).

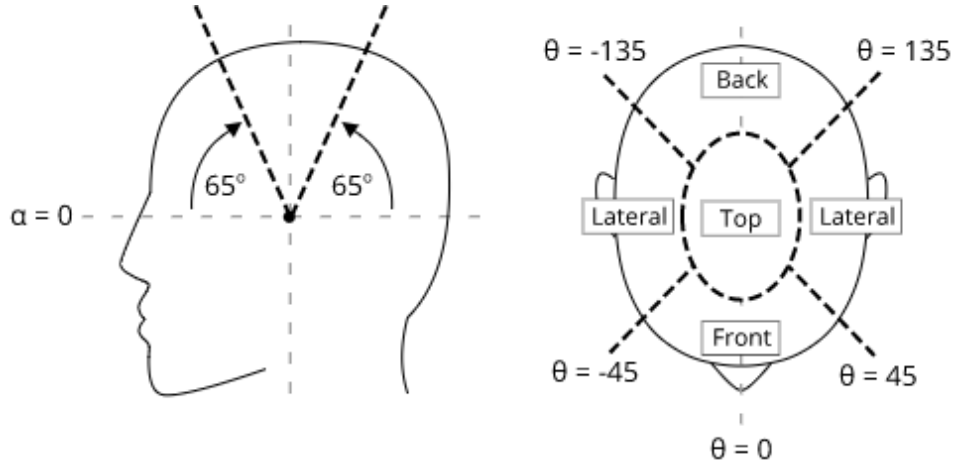


Figure 3.9: The partitioning of the head based on the defined impact groups.

The combined percentage of impact associated with each impact area was determined and summarized in Table 3.2, and the average value for each location was calculated.

Table 3.2: Summary of impact exposure probability from different football research studies.

Dataset	Impact Area			
	Front	**Back**	**Lateral**	**Top**
Youth (Daniel, et al., 2012)	31%	14%	36%	18%
Collegiate (Rowson & Duma, 2011; Crisco, et al., 2010)	37%	29%	18%	16%
National Football League (NFL) (Pellman, et al., 2003)	10%	27%	45%	17%
Average impact exposure	**26%**	**23%**	**33%**	**18%**

Based on Pellman study, the distribution of the impact concentration on the outer shell of the helmet was identified (Pellman, et al., 2003). Taking into account Pellman study and impact exposure probability in Table 3.2, four impact locations on the helmet were chosen (Figure 3.10). The four chosen locations are situated on each specific area. The helmeted headform was placed on the test rig and oriented such that each chosen location was contacting the anvil. Then, for each impact location, the helmet was removed while the headform orientation was preserved. Next, the locations on the headform were identified by contacting the anvil. As shown in Figure 3.11, locations #1,

2, 3, and 4 are situated on the lateral, top, front, and back sections of the headform, respectively.

(a) (b) (c)

Figure 3.10: Locations of impact for football helmet impact test.

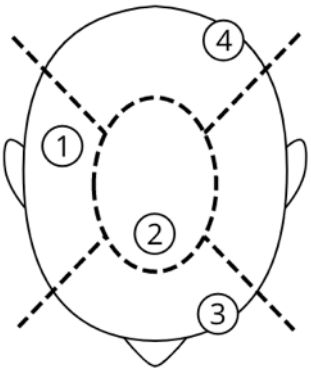
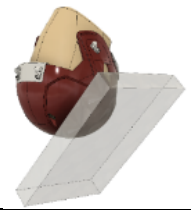
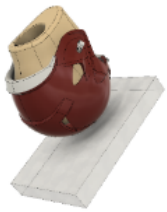
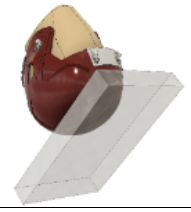
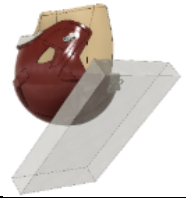
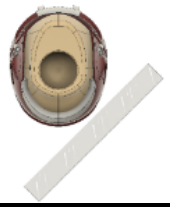
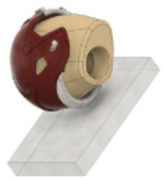
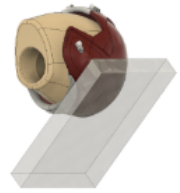


Figure 3.11: The selected impact locations on the headform.

The pre-impact orientation of the respective impact locations is illustrated in Table 3.3.

Table 3.3: Dummy headform equipped with football helmet suspended above the anvil showing different views of the locations of impact.

Location	View 1	View 2	View 3
1			
2			
3			
4			

The dummy headform pre-impact orientation at every impact location is specified in Table 3.4.

Table 3.4: Dummy headform pre-impact orientations for football helmet test.

Location #	Z	Y	X
1	-90*	24	0
2	0	37.5*	0
3	30*	0	0
4	90*	90	0

* Rotation performed first, moving reference frame

The impact exposure coefficients, which later are going to be used in calculating the helmet's grade are summarized in Table 3.5.

Table 3.5: Impact exposure coefficients for football helmet test.

	Location [n]	***Exposure [E(n)]***	***Description***
American football Helmet	**1**	0.34	Left
	2	0.17	Top
	3	0.28	Front
	4	0.22	Bottom

The linear and rotational acceleration of the head are recorded and the average peak acceleration values are measured. At every impact location, a minimum of three trials or tests are performed. The acquired trials must be within 10% of each other. Since the football helmet is designed to receive multiple impacts, one helmet is required per

helmet model. In total, twelve tests were performed per helmet model in a sequential order starting from location 1, moving on to 4 and repeat three times.

3.3.2. Hockey Helmet Test Methodology

ASTM (1045-99) and CSA (Z262.1-M90) standards are most widely used to test and certify hockey helmets. ASTM and CSA recommend 4.5 m/s impact speed while NOCSAE standard for hockey helmet recommends 5.5 m/s impact speed (Whyte, et al., 2019; NOCSAE, DOC, 2016). Other research studies have shown that a dummy headform without a neck can be used as a suitable replacement for a full-body dummy as long as it is tested at a higher speed (Chinn, et al., 2001). Therefore, the impact speed for hockey helmet test methodology is chosen to be 6.5 m/s (±3%) as the most agreeable impact speed across most of the standards and research studies (Rowson, et al., 2015; Whyte, et al., 2019; Allison, et al., 2017; Clark, et al., 2016; Potvin, et al., 2019).

All of the standards mentioned above call for a flat anvil impact test on an MEP pad. The impact opponent is chosen to be an MEP pad to simulate falls on ice or player's shoulder (Hoshizaki, et al., 2012; Kendall, et al., 2012). The impact angle was chosen to be at 45° as it provides even distribution of normal and tangential impact forces (Halldin, 2015).

Helmets are tested without the visors, facemasks, or other attachments and are fitted according to the manufacturer specifications. The 50th percentile Hybrid III headform has 59.7 cm head circumference, and the dummy headform fits well with most medium to large size helmets.

Determination of the Locations of Impact

Researchers have been conducting studies on hockey players where they equipped the players with instrumented helmets to determine the impact exposure rate on different regions of the helmet (Brainard, et al., 2012; Gwin, et al., 2009; Wilcox, et al., 2014; Mihalik, et al., 2012). Similar to football-related research, the head is partitioned into five areas (CRD), as shown in Figure 3.11. The research mentioned above were summarized in Table 3.6 and the probability of receiving an impact during an accident on different areas of the helmet was found. Impacts to the front, back, and

side of the helmet were similar in frequency, while impacts to the top were about four times less frequent.

Table 3.6: Summary of impact exposure probability from different hockey research studies.

Author	Description	Front	Back	Side	Top	Total
Brainard (Brainard, et al., 2012)	NCAA Men In-situ	4583	4970	4318	1410	**15281**
	NCAA Women In-situ	3704	4047	4008	1138	**12897**
Gwin (Gwin, et al., 2009)	NCAA Men In-situ	626	1458	1430	342	**3856**
Wilcox (Wilcox, et al., 2014)	NCAA Men In-situ	5566	5646	5984	1491	**18687**
	NCAA Women In-situ	4258	4548	4054	1322	**14182**
Mihalik (Mihalik, et al., 2012)	Youth In-situ	3974	2804	4313	1162	**12253**
Total (n)		*22711*	*23473*	*24107*	*6865*	***77156***
Total (%)		*29.4*	*30.4*	*31.2*	*8.9*	***100***

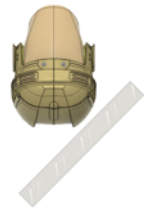
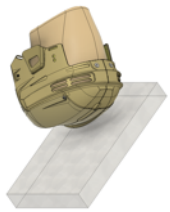
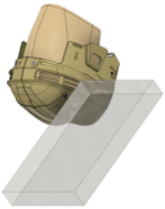
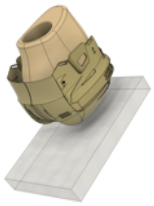
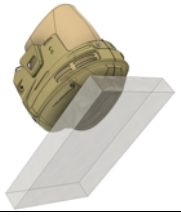
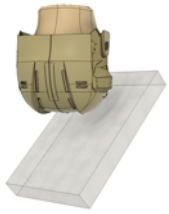
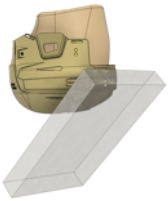
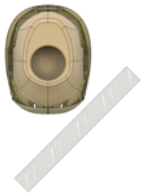
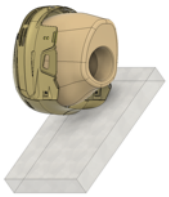
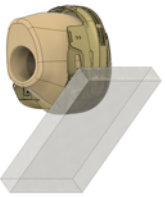
The areas around the head with the most impact concentration were determined to find candidates of the locations of impact. The locations of impact on the helmet are shown in Figure 3.12. Similar to determining the location of impact on a football helmet, the helmeted headform was placed on the test rig and oriented such that each chosen location was contacting the anvil. Then, for each impact location, the helmet was removed while the headform orientation was preserved. Next, the locations on the headform were identified by contacting the anvil. Similar to football helmet impact locations, as shown in Figure 3.11, locations #1, 2, 3, and 4 are situated on the lateral, top, front, and back sections of the headform, respectively.

(a) (b) (c)

Figure 3.12: Locations of impact for hockey helmet impact test.

The pre-impact orientation of the respective impact locations is illustrated in Table 3.7.

Table 3.7: Dummy headform equipped with hockey helmet suspended above the anvil showing different views of the locations of impact.

Location	View 1	View 2	View 3
1			
2			
3			
4			

The dummy headform pre-impact orientation at every impact location is specified in Table 3.4. The impact exposure coefficients, which later are going to be used in calculating the helmet's grade are summarized in Table 3.8.

Table 3.8: Impact exposure coefficients for hockey helmet test.

	Location [n]	***Exposure [E(n)]***
Hockey Helmet	**1**	0.31
	2	0.09
	3	0.29
	4	0.30

The linear and rotational acceleration of the head are recorded and the average peak acceleration values are measured. At every impact location, a minimum of three trials or tests are performed. The acquired trials must be within 10% of each other. Since the hockey helmet is designed to receive multiple impacts, one helmet is required per helmet model. In total, twelve tests were performed per helmet model in a sequential order starting from location 1, moving on to 4 and repeat three times.

3.3.3. Cycling Helmet Test Methodology

Consumer Product Safety Commission (CPSC) certification standard is one of the most widely used standards to certify a bicycle helmet (Consumer Product Safety Commission, 1998). Recently, the European Committee for Standardization (CEN) proposed a new certification standard, EN13087-11. The proposed EN13087-11 standard requires a vertical helmeted impact against a fixed 45° steel anvil. A 45° impact angle was chosen to provide even distribution of normal and tangential impact forces (Halldin, 2015; Bourdet, et al., 2012; Bourdet, et al., 2014; Peng, et al., 2012). In addition, some research have shown that the most common impact angles in cycling accidents were approximately between 30° and 60° to the vertical (Bourdet, et al., 2012; Bourdet, et al., 2014; Peng, et al., 2012).

The importance of the surface condition of the anvil has been studied. The study has shown that an 80-grit Aluminum oxide sandpaper can in general simulate a fair scenario of road conditions (Petersen, 2018). The use of sandpaper is similar to that in the United Nations Economic Commission for Europe (ECE) standard for motorcycle helmets and other bicycle oblique impact research (UN ECE Regulation 22/05, 2002; Halldin, et al., 2001; Bliven, et al., 2019; Bland, et al., 2018). Therefore, the impact opponent for cycling helmet test methodology is chosen to be a fixed 45° steel anvil covered with an 80-grit Aluminum oxide sandpaper. After each impact test, the sandpaper is replaced.

The impact speed is chosen to be 6.5 m/s (±3%) based on studies of severe real-life impacts (Halldin, 2015; Williams, 1991). However, it is possible that in the future, more tests with different (higher or lower) impact speed can be performed.

Helmets are tested without the visors, facemasks, or other attachments and are fitted according to the manufacturer specifications. The 50th percentile Hybrid III headform has 59.7 cm head circumference. It is found that the dummy headform fits well with most medium to large size helmets.

Determination of the Locations of Impact

Several research studies have been conducted to determine the impact exposure rate on different regions of the helmet (Bland, et al., 2018; Williams, 1991; McIntosh, et al., 1998; Depreitere, et al., 2004; Ching, et al., 1997; Smith, et al., 1994). All the aforementioned research studies performed either a survey to the victim of cycling accidents or helmet forensic analysis. By summarizing the previously mentioned research, the probability of receiving an impact during an accident on the pre-determined areas was generated (Figure 3.13). The impact probability distribution on a helmet is symmetrical with respect to the head's sagittal plane.

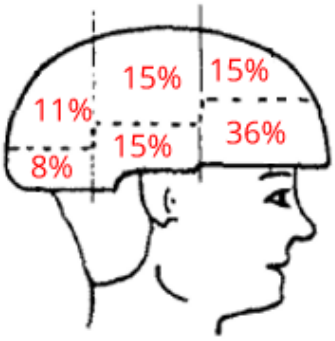


Figure 3.13: Impact exposure probability on a bicycle helmet (adapted from Williams) (Williams, 1991).

Similar to determining the location of impact on football and hockey helmet, the areas around the head with the most impact concentration were determined to find candidates of the locations of impact. According to CPSC and CEN standards, the distance between each two locations of impact has to be a minimum of 120 mm (Bland, et al., 2018; Consumer Product Safety Commission, 1998). The locations of impact on the helmet are shown in Figure 3.14. Then, the helmeted headform was placed on the test rig and oriented such that each chosen location was contacting the anvil. For each impact location, the helmet was removed while the headform orientation was preserved. Next, the locations of impact on the headform (Figure 3.15) were identified by contacting the anvil.

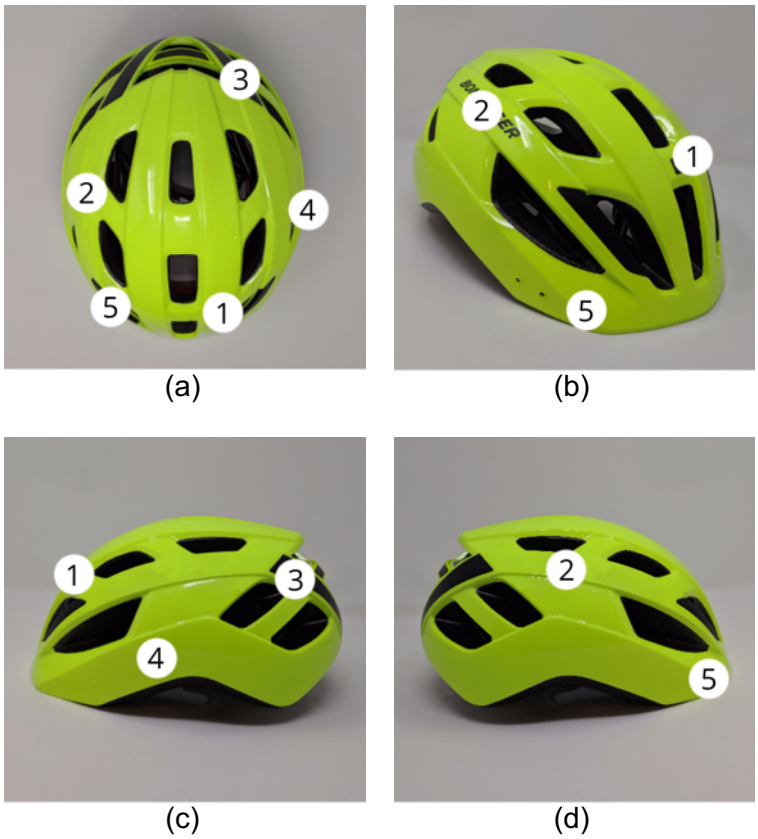


Figure 3.14: Locations of impact for cycling helmet impact test.

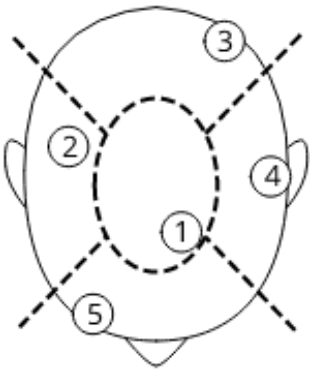
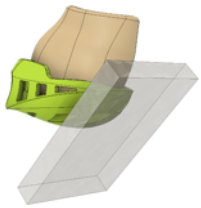
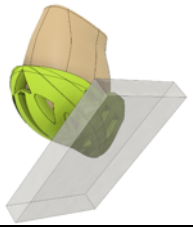
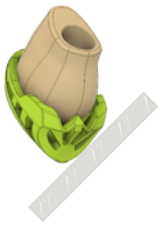
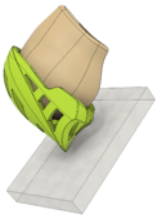
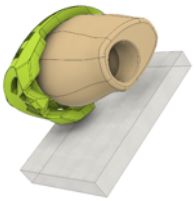
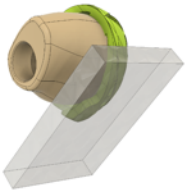
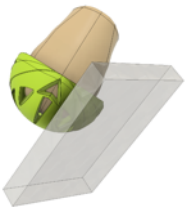


Figure 3.15: The selected impact locations on the headform.

The pre-impact orientation of the respective impact locations is illustrated in Table 3.9.

Table 3.9: Dummy headform equipped with cycling helmet suspended above the anvil showing different views of the locations of impact.

Location	View 1	View 2	View 3
1			
2			
3			
4			
5			

The dummy headform pre-impact orientation at every impact location is described in Table 3.10.

Table 3.10: Dummy headform pre-impact orientations for cycling helmet test..

Location #	Z	Y	X
1	-30*	-18	0
2	-90*	18	0
3	125*	37.5	0
4	45	0	90*
5	-45*	0	-32

* Rotation performed first, moving reference frame

The impact exposure coefficients, which later are going to be used in calculating the helmet's grade are summarized in Table 3.11.

Table 3.11: Impact exposure coefficients for cycling helmet test.

	Location [n]	***Exposure [E(n)]***
Cycling Helmet	**1**	0.16
	2	0.16
	3	0.12
	4	0.16
	5	0.39

The linear and rotational acceleration of the head are recorded and the average peak acceleration values are measured. At every impact location, a minimum of three trials or tests are performed. The acquired trials must be within 10% of each other. Since cycling helmet is designed to receive a single impact on each location, a minimum of three helmets are required per helmet model. In total, fifteen tests were performed per helmet model in a sequential order starting from location 1, moving on to 5 and repeat three times. For some helmet design, having the 120 mm distance between one location to another may not be possible. In this case, more helmets will be required as the five locations should not be tested on one helmet.

3.4. Development of the Grade Equation

As mentioned before, the grading system will consider both linear and rotational acceleration and their durations. The duration if acceleration pulse is represented by the velocity gain of the acceleration pulse. The velocity gain can be computed by integrating the acceleration pulse, or by calculating the area under the curve. Therefore, in order to take into account the duration of an acceleration pulse, as shown in Figure 3.16, the Area Under the Curve (AUC) between points A and B is calculated. Point A and B are selected from which the magnitude of the acceleration pulse is above a threshold of 5 g and 0.5 krad/s^2 for linear and rotational acceleration, respectively. The threshold was chosen to eliminate error from sensor's noise, ensuring the AUC is calculated only for the impact episode. Ultimately, two pieces of information are extracted from one acceleration measurement: peak and AUC. The AUC can be calculated using numerical integration with the help of computational tools such as MATLAB.

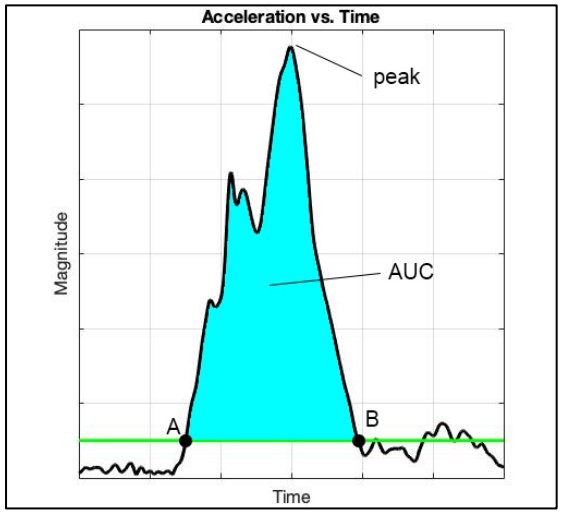


Figure 3.16: Extracting information from impact attenuation test result

The grade equation takes the peak and AUC of the acceleration pulse as its input. Each of the inputs is compared with a corresponding baseline value. The baseline value is obtained from performing the corresponding impact test on a non-helmeted (bare) dummy head. At every test scenario or impact location (n), the peak linear and rotational acceleration (a and α) are measured, and the AUC linear and rotational ($AUCL$ and $AUCR$) are calculated. Since a minimum of three tests per impact scenario are required, the average values of each input parameters are calculated. Then, the $Linear\ Grade(n)$ and $Rotational\ Grade(n)$ are calculated using (3.1) and (3.2), respectively.

$$\boldsymbol{Linear\ Grade} = \sum \boldsymbol{E(n)} \times \frac{\boldsymbol{a_{,baseline}(n)}}{\boldsymbol{a_{measured}(n)}} \times \frac{\boldsymbol{AUCL_{baseline}(n)}}{\boldsymbol{AUCL_{measured}(n)}} \tag{3.1}$$

$$\boldsymbol{Rotational\ Grade} = \sum \boldsymbol{E(n)} \times \frac{\boldsymbol{\alpha_{,baseline}(n)}}{\boldsymbol{\alpha_{measured}(n)}} \times \frac{\boldsymbol{AUCR_{baseline}(n)}}{\boldsymbol{AUCR_{measured}(n)}} \tag{3.2}$$

Where a and α are the linear and rotational acceleration respectively. The Exposure values ($E(n)$) are summarized in Table 3.5, Table 3.8, and Table 3.11. To reflect the magnification effect of both the pulse magnitude and duration, the peak acceleration value is multiplied by the AUC value. Finally, the final grade is calculated using equation (3.3).

$$\boldsymbol{Final\ Grade = Linear\ Grade \times Rotational\ Grade} \quad \textbf{(3.3)}$$

It was found that the brain deformation is more severe when the linear and rotational motion were applied together rather than independently (Gennarelli, et al., 1972; Ueno & Melvin, 1995; Pellman, et al., 2003). In other words, the damage from both accelerations magnifies each other. To reflect this modification, multiplications are used in calculating the Final Grade, as seen in (3.3).

The Final Grade is a dimensionless number representing how good a helmet attenuates severe impacts. The larger the number, the higher the performance. Mathematically, the range of the Final Grade is zero to infinity. The final grade represents the overall impact attenuating capability of a helmet, which consists of linear acceleration, rotational acceleration, linear velocity gain, and rotational velocity gain.

3.4.1. Baseline Values

Baseline values were experimentally determined by performing impact attenuation tests on the Hybrid III dummy headform without any helmets (bare dummy) at the predefined impact locations. The bare dummy was tested on both the 45° steel anvil covered with 80-grit Aluminum oxide sandpaper and the MEP pad at 6.5 m/s. The data were filtered with a fourth-order order Butterworth low-pass filter with 1650 Hz or 1000 Hz cut-off frequency (depending on test methodology). The Hybrid III dummy headform was covered with a vinyl skin that was soft (prone to be shaven by abrasive, if tested without protection). Therefore, to mitigate the damage from the harsh impact, especially from the roughness of the sandpaper, the bare dummy was covered with a 0.2 mm thick polyvinyl chloride (PVC) around the contacting area.

Table 3.12 summarizes the baseline values. Three trials were conducted per each impact location with a standard deviation that was less than 10% of the average to ensure repeatability.

Table 3.12: Summary of bare dummy headform testing used as the baseline values for the grading equations.

	Location	*Baseline Values*			
		Linear Acceleration (g)	***AUC Linear (g s)***	***Rotational Acceleration (krad/s^2)***	***AUC Rotational (krad/s)***
Cycling Helmet (Steel Covered with Sandpaper)	**1**	678.93	0.6466	16.79	0.0334
	2	626.68	0.5998	32.77	0.0506
	3	537.78	0.6202	25.86	0.0435
	4	566.64	0.5775	61.93	0.0970
	5	554.16	0.5750	45.69	0.0723
Football and Hockey Helmet (MEP)	**1**	339.26	0.6748	18.12	0.0598
	2	375.89	0.7025	11.13	0.0284
	3	343.18	0.7067	11.50	0.0316
	4	311.40	0.6885	18.41	0.0651

A further experiment was done in order to make sure the 0.2 mm PVC would not affect the testing result. A set of bare dummy test (onto 45° MEP pad) without the 0.2 mm PVC at lower impact speed (4.3 m/s) was carried on. Table 3.13 summarizes the test results.

Table 3.13: Test result for bare dummy headform with and without PVC cover.

	# Test	*Ave. Peak Lin. Acc. (g)*	*% diff*	*p-value*	*Ave. Peak Rot. Acc. (krad/s^2)*	*% diff*	*p-value*
Without PVC	3	197.84	-	-	6.89	-	-
With PVC	3	194.22	1.8%	0.1105	7.43	7.8%	0.0062

The result showed that the 0.2 mm PVC would increase the rotational acceleration of the head during impact by 8% and reduce the linear acceleration by 2%. These errors are considered to be minor. Therefore, the current baseline data is still valid to be used in the grade equation.

3.4.2. Letter Grade

The letter grade provides helmet user with a more comprehensive and easy-to-understand helmet performance result. To obtain the letter grade, first, the final grade of each helmet tested is normalized, as seen in (3.4). The maximum grade in (3.4) is the final grade of the best performing helmet of the same type. The maximum grade is 'floating' as its value can change when more helmets are tested.

$$\boldsymbol{Normalized\ Final\ Grade} = \frac{\boldsymbol{Final\ Grade}}{\boldsymbol{Maximum\ Grade}} \quad \textbf{(3.4)}$$

Equation (3.4) shows that the normalized final grade value ranges between 0.00 and 1.00. The obtained normalized final grade of each helmet is categorized into five different categories. The the letter grade is obtained by comparing the normalized final grade with Table 3.14.

Table 3.14: KRS letter grade category.

Normalized Final Grade (nFG)	*Letter Grade*
0.80 < nFG ≤ 1.00	A+
0.60 < nFG ≤ 0.80	A
0.40 < nFG ≤ 0.60	B
0.20 < nFG ≤ 0.40	C
0.00 < nFG ≤ 0.20	D

3.5. Summary

A Kinematic Rating System (KRS) for football, hockey, and cycling helmets were developed. The test methods in the KRS takes into account the most common impact speed, angle, and surface within each helmet type. The impact tests were performed on an oblique impact test rig, highlighting the normal and tangential component of the impact force. The KRS takes into account the head kinematics, such as linear and rotational acceleration during impact. The KRS uses the peak value and duration of the acceleration pulses to gauge the severity of an impact. The KRS does not attempt to correlate impact severity and injury severity. Therefore, KRS provides a non-biased way of evaluating helmet performance.

The following chapter will discuss the validation of the KRS. Some helmets from each category were tested according to the KRS. The results were compared with results from the available helmet rating system, such as STAR rating system.

Chapter 4.

Validation of the Helmet Evaluation Method

Four football, four hockey, and six cycling helmets were tested according to the KRS. The helmets were graded based on the equations that were also defined previously.

4.1. Results

4.1.1. Football Helmet Test

Four football helmets were tested. The information on the tested football helmets can be found in Table 4.1:.

Table 4.1. Description of the tested football helmets.

No.	*Helmet Name*	*Picture*	*Size*	*Price (USD)*
1	Schutt Vengeance Z10		Large	210.00

2	Xenith X2E+		Large	289.00
3	Riddell Speed		Large	279.99
4	VICIS ZERO1		Large	950.00

One football helmet was used for testing each helmet model at four impact locations. The results of all helmets were summarized in Table 4.2 to Table 4.5.

Table 4.2: KRS result for Schutt Vengeance Z10.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	86.50	0.7593	3.48	4.19	0.0270	9.45
2	81.09	0.6534	4.97	5.13	0.0292	2.08
3	112.34	0.6916	3.12	4.54	0.0304	2.59
4	76.46	0.6719	4.16	5.01	0.0451	5.23
		Linear Grade	***3.81***		***Rotational Grade***	***5.37***
			Final Grade: 20.4			

Table 4.3: KRS result for Xenith X2E+.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	105.65	0.8014	2.70	4.23	0.0239	10.58
2	82.29	0.7165	4.47	4.46	0.0258	2.72
3	95.72	0.7562	3.35	4.01	0.0267	3.33
4	86.75	0.7434	3.32	4.07	0.0363	7.98
		Linear Grade	***3.33***		***Rotational Grade***	***6.68***
			Final Grade: 22.3			

Table 4.4: KRS result for Riddell Speed.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	97.84	0.7289	3.20	4.95	0.0261	8.30
2	92.60	0.6625	4.29	5.39	0.0295	1.96
3	101.05	0.6937	3.47	5.41	0.0333	2.00
4	82.95	0.6884	3.74	5.38	0.0302	7.27
		Linear Grade	***3.59***		***Rotational Grade***	***5.29***
			Final Grade: 19.0			

Table 4.5: KRS result for VICIS ZERO1.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	89.91	0.6399	3.97	6.98	0.0447	3.44
2	74.37	0.5991	5.91	4.52	0.0375	1.85
3	76.42	0.6902	4.58	4.07	0.0330	2.66
4	94.31	0.7663	2.96	7.33	0.0492	3.29
		Linear Grade	***4.25***		***Rotational Grade***	***2.92***
			Final Grade: 12.4			

Out of all tested helmets, Xenith X2E+ received the highest final grade (22.3), and VICIS ZERO1 received the worst (12.4). Although VICIS ZERO1 received the best linear grade (4.25), the rotational grade is significantly lower than the rest of the helmets

(2.92). The letter grade of each of the tested helmets was obtained by normalizing each helmet grade with the highest grade (22.3) and comparing the normalized grade with Table 3.14. The football helmet results are summarized in Table 4.6.

Table 4.6: Summary of football helmet KRS results.

No.	*Helmet Name*	*Final Grade*	*Normalized Final Grade*	*Letter Grade*
1	Schutt Vengeance Z10	20.4	0.91	A+
2	Xenith X2E+	22.3*	1.00	A+
3	Riddell Speed	19.0	0.85	A+
4	VICIS ZERO1	12.4	0.56	B

* The maximum final grade in football helmet category.

4.1.2. Hockey Helmet Test

Four hockey helmets were tested. The information on the tested hockey helmets can be found in Table 4.7.

Table 4.7: Description of the tested hockey helmets.

No.	*Helmet Name*	*Picture*	*Size*	*Price (CAD)*
1	CCM FL40		Large	54.99
2	Bauer 5100		Large	104.99

3	CCM FL500		Large	249.99
4	Bauer REAKT 200		Large	349.99

One hockey helmet was tested per helmet model at four impact locations. The results of all helmets were summarized in Table 4.8 to to Table 4.11.

Table 4.8: KRS result for CCM FL40.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	126.12	0.7606	2.38	5.74	0.0406	4.59
2	183.38	0.7464	1.93	8.28	0.0341	1.11
3	101.48	0.6947	3.45	6.86	0.0373	1.41
4	89.33	0.6971	3.46	5.80	0.0415	4.92
		Linear Grade	***2.98***		***Rotational Grade***	***3.44***

Final Grade: 10.2

Table 4.9: KRS result for Bauer 5100.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	130.52	0.7217	2.43	6.89	0.0363	4.29
2	178.29	0.6763	2.19	8.20	0.0291	1.32
3	137.78	0.6728	2.62	6.32	0.0290	1.97
4	111.38	0.7116	2.70	5.94	0.0350	5.71
		Linear Grade	***2.54***		***Rotational Grade***	***3.77***

Final Grade: 9.6

Table 4.10: KRS result for CCM FL500.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	84.27	0.6953	3.89	4.53	0.0298	7.92
2	91.25	0.6242	4.63	6.21	0.0367	1.37
3	87.62	0.6762	4.08	5.34	0.0349	1.92
4	81.79	0.7201	3.63	5.36	0.0379	5.82
	Linear Grade		***3.93***	***Rotational Grade***		***4.93***

Final Grade: 19.4

Table 4.11: KRS result for Bauer REAKT 200.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	89.76	0.7531	3.38	5.06	0.0339	6.25
2	86.69	0.6419	4.77	5.75	0.0310	1.76
3	90.06	0.7161	3.75	5.08	0.0338	2.09
4	79.12	0.7170	3.78	4.46	0.0319	8.30
	Linear Grade		***3.73***	***Rotational Grade***		***5.24***

Final Grade: 19.6

Although CCM FL500 received the best linear grade (3.93), Bauer REAKT 200 received a significantly higher rotational grade (5.24). Therefore, out of all tested helmets, Bauer REAKT 200 received the highest final grade (19.6), while closely being followed by CCM FL500 (19.4). Bauer 5100 performed the worse with a final grade of 9.6. The letter grade of each of the tested helmets was obtained by normalizing each helmet grade with the highest grade (19.6) and comparing the normalized grade with

Equation (3.4) shows that the normalized final grade value ranges between 0.00 and 1.00. The obtained normalized final grade of each helmet is categorized into five different categories. The the letter grade is obtained by comparing the normalized final grade with Equation (3.4) shows that the normalized final grade value ranges between 0.00 and 1.00. The obtained normalized final grade of each helmet is categorized into five different categories. The the letter grade is obtained by comparing the normalized final grade with Table 3.14. The hockey helmet results are summarized in Table 4.12.

Table 4.12: Summary of hockey helmet KRS results.

No.	*Helmet Name*	*Final Grade*	*Normalized Final Grade*	*Letter Grade*
1	CCM FL40	10.2	0.52	B
2	Bauer 5100	9.6	0.49	B
3	CCM FL500	19.4	0.99	A+
4	Bauer REAKT 200	19.6*	1.00	A+

* The maximum final grade in hockey helmet category.

4.1.3. Cycling Helmet Test

Six cycling helmets were tested. The information on the tested cycling helmets can be found in Table 4.13.

Table 4.13: Description of the tested cycling helmets.

No.	*Helmet Name*	*Picture*	*Size*	*Price (CAD)*
1	Alibaba*		One-size	-
2	Specialized Covert		M	-

3	Schwinn Excursion		U	39.99
4	Bontrager Solstice		M/L	54.99
5	Bontrager Spectre WaveCel		M	194.99

6	Bontrager Ballista MIPS		M	269.99

* Obtained directly from a manufacturer in China.

Taking into account the 120 mm minimum distance between each location, a minimum of three cycling helmet was tested per helmet model at five impact locations. The results of all helmets were summarized in Table 4.14 to Table 4.19.

Table 4.14: KRS result for Alibaba helmet.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	176.50	0.6495	3.81	12.15	0.0371	1.24
2	186.71	0.6898	2.92	12.49	0.0400	3.29
3	152.82	0.6803	3.20	8.00	0.0258	5.42
4	177.71	0.5854	3.15	22.40	0.0585	4.59
5	190.21	0.7117	2.35	12.17	0.0551	4.90
	Linear Grade		***2.91***	***Rotational Grade***		***4.05***

Final Grade: 11.8 (D)

Table 4.15: KRS result for Specialized Covert.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	155.48	0.6431	4.37	12.03	0.0376	1.24
2	174.30	0.6945	3.11	10.71	0.0397	3.87
3	167.46	0.7052	2.77	10.52	0.0307	3.61
4	164.89	0.5788	3.43	19.34	0.0546	5.67
5	148.42	0.6609	3.25	10.73	0.0453	6.82
	Linear Grade		***3.38***	***Rotational Grade***		***4.86***

Final Grade: 16.4

Table 4.16: KRS result for Schwinn Excursion.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	117.89	0.6618	5.60	7.06	0.0327	2.40
2	153.76	0.7115	3.27	11.57	0.0400	3.59
3	143.38	0.6814	3.43	8.89	0.0316	3.99
4	157.25	0.6093	3.41	18.95	0.0545	5.81
5	151.42	0.6997	3.02	8.32	0.0359	11.22
	Linear Grade		***3.59***	***Rotational Grade***		***6.79***

Final Grade: 24.4

Table 4.17: KRS result for Bontrager Solstice.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	123.57	0.6379	5.55	8.01	0.0336	2.07
2	158.03	0.7004	3.39	8.68	0.0390	4.56
3	116.83	0.6286	4.54	6.94	0.0326	4.94
4	164.78	0.6024	3.31	20.52	0.0583	5.02
5	126.11	0.6567	3,84	8.73	0.0488	7.68
	Linear Grade		***4.04***	***Rotational Grade***		***5.49***

Final Grade: 22.2

Table 4.18: KRS result for Bontrager Spectre WaveCel.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	109.79	0.6468	6.15	5.59	0.0282	3.51
2	108.53	0.6745	5.12	5.48	0.0347	8.59
3	94.51	0.6442	5.48	4.37	0.0223	11.32
4	114.03	0.5806	4.93	13.53	0.0487	9.07
5	105.99	0.6570	4.57	5.63	0.0331	17.50
	Linear Grade		***5.08***	***Rotational Grade***		***11.65***

Final Grade: 59.2

Table 4.19: KRS result for Bontrager Ballista MIPS.

Loc. (n)	*Ave. Lin. Acc. (g)*	*Ave. Lin. AUC (g/s)*	*Linear Grade (n)*	*Ave. Rot. Acc. (krad/s^2)*	*Ave.Rot. AUC (krad/s)*	*Rot. Grade (n)*
1	106.28	0.6387	6.43	3.09	0.0195	9.00
2	103.68	0.6648	5.46	7.80	0.0336	6.26
3	95.02	0.6089	5.75	4.05	0.0277	9.84
4	145.98	0.5031	4.45	14.79	0.0448	9.02
5	117.32	0.6363	4.26	5.12	0.0317	20.15
	Linear Grade		***5.02***	***Rotational Grade***		***13.02***

Final Grade: 65.3

Out of all tested helmets, Bontrager Ballista MIPS received the highest final grade (65.3). Bontrager Spectre Wavecel came second with a final grade of 59.2. Both helmets received an almost similar linear grade, but Bontrager Ballista MIPS received higher rotational grade compared to Bontrager Spectre Wavecel (13.02 vs. 11.65). The Alibaba helmet received the worst linear, rotational, and final grade (2.91, 4.05, and 11.8, respectively). The letter grade of each of the tested helmets was obtained by normalizing each helmet grade with the highest grade (65.3) and comparing the normalized grade with Equation (3.4) shows that the normalized final grade value ranges between 0.00 and 1.00. The obtained normalized final grade of each helmet is categorized into five different categories. The the letter grade is obtained by comparing the normalized final grade withEquation (3.4) shows that the normalized final grade value ranges between 0.00 and 1.00. The obtained normalized final grade of each helmet is categorized into five different categories. The the letter grade is obtained by comparing the normalized final grade with Table 3.14. The cycling helmet results are summarized in Table 4.20.

Table 4.20: Summary of cycling helmet KRS results.

No.	*Helmet Name*	*Final Grade*	*Normalized Final Grade*	*Letter Grade*
1	Alibaba*	11.8	0.18	D
2	Specialized Covert	16.4	0.25	C
3	Schwinn Excursion	24.4	0.37	C
4	Bontrager Solstice	22.2	0.34	C
5	Bontrager Spectre WaveCel	59.2	0.91	A+
6	Bontrager Ballista MIPS	65.3*	1.00	A+

* The maximum final grade in cycling helmet category.

4.2. Discussion and Analysis

Over the years, the properties and design of the padding have changed. For example, modern football helmets utilize foam padding (air cushion) that can be inflatable or reversible in some designs. With the advancement in helmet design and new material used, the helmet design is becoming more complex. For example, the air cushion of Xenith X2E+ is fixed on a hollow plastic block that is attached to the outer shell. VICIS ZERO1 structure consists of narrow columns that are arranged radially between the inner shell and the softer outer shell. The columns are designed to buckle upon receiving an impact; the softer outer shell allows the flexing of the helmet. Bontrager Ballista MIPS utilizes a thin plastic interface to facilitate sliding between the head and the helmet. Bontrager Wavecel uses an auxetic material in combination with the conventional foam to allow better compression and shearing motion between the head and the helmet. Therefore, the response of the helmets under impact tests are diversifying and cannot be assumed the same. This results in the different magnitude of

acceleration and the corresponding duration of an impact on each design, as seen in the impact testing results throughout this research.

The results of the KRS were compared with results from the available STAR rating system. The comparison shows that most of KRS results disagree with STAR rating system results. More discussion and analysis were done in the next subsections.

4.2.1. Comparison between KRS and STAR Rating System Results

KRS final grade is a dimensionless number representing how well a helmet attenuates severe impacts compared to when no helmet is present in an impact. The larger the number, the higher the performance. Unlike the STAR rating system, where better helmets receiver lower number, the larger the grade, the better the performance. As an example, a 50-grade helmet is twice better than a 25-grade helmet, according to KRS. However, one cannot say that the linear acceleration attenuation capability of a 50-grade helmet is twice better than a 25-grade helmet. Furthermore, the final grade of KRS is compared to a floating maximum, this eliminates having an absolute fixed grading category.

Football Helmet

According to the Virginia Tech STAR rating system, all tested football helmets are rated five-star. Out of the four tested helmets, VICIS ZERO1 received the best rating (1.92), with Xenith X2E+, Schutt Vengeance Z10, and Riddell Speed came second (2.92), third (6.28), and fourth (6.67) respectively. Most of the NFLPA testing result agrees with STAR. It is also important to note that VICIS ZERO1 is one of the most expensive helmets available in the market. However, KRS results do not agree with Virginia Tech results. According to KRS, VICIS ZERO1 performed the worse out of all tested helmets due to its poor rotational acceleration and velocity performance. The comparison of the KRS and STAR rating system results of the four tested helmets are summarized in Table 4.21.

Table 4.21: Comparison of KRS and STAR rating system for the tested football helmets.

Helmet	*KRS Grade*	*KRS Rank*	*STAR Rating*	*STAR Rank*	*NFLPA Rank*
Schutt Vengeance Z10	**A+ (20.4)**	2	★★★★★**(6.28)**	3	4
Xenith X2E+	**A+ (22.3)**	1	★★★★★**(2.92)**	2	2
Riddell Speed	**A+ (19.0)**	3	★★★★★**(6.67)**	4	3
VICIS ZERO1	**B (12.4)**	4	★★★★★**(1.92)**	1	1

The STAR and KRS results for football were also compared with NFLPA results. According to NFLPA helmet laboratory performance testing, VICIS ZERO1 performed better than all helmets tested in 2017-2019 (NFLPA, 2019). The STAR rating system and NFLPA testing methodologies are fairly similar using the linear impactor test, and therefore most of both results are in sync. In addition, the rotational acceleration measured during testing with an impactor mostly represents the rotational acceleration caused by a normal force to the helmet and not a tangential force. Therefore, if a helmet mitigates linear acceleration well, it will also provide good results for rotational acceleration caused by the normal force. However, most impacts in real-life happens at an angle and therefore inducing normal and tangential force. KRS provides a more comprehensive testing methodology by addressing the tangential component of the impact force beside the linear or compressive component. According to KRS, VICIS ZERO1 performed the worst out of all tested helmets.

Helmets with similar structures, such as Schutt Vengeance Z10, Xenith X2E+, and Riddell Speed, received similar grades with low variations according to the STAR rating system and KRS. However, VICIS ZERO1 received conflicting results. The unique structure of VICIS ZERO1 allows for more significant attenuation of compression forces. This was shown in impact testing, such as in the STAR rating system and NFLPA, where the compression force is the major component of the impact force and the rotational acceleration is caused by the normal force and not a tangential force. According to KRS result, VICIS ZERO1 also received a superior linear grade as compared with other tested helmets. However, when subjected to tangential force, VICIS ZERO1 received a relatively poor rotational grade.

As seen in Table 4.22 Table 4.23, Figure 4.1a-d, and Figure 4.2a-d, VICIS ZERO1 attenuated linear acceleration better than the rest of the tested helmets locations 2 and 3 (as seen in Figure 4.1b and Figure 4.1c). The AUC linear of VICIS ZERO1 was the best in locations 1, 2, and 3. Considering both peak linear, AUC linear, and impact exposure coefficient, VICIS ZERO1 received the highest linear grade of 6.24. In contrast, as shown in Table 4.22, VICIS ZERO1 rotational acceleration attenuation capability was poor in locations 1 and 4 (Figure 4.2a and Figure 4.2d). A helmet that received the lowest peak accelerations (Schutt Vengeance Z10 for linear acceleration) did not necessarily end up with the highest grade. This is due to the introduction of AUC in the grade calculations.

Table 4.22: Linear grade comparison of different football helmets showing the linear grade while the linear acceleration (g) and AUC linear (m/s) are shown inside the parentheses.

Helmet	***Location***			
	1	***2***	***3***	***4***
Schutt Vengeance Z10	3.48 (86.50 g - 7.45 m/s)	4.97 (81.09 g - 6.41 m/s)	3.12 (112.34 g - 6.78 m/s)	4.16 (76.46 g - 6.59 m/s)
Xenith X2E+	2.70 (105.65 g - 7.86 m/s)	4.47 (82.29 g - 7.03 m/s)	3.35 (95.72 g - 7.42 m/s)	3.32 (86.75 g - 7.29 m/s)
Riddell Speed	3.20 (97.84 g - 7.15 m/s)	4.29 (92.60 g - 6.50 m/s)	3.47 (101.05 g - 6.81 m/s)	3.74 (82.95 g - 6.75 m/s)
VICIS ZERO1	3.97 (89.10 g - 6.28 m/s)	5.91 (74.37 g - 5.88 m/s)	4.58 (76.42 g - 6.77 m/s)	2.96 (94.31 g - 7.52 m/s)

Table 4.23: Rotational grade comparison of different football helmets showing the linear grade while the Rotational acceleration (krad/s^2) and AUC Rotational (krad/s) are shown inside the parentheses.

Helmet	***Location***			
	1	***2***	***3***	***4***
Schutt Vengeance Z10	9.45 (4.19 krad/s^2 - 0.0270 krad/s)	2.08 (5.13 krad/s^2 - 0.0292 krad/s)	2.59 (4.54 krad/s^2 - 0.0304 krad/s)	5.23 (5.01 krad/s^2 - 0.0451 krad/s)
Xenith X2E+	10.58 (4.23 krad/s^2 - 0.0239 krad/s)	2.72 (4.46 krad/s^2 - 0.0258 krad/s)	3.33 (4.01 krad/s^2 - 0.0267 krad/s)	7.98 (4.07 krad/s^2 - 0.0363 krad/s)
Riddell Speed	8.30 (4.95 krad/s^2 - 0.0261 krad/s)	1.96 (5.39 krad/s^2 - 0.0295 krad/s)	2.00 (5.41 krad/s^2 - 0.0333 krad/s)	7.27 (5.38 krad/s^2 - 0.0302 krad/s)
VICIS ZERO1	3.44 (6.98 krad/s^2 - 0.0447 krad/s)	1.85 (4.52 krad/s^2 - 0.0375 krad/s)	2.66 (4.07 krad/s^2 - 0.0330 krad/s)	3.29 (7.33 krad/s^2 - 0.0492 krad/s)

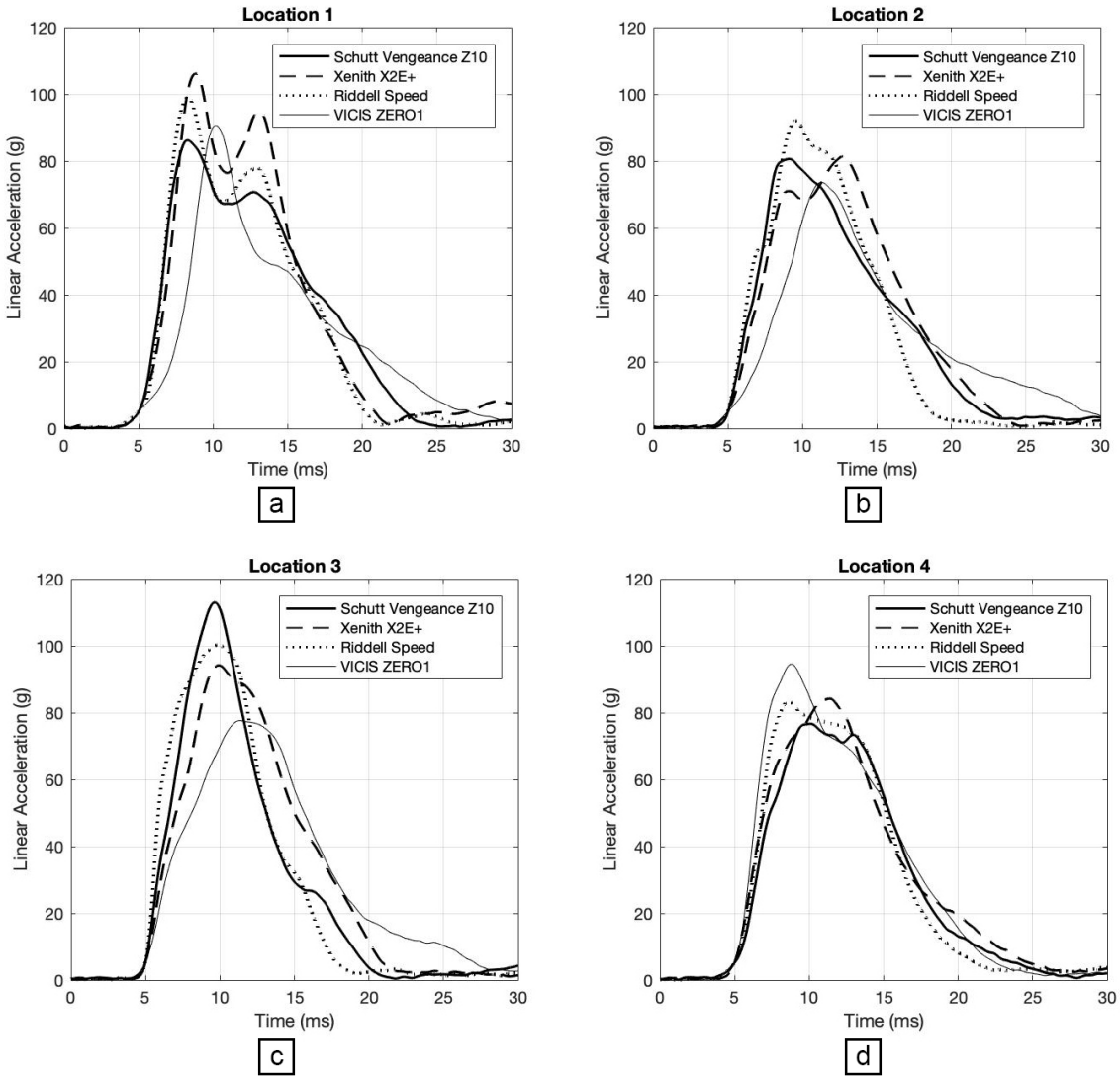


Figure 4.1: Plot of average rotational acceleration versus time of all football helmets tested on (a) location 1, (b) location 2, (c) location 3, and (d) location 4.

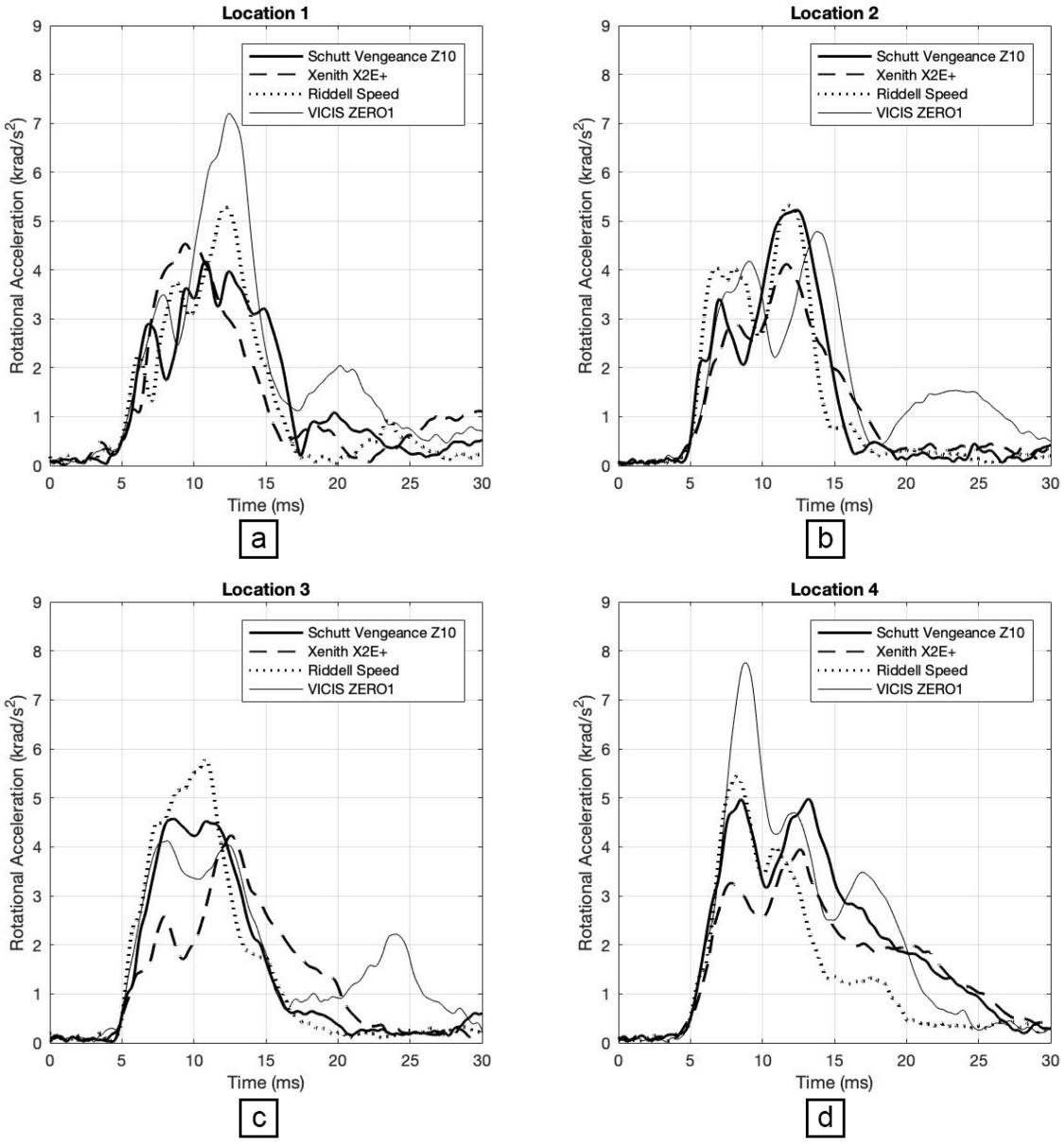


Figure 4.2: Plot of average linear acceleration versus time of all football helmets tested on (a) location 1, (b) location 2, (c) location 3, and (d) location 4.

Although in football impacts, linear acceleration is highly correlated to rotational acceleration (Rowson & Duma, 2011) due to the impact mostly induces compression forces, it does not mean that helmets should not be tested for tangential forces because such impact exists and is severe. KRS results show that the design some helmets, such as VICIS ZERO1, requires to be improved to better protect users against tangential forces.

Hockey Helmet

Two out of four tested hockey helmets (CCM FL500 and Bauer REAKT 200) are rated five-star in the STAR rating by Virginia Tech while CCM FL40 and Bauer 5100 are zero-star and two-star, respectively. According to KRS, CCM FL500 and Bauer REAKT 200 received a final grade of A+, while CCM FL40 and Bauer 5100 were graded B. The comparison between KRS and STAR rating system results can be seen in Table 4.24.

Table 4.24: Comparison of KRS and STAR rating system for the tested hockey helmets.

Helmet	*KRS Grade*	*KRS Rank*	*STAR Rating*	*STAR Rank*
CCM FL40	**B (10.2)**	3	**0 (12.19)**	4
Bauer 5100	**B (9.6)**	4	★★ **(6.31)**	3
CCM FL500	**A+ (19.4)**	2	★★★★★ **(1.67)**	1
Bauer REAKT 200	**A+ (19.6)**	1	★★★★★ **(3.30)**	2

As shown in Table 4.24, the results of the KRS and the STAR rating system in each category are not in sync. For example, according to the STAR rating system, CCM FL40 performed the worst while, according to KRS, Bauer 5100 performed the worst. Similar to KRS for football helmet, KRS for hockey provides a more complete testing methodology by addressing the tangential component of the impact force beside the linear or compressive component.

Table 4.25 and Table 4.26 present the detailed linear and rotational results of all tested hockey helmets. Based on the peak and duration of linear acceleration, CCM FL500 performed the best linearly in two out of four impact locations, followed by Bauer REAKT 200, resulting in CCM FL500 received the best linear grade. On the other hand, Bauer REAKT 200 performed the best rotationally in three out of four impact locations. Overall, the final grades of CCM FL500 and Bauer REAKT200 were close.

Table 4.25: Linear grade comparison of different hockey helmets showing the linear grade while the linear acceleration (g) and AUC linear (m/s) are shown inside the parentheses.

Helmet	*Location*			
	1	*2*	*3*	*4*
CCM FL40	2.38 (126.12 g - 7.46 m/s)	1.93 (183.38 g - 7.32 m/s)	3.45 (101.48 g - 6.81 m/s)	3.46 (89.33 g - 6.84 m/s)
Bauer 5100	2.43 (130.52 g - 7.08 m/s)	2.19 (178.29 g - 6.63 m/s)	2.62 (137.78 g - 6.60 m/s	2.70 (111.38 g - 6.98 m/s)
CCM FL500	3.89 (84.27 g - 6.82 m/s)	4.63 (91.25 g - 6.12 m/s	4.08 (87.62 g - 6.63 m/s)	3.63 (81.79 g - 7.06 m/s)
Bauer REAKT 200	3.38 (89.76 g - 7.39 m/s)	4.77 (86.69 g - 6.30 m/s)	3.75 (90.06 g - 7.02 m/s)	3.78 (79.12 g - 7.03 m/s)

Table 4.26: Rotational grade comparison of different hockey helmets showing the linear grade while the Rotational acceleration (krad/s^2) and AUC Rotational (krad/s) are shown inside the parentheses.

Helmet	*Location*			
	1	*2*	*3*	*4*
CCM FL40	4.59 (5.74 krad/s^2 - 0.0406 krad/s)	1.11 (8.28 krad/s^2 - 0.0341 krad/s)	1.41 (6.86 krad/s^2 - 0.0373 krad/s)	4.92 (5.80 krad/s^2 - 0.0415 krad/s)
Bauer 5100	4.29 (6.89 krad/s^2 - 0.0363 krad/s)	1.32 (8.20 krad/s^2 - 0.0291 krad/s)	1.97 (6.32 krad/s^2 - 0.0290 krad/s)	5.71 (5.94 krad/s^2 - 0.0350 krad/s)
CCM FL500	7.98 (4.53 krad/s^2 - 0.0298 krad/s)	1.37 (6.21 krad/s^2 - 0.0367 krad/s)	1.92 (5.34 krad/s^2 - 0.0349 krad/s)	5.82 (5.36 krad/s^2 - 0.0379 krad/s)
Bauer REAKT 200	6.25 (5.06 krad/s^2 - 0.0339 krad/s)	1.76 (5.75 krad/s^2 - 0.0310 krad/s)	2.09 (5.08 krad/s^2 - 0.0338 krad/s)	8.30 (4.46 krad/s^2 - 0.0319 krad/s)

Just like football, hockey helmet structure can also vary depending on the model. Generally, the structure consists of an outer shell made of hard plastic, impact-absorbing liners such as foams with single or multiple densities, and liners for comfort or any other functions. CCM FL40 is an example of the simplest hockey helmet design with just an

outer shell and foam for both comfort and absorbing impact. On the other hand, Bauer REAKT 200 showcases a more complex design with multi-density foam and "floating" liner. According to KRS and STAR rating system results, improving helmet design, which can be seen in CCM FL40 and Bauer REAKT 200 comparison study, can result in a better performing helmet.

All the tested hockey helmets' size is adjustable in the sagittal direction. The helmet design consists of two separate outer shells, generally coupled at the mid-coronal plane. Most of the coupling methodology involves metal fixation components. Since hockey helmets are designed to receive multiple impacts, the current outer shell design is problematic. It was found that after receiving some numbers of impacts, the integrity of the outer shell could be compromised, resulting in two moving shells that could potentially crush the user's head instead of protecting it during impact. In recent years, Warrior has come up with a helmet called Alpha One that consists of one outer shell. However, due to limited financial support, we have not tested the helmet mentioned above.

Cycling Helmet

Although six cycling helmets were tested, only four are available for comparison with the STAR rating system, which are Schwinn Excursion, Bontrager Solstice, Bontrager Ballista MIPS, Bontrager Spectre Wavecel. Table 4.27 showed the comparison between the KRS and the STAR rating system. Two out of six tested cycling helmets (Bontrager Spectre Wavecel and Bontrager Ballista MIPS) are rated five-star by the STAR rating while Schwinn Excursion and Bontrager Solstice are three-star, respectively. There are no available STAR rating for Alibaba and Specialized Covert. When tested according to KRS, Bontrager Spectre Wavecel and Bontrager Ballista MIPS received A+ grades while Schwinn Excursion and Bontrager Solstice received C grades. Most of the KRS results agree with the STAR rating system. According to KRS, Bontrager Spectre performed the best, while according to STAR rating system, Bontrager Ballista MIPS performed the best.

Table 4.27: Comparison of KRS and STAR rating system for the tested cycling helmets.

Helmet	*KRS Grade*	*KRS Rank*	*STAR Rating*	*STAR Rank*
Alibaba	**D (11.8)**	6	N/A	N/A
Specialized Covert	**C (16.4)**	5	N/A	N/A
Schwinn Excursion	**C (24.4)**	3	★★★ **(19.4)**	3
Bontrager Solstice	**C (22.2)**	4	★★★ **(19.9)**	4
Bontrager Spectre Wavecel	**A+ (59.2)**	2	★★★★★ **(10.8)**	1
Bontrager Ballista MIPS	**A+ (65.3)**	1	★★★★★ **(10.9)**	2

Table 4.28 and Table 4.29 present the detailed linear and rotational results of all tested cycling helmets. Based on the AUC calculation of linear and rotational acceleration, Bontrager Spectre is inferior compared to Bontrager Ballista MIPS. Moreover, the peak acceleration values of Bontrager Ballista MIPS were mostly lower. On the other hand, the performance of the Schwinn Excursion and Bontrager Solstice was also similar. Both helmets received a C grade.

Table 4.28: Linear grade comparison of different cycling helmets showing the linear grade while the linear acceleration (g) and AUC linear (m/s) are shown inside the parentheses.

Helmet	***Location***				
	1	***2***	***3***	***4***	***5***
Alibaba	3.83 (176.50 g - 6.37 m/s)	2.92 (186.71 g - 6.77 m/s)	3.20 (152.82 g - 6.67 m/s)	3.15 (177.71 g - 5.74 m/s)	2.35 (190.21 g - 6.98 m/s)
Covert	4.38 (155.48 g - 6.31 m/s)	3.11 (174.30 g - 6.81 m/s)	2.77 (167.46 g - 6.92 m/s)	3.43 (164.89 g - 5.68 m/s)	3.25 (148.42 g - 6.48 m/s)
Excursion	5.62 (117.89 g - 6.49 m/s)	3.27 (153.76 g - 6.98 m/s	3.43 (143.38 g - 6.68 m/s)	3.41 (157.25 g - 5.98 m/s)	3.02 (151.42 g - 6.86 m/s)
Solstice	5.56 (123.57 g - 6.26 m/s)	3.39 158.03 g - 6.87 m/s)	4.54 (116.83 g - 6.17 m/s)	3.31 (164.78 g - 5.91 m/s)	3.84 (126.11 g - 6.44 m/s)
Spectre	6.17 (109.79 g - 6.35 m/s)	5.12 (108.53 g - 6.62 m/s)	5.48 (94.51 g - 6.32 m/s)	4.93 (114.03 g - 5.70 m/s)	4.57 (105.99 g - 6.45 m/s)
Ballista	6.45 (106.28 g - 6.27 m/s)	5.46 (103.68 g - 6.52 m/s)	5.75 (95.02 g - 5.97 m/s)	4.45 (145.98 g - 4.94 m/s)	4.26 (117.32 g - 6.24 m/s)

Table 4.29: Rotational grade comparison of different cycling helmets showing the linear grade while the Rotational acceleration (krad/s^2) and AUC Rotational (krad/s) are shown inside the parentheses.

Helmet	*Location*				
	1	*2*	*3*	*4*	*5*
Alibaba	1.24 (12.15 krad/s^2 - 0.0371 krad/s)	3.29 (12.49 krad/s^2 - 0.0400 krad/s)	5.42 (8.00 krad/s^2 - 0.0258 krad/s)	4.59 (22.40 krad/s^2 - 0.0585 krad/s)	4.90 (12.17 krad/s^2 - 0.0551 krad/s)
Covert	1.24 (12.03 krad/s^2 - 0.0376 krad/s)	3.87 (10.71 krad/s^2 - 0.0397 krad/s)	3.61 (10.52 krad/s^2 - 0.0307 krad/s)	5.67 (19.34 krad/s^2 - 0.0546 krad/s)	6.82 (10.73 krad/s^2 - 0.0453 krad/s)
Excursio n	2.40 (7.06 krad/s^2 - 0.0327 krad/s)	3.59 (11.57 krad/s^2 - 0.0400 krad/s)	3.99 (8.89 krad/s^2 - 0.0316 krad/s)	5.81 (18.94 krad/s^2 - 0.0545 krad/s)	11.22 (8.32 krad/s^2 - 0.0359 krad/s)
Solstice	2.07 (8.01 krad/s^2 - 0.0336 krad/s)	4.56 (8.68 krad/s^2 - 0.0390 krad/s)	4.94 (6.94 krad/s^2 - 0.0326 krad/s)	5.02 (20.52 krad/s^2 - 0.0583 krad/s)	7.68 (8.73 krad/s^2 - 0.0488 krad/s)
Spectre	3.51 (5.59 krad/s^2 - 0.0282 krad/s)	8.59 (5.48 krad/s^2 - 0.0347 krad/s)	11.32 (4.37 krad/s^2 - 0.0223 krad/s)	9.07 (13.53 krad/s^2 - 0.0487 krad/s)	17.50 (5.63 krad/s^2 - 0.0331 krad/s)
Ballista	9.00 (3.09 krad/s^2 - 0.0195 krad/s)	6.26 (7.80 krad/s^2 - 0.0336 krad/s)	9.84 (4.05 krad/s^2 - 0.0277 krad/s)	9.02 (14.79 krad/s^2 - 0.0448 krad/s)	20.15 (5.12 krad/s^2 - 0.0317 krad/s)

The AUC contains information about the duration of the respective impact. Impact severity is better detemined when both the peak and duration of the acceleration pulse are considered in the performance evaluation of a helmet. By including the AUC in the KRS, a helmet is graded based on the peak and duration of acceleration pulse. Therefore, KRS provides a better estimate of a helmet's ability to reduce impact severity.

Both KRS and the STAR rating system use similar equipment. Some differences between the test methods in the KRS and STAR rating system are the dummy headform, the selection of the location of impact, and the impact speed. As seen in Table 4.30, the STAR rating system calls for a helmet to be tested on six impact locations and two speed levels. At each test scenario, the helmet is required to be tested twice, producing a total of 24 impacts per helmet model (Bland, et al., 2018). Only two data points per impact scenario are obtained. Two data points may not provide a reasonable estimation of the actual value and therefore, may result in lower result accuracy.

Table 4.30: KRS and STAR rating system test methods comparison

	KRS	*STAR Rating System*
Test Rig	Suspension-based oblique impact test rig	Oblique impact test rig with basket frame
Dummy Headform	Hybrid III 50th percentile headform without neck	NOCSAE headform without neck
# Impact Location	5	6
Impact Speed	6.5 m/s	4.8 and 7.3 m/s

On the other hand, the KRS calls for a helmet to be tested on five impact locations at one speed. At each test scenario, the helmet is required to be tested three times, producing a total of 15 impacts per helmet model. With three data points per impact scenario, the KRS can have better accuracy. The number of required data points in the KRS is significantly lower, which may result in lower testing cost.

4.2.2. KRS Impact Speed

KRS method requires the helmet to be tested at one impact speed (6.5 m/s), while STAR Rating system requires two. The KRS impact speed is higher than the speeds recommended in CSA, CPSC, and NOCSAE standards to represent more severe impacts. Some research studies suggested that a helmet that performs well in severe impacts is considered protective for all impacts of equal or lesser severity (Whyte, et al., 2019; Becker, et al., 2015). Studies have shown linearity of impact response (linear acceleration) versus impact severity (impact speed) for up to a certain speed (DeMarco, et al., 2010; DeMarco, et al., 2016; Rowson, et al., 2013). The linearity only continues for up to the helmet liner stops buckling or starts to bottom out and loses its protective capability. Most of the tested helmets have not reached the point where they lose their protective capability when tested with an impact speed of 6.5 m/s. There is no report regarding the relationship of impact speed with rotational acceleration and rotational velocity. By graphing and analyzing results shown in COST 327 report (Chinn, et al., 2001) for various motorcycle helmets, the relationship between impact speed and rotational acceleration and rotational velocity was shown to be linear, as seen in Figure 4.3. The R^2 values for the rotational acceleration of four helmet models were between

0.9903 and 0.9991. For the rotational velocity, the R^2 values were between 0.8995 and 0.9952. However, the COST 327 did not report peak linear acceleration.

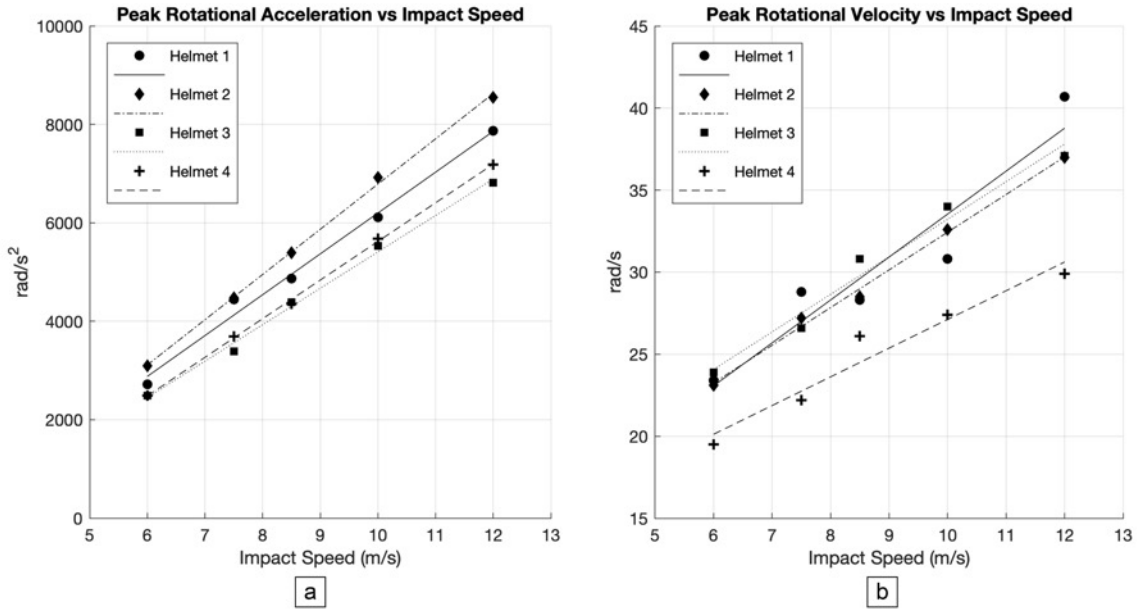


Figure 4.3: Plot of speeds versus (a) rotational acceleration, (b) rotational velocity of four motorcycle helmets from COST 327 report (Chinn, et al., 2001).

Furthermore, results of a football helmet oblique impact test at different speeds and impact locations (performed in-house) indicated that the relationship between impact speed and helmet kinematic response (linear acceleration, rotational acceleration, and rotational velocity) was linear as shown in Figure 4.4. The R^2 values for each fit line are summarized in Table 4.31. Considering the aforementioned studies, performing oblique impact test at one velocity that is considered severe seems to be sufficient.

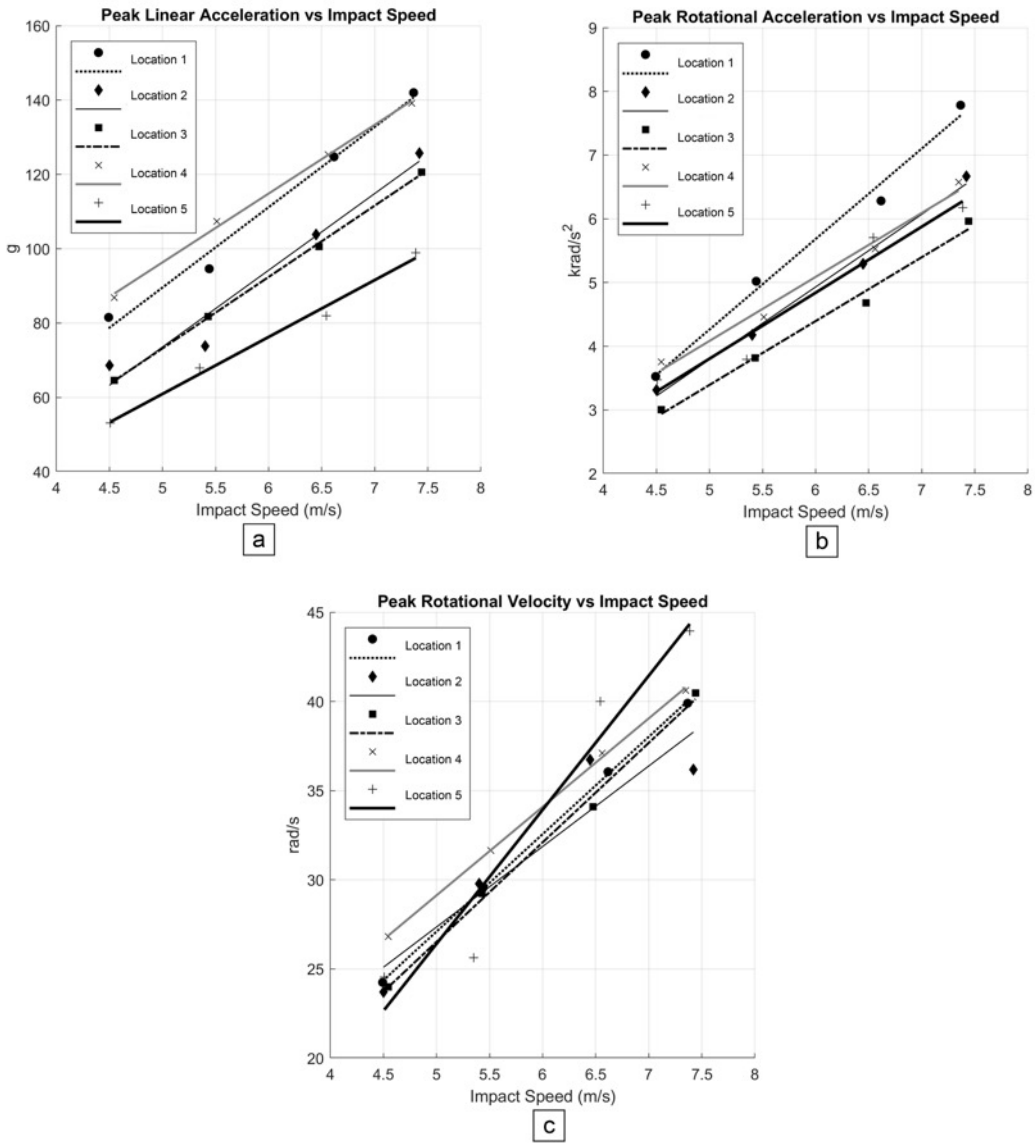


Figure 4.4: Plot of impact speeds versus (a) linear acceleration, (b) rotational acceleration, and (c) rotational velocity, fitted with a polynomial degree 1.

Table 4.31: R^2 values of the best fit lines of the response

	Lin. Acc.	Rot. Acc.	Rot. Vel.
Impact Loc.	R^2	R^2	R^2
1	0.9867	0.9879	0.9997
2	0.9545	0.9924	0.8683
3	0.9993	0.9883	0.9956
4	0.9973	0.9855	0.9993
5	0.9899	0.9505	0.9347

4.3. Summary

The KRS test method evaluates a helmet's ability to reduce both the tangential and normal components of the impact force. The STAR rating system for football and hockey, does not take into account the helmet's ability to reduce tangential forces.

Four football, four hockey, and six cycling helmets were tested according to the test methods defined in the previous section. The results showed that helmet structure plays a key role in determining the helmet kinematic response to both compression and tangential forces. Furthermore, the results were compared with the STAR rating system. The comparison showed some disagreements.

Chapter 5.

Conclusion and Recommendation

5.1. Conclusion

A new kinematic-based helmet grading method, the Kinematic Rating System (KRS), was proposed and it was developed for football, hockey, and cycling helmet. The grading method evaluates a helmet based purely on its impact attenuating capability. Prediction of risk of injury is based on factors that are uniquely dependant on the subject (the hemet wearer), and rating a helmet based on such information leave a lot of room for error. KRS result does not predict the occurrence of head injury and concussion, and only provide information on how well a helmet performs. This makes KRS more objective and non-biased. Also, KRS takes into account all all the crucial factors of a helmet response such as acceleration and the duration of an impact scenario. This allows the KRS evaluation of a helmet to be accurate, and only based on factors related to the helmet.

A series of oblique impact tests were performed on a dummy headform equipped with the helmet of interest. The impact tests were specifically designed to simulate severe real-life impacts. The tests were conducted using the drop rig described in Section 3.2. The test methods are summarized in Table 5.1.

Table 5.1: Summary of the test methodology developed for cycling, football, and hockey helmet.

Helmet Type	Impact Opponent (Anvil)	Anvil Angle	Impact Speed	# Helmet required	# Impact Locations per Helmet
Cycling	Steel covered with 80-grit Aluminum Oxide sandpaper	45°	6.5 m/s	3	5
Football	Modular Elastomer Programmer (MEP) Pad	45°	6.5 m/s	1	4
Hockey	Modular Elastomer Programmer (MEP) Pad	45°	6.5 m/s	1	4

KRS required that the kinematic parameters, linear and rotational acceleration, of the dummy headform center of gravity during impact are recorded. From the recorded data of each acceleration, more information is extracted: peak value and Area Under the Curve (AUC) above a certain threshold. The extracted information is used as the inputs of the grading equations.

KRS provides a holistic test method that takes into account a helmet's ability to reduce the tangential and normal component of the impact force. Other test methods, such as in STAR for football and hockey, do not highlight the tangential component of the impact force. Therefore, when compared to KRS, the results were not in sync. In other words, KRS exposes the blind spot of modern helmet testing methods. KRS can be used by the customer to make an informed decision before purchasing any helmets and by helmet designer to design a better performing helmet. In summary, the main benefits of KRS are:

- Purely based on helmet's kinematic response (not predicting injury)
- Takes into account both the magnitude and duration of an acceleration pulse when rating a helmet
- Takes into account a helmet's ability to reduce the tangential and normal component of the impact force

5.2. Recommendation and Future Work

Due to limited budget, KRS was validated by testing four football, four hockey, and six cycling helmets only. KRS needs to be validated by testing more football, hockey, and cycling helmets. Furthermore, KRS can be developed for other types of helmet such as, but not limited to, motorcycle, ski, and military.

The speed of impact for the current version of the KRS was chosen to be 6.5 m/s. Test with one other speeds may be performed, possibly one lower than the current 6.5 m/s. By testing for two different speed levels, the helmet behaviour at the speeds in between can be predicted. This will result in more comprehensive helmet performance analysis. Moreover, if there would be more supporting evidence, impact test on different anvil angles may also be performed.

Currently, there is no way of quantifying how much a kinematic parameter is more important than the other. When more information becomes available, the grade equation can include weighing factors $\boldsymbol{A} - \boldsymbol{L}$, as seen in (5.5.1) – (5.5.2).

$$\boldsymbol{Linear\ Grade} = \sum \boldsymbol{E(n)} \times \left(\boldsymbol{A}\frac{\boldsymbol{a_{,baseline}(n)}}{\boldsymbol{a_{measured}(n)}}\right)^{\boldsymbol{G}} \times \left(\boldsymbol{B}\frac{\boldsymbol{AUCL_{baseline}(n)}}{\boldsymbol{AUCL_{measured}(n)}}\right)^{\boldsymbol{H}} \quad \textbf{(5.5.3)}$$

$$\boldsymbol{Rotational\ Grade} = \sum \boldsymbol{E(n)} \times \left(\boldsymbol{C}\frac{\boldsymbol{\alpha_{,baseline}(n)}}{\boldsymbol{\alpha_{measured}(n)}}\right)^{\boldsymbol{I}} \times \left(\boldsymbol{D}\frac{\boldsymbol{AUCR_{baseline}(n)}}{\boldsymbol{AUCR_{measured}(n)}}\right)^{\boldsymbol{J}} \quad \textbf{(5.5.4)}$$

$$\boldsymbol{Final\ Grade} = (\boldsymbol{E} \times \boldsymbol{Linear\ Grade})^{\boldsymbol{K}} \times (\boldsymbol{F} \times \boldsymbol{Rotational\ Grade})^{\boldsymbol{L}} \quad \textbf{(5.5.5)}$$

References

Aare, M. & Halldin, P., 2003. A new laboratory rig for evaluating helmets subject to oblique impacts. *Traffic Injury Prevention,* 4(3), pp. 240-248.

Adanty, K., Clark, J. M., Post, A. & Gilchrist, M. D., 2019. Comparing Two Proposed Protocols to Test the Oblique Response of Cycling Helmets to Fall Impacts. *International Journal of Crashworthiness.*

Albrecht, J. S. et al., 2016. Increased Rates of Mild Traumatic Brain Injury Among Older Adults in US Emergency Departments, 2009-2010. *J Head Trauma Rehabil,* 31(5), pp. E1-7.

Allison, M. A. et al., 2017. Validation of a helmet-based system to measure head impact biomechanics in ice hockey. *Med Sci Sports Exerc,* 46(1), pp. 115-123.

Alosco, M. L. et al., 2018. Age of First Exposure to Tackle Football and Chronic Traumatic Encephalopathy. *American Neurological Association,* Volume 83, pp. 886-901.

ASTM International, 2015. *Standard Performance Specification for Ice Hockey Helmets,* West Conshohocken, PA, Standard No. F1045-15.: ASTM International.

Attewell, R. G., Glase, K. & McFadden, M., 2001. Bicycle helmet efficecy: a meta-analysis. *Accident Analysis and Prevention,* Volume 33, pp. 345-352.

Bambach, M. R., Mitchell, R. J., Grzebieta, R. H. & Olivier, J., 2013. The effectiveness of helmets in bicycle collisions with motor vehicles: A case–control study. *Accident Anlaysis and Prevention,* Volume 53, pp. 78-88.

Bandte, A. et al., 2017. Sport-Related Traumatic Brain Injury with and without Helmets in Children. *World Neurosurg.*

Bartsch, A. et al., 2012. Hybrid III anthropomorphic test device (ATD) response to head impacts and potential implications for athletic headgear testing. *Accident Analysis and Prevention,* Volume 48, pp. 285-291.

Becker, E. B., Anishchenko, D. V. & Palmer, S. B., 2015. *Motorcycle Helmet Impact Response at Various Levels of Severity for Different Standard Certifications.* Lyon, France, IRCOBI Conference.

Beckwith, J. G. et al., 2013. Head Impact Exposure Sustained Bby Football Players on Days of Diagnosed Concussion. *Med Sci Sports Exerc,* April, 45(4), pp. 737-746.

Benson, B. W. et al., 2009. Is protective equipment useful in preventing concussion? A systematic review of the literature. *Br J Sports Med,* Volume 43, pp. i56-i57.

Benson, B. W. et al., 2011. A prospective study of concussions among National Hockey League players during regular season games: the NHL- NHLPA Concussion Program. *CMAJ,* pp. 905-911.

Beusenberg, M. et al., 2001. *Head, Neck, and Body Coupling in Reconstructions of Helmeted Head Impacts.* Isle of Man, UK, IRCOBI Conference.

Blackman, E. G., Hale, M. E. & Lisanby, S. H., 2007. *Improving TBI Protection Measures and Standards for Combat Helmets.* Defense Science Study Group paper, Section 3.3.5.

Bland, M. L., McNally, C. & Rowson, S., 2018. Differences in Impact Performance of Bicycle Helmets During Oblique Impacts. *J Biomech Eng,* 140(9).

Bland, M. L., McNally, C. & Rowson, S., 2018. *Headform and Neck Effects on Dynamic Response in Bicycle Helmet Oblique Impact Testing.* Athens, Greece, IRCOBI Conference.

Bland, M. L., McNally, C. & Rowson, S., 2018. *STAR Methodology for Bicycle Helmets,* Blacksburg, VA: Virginia Tech, Virginia Tech Helmet Lab.

Bliven, E. et al., 2019. Evaluation on a Novel Bicycle Helmet Concept in Oblique Impact Testing. *Accident Analysis and Prevention,* Volume 124, pp. 58-65.

Bourdet, N., Deck, C. & Carreira, R. P., 2012. Head Impact Conditions in the Case of Cyclist Falls. *Journal of Sports Engineering and Technology,* 226(3-4).

Bourdet, N. et al., 2014. In-depth Real-World Bicycle Accident Reconstructions. *International Journal of Crashworthiness,* 19(3), pp. 222-232.

Brain Injury Canada, n.d. *Brain Injury Canada: Education, Awareness, Advocacy.* [Online]
Available at: https://www.braininjurycanada.ca/
[Accessed 10 02 2018].

Brain Injury Society of Toronto, 2018. *Brain Injury Facts & Figures.* [Online]
Available at: http://www.bist.ca/brain-injury-facts-figures/
[Accessed 10 02 2018].

Brainard, L. L. et al., 2012. Gender Differences in Head Impact Sustained by Collegiate Ice Hockey Players. *Med Sci Sports Exerc.,* 44(2), pp. 297-304.

Breedlove, K. M., Breedlove, E. L., Bowman, T. G. & Nauman, E. A., 2016. Impact attenuation capabilities of football and lacrosse helmets. *Journal of Biomechanics,* Volume 49, pp. 2838-2844.

Broglio, S. P. et al., 2010. The Biomechanical Properties of Concussions in High School Football. *Med Sci Sports Exerc,* November, 42(11), pp. 2064-2071.

Brolinson, P. G. et al., 2006. Analysis of Linear Head Accelerations from Collegiate Football Impacts. *Current Sports Medicine Reports,* January, 5(1), pp. 23-28.

BS 6658:1985, 1985. *Protective Helmets for Vehicle Users,* London: British Standard Institution.

Cadex Inc., 2019. *EN960 - Full Headform - Magnesium K1A.* [Online] Available at: http://www.cadexinc.com/en960_full_magnesium_k1a_headform.php [Accessed 01 07 2019].

Camacho, D. L. et al., 1997. Experimental Flexibility Measurements for the Development of a computational Head-Neck Model Validated for Near-Vertex Head Impact. *SAE 973345.*

Cantu, R. C., 1996. Head injuries in sports. *Br J Sports Med,* Volume 30, pp. 289-296.

Ching, R. P. et al., 1997. Damage to Bicycle Helmets Involved with Crashes. *Accident Analysis and Prevention,* 29(5), pp. 555-562.

Chinn, B. et al., 2001. *COST 327 - Motorcycle Safety Helmet,* Luxembourg: Office for Official Publications of the European Communities.

Clark, J. M., Post, A., Hoshizaki, T. B. & Gilchrist, M. D., 2016. Protective Capacity of Ice Hockey Helmets against Different Impact Events. *Annals of Biomedical Engineering,* 44(12), pp. 3693-3704.

Cobb, B. R. et al., 2015. Quantitative Comparison of Hybrid III and NOCSAE Headform Shape Characteristics and Implications on Football Helmet Fit. *Proc. Inst. Mech. Eng., Part P,* pp. 39-46.

Connor, T. A. et al., 2016. *Current Standards for Sports and Automotive Helmets: A Review.* Ref. Ares 3151745.

Consumer Product Safety Commission, 1998. *Safety Standard for Bicycle Helmets; Final Rule, 16 CFR Part 1203,* US: Consumer Product Safety Commission.

Conte Jaswal, 2017. *Brain Injury Statistics Canada: The Shocking Truth.* [Online] Available at: http://contelawyers.ca/brain-injury-statistics-canada-2017/ [Accessed 10 02 2018].

Cripton, P. A. et al., 2014. Bicycle helmets are highly effective at preventing head injury during head impact: Head-form accelerations and injury criteria for helmeted and unhelmeted impacts. *Accident Analysis and Prevention,* Volume 70, pp. 1-7.

Crisco, J. J. et al., 2010. Frequency and Location of Head Impact Exposures in Individual Collegiate Football Players. *Journal of Athletic Training,* 45(6), pp. 549-559.

Crisco, J. J. et al., 2012. Magnitude of Head Impact Exposures in Individual Collegiate Football Players. *Journal of Applied Biomechanics,* Volume 28, pp. 174-183.

CSA, 2015. *Ice Hockey Helmets,* Toronto, ON, Canada, Standard No. Z262.1-15: Canadian Standards Association.

Cusimano, M. D. et al., 2013. Mechanisms of Team-Sport-Related Brain Injuries in Children 5 to 19 Years Old: Opportunities for Prevention. *PLOS ONE,* 8(3).

Daniel, R. W., Rowson, S. & Duma, S. M., 2012. Head impact exposure in youth football. *Annals of Biomedical Engineering,* 40(4), pp. 976-981.

Deck, C. et al., 2012. *Proposal of an improved bicycle helmet standards.* Politecnico - Milano, s.n.

Deck, C. & Willinger, R., 2008. Improved head injury criteria based on head FE model. *Int J Crash,* Volume 13, pp. 667-679.

Delaney, S., Lacroiz, V., Leclerc, S. & Johnston, K., 2002. Concussion among university football and soccer players. *Clin J Sport Med,* Volume 12, pp. 331-338.

DeMarco, A. L., Chimich, D. D., Gardiner, J. C. & Nightingale, R. W., 2010. The impact response of motorcycle helmets at different impact severities. *Accident Analysis and Prevention,* Volume 42, pp. 1778-1784.

DeMarco, A. L., Chimich, D. D., Gardiner, J. C. & Siegmund, G. P., 2016. The impact response of traditional and BMX-style bicycle helmets at different impact severities. *Accident Analysis and Prevention,* Volume 92, pp. 175-183.

Denny-Brown, D. & Russell, R., 1941. Brain. *Experimental Cerebral Concussion,* Volume 64, pp. 93-164.

Depreitere, B. et al., 2004. Bicycle-related Head Injury: A Study of 86 Cases. *Accident Analysis and Prevention,* Volume 36, pp. 561-567.

Ebrahimi, I., Golnaraghi, F. & Wang, G., 2015. Factors influencing the oblique impact test of motorcycle helmets. *Traffic Injury Prevention,* 16(4), pp. 404-408.

Elliott, J., Anderson, R., Collins, S. & Heron, N., 2019. Sports-related concussion (SRC) assessment in road cycling: a systematic review and call to action. *BMJ Open Sport & Exercise Medicine,* Volume 5, p. e000525.

Faul, M., Xu, L., Wald, M. M. & Coronado, V. G., 2010. *Traumatic Brain Injury in the United States: Emergency Department Visits, Hospitalizations and Deaths 2002-2006,* s.l.: National Center for Injury Prevention and Control, Centers for Disease Control and Prevention.

Finan, J. D., Nightingale, R. W. & Myers, B. S., 2008. The Influence of Reduced Friction on Head Injury Metrics in Helmeted Head Impacts. *Traffic Injury Prevention,* 9(5), pp. 483-488.

Folksam, 2019. *Bicycle Helmets 2019 Tested by Folksam,* Stockholm, Sweden: Folksam.

Forbes, A. E., Schutzer-Weissmann, J., Menassa, D. A. & Wilson, M. H., 2017. Head injury patterns in helmeted and non- helmeted cyclists admitted to a London Major Trauma Centre with serious head injury. *PLoS ONE,* 12(9), p. e0185367.

Funk, J. R., Duma, S. M., Manoogian, S. J. & Rowson, S., 2007. Biomechanical Risk Estimates for Mild Traumatic Brain Injury. *Annu. Proc. Assoc. Adv. Automot. Med.,* Volume 51, pp. 343-361.

Gadd, C. W., 1966. Use of a Weighted-Impulse Criterion for Estimating Injury Hazard. *10th Stapp, SAE,* pp. 164-174.

Gennarelli, T. A. et al., 1982. Diffuse Axonal Injury and Traumatic Coma in the Primate. *Annals of Neurology,* 12(6), pp. 564-574.

Gennarelli, T., Thibault, L. & Ommaya, A., 1972. Pathophysiologic Responses to Rotational and Translational Accelerations of the Head. *SAE Technical Paper 720970.*

Gennarelli, T. et al., 1987. Directional dependence of axonal brain injury due to centroidal and non-centroidal acceleration. *SAE Technical Paper 872197.*

Ghajari, M., Peldschus, S., Galvanetto, U. & Iannucci, L., 2013. Effects of the Presence of the Body in Helmet Oblique Impacts. *Accident Analysis and Prevention,* Volume 50, pp. 263-271.

Giza, C. C. & Hovda, D. A., 2001. The Neurometabolic Cascade of Concussion. *Journal of Athletic Training,* 36(3), pp. 228-235.

Gough, C., 2018. *Total number of registered ice hockey players in Canada from 2010/11 to 2017/18.* [Online]
Available at: https://www.statista.com/statistics/282125/number-of-registered-ice-hockey-players-in-canada/
[Accessed 08 07 2019].

Gough, C., 2018. *Total number of registered ice hockey players in the United States from 2010/11 to 2017/18.* [Online]
Available at: https://www.statista.com/statistics/282122/number-of-registered-ice-hockey-players-in-the-united-states/
[Accessed 08 07 2019].

Greenwald, R. M., Gwin, J. T., Chu, J. J. & Crisco, J. J., 2008. Head Impact Severity Measures for Evaluating Mild Traumatic Brain Injury Risk Exposure. *Neurosurery,* 62(4), pp. 789-798.

Gurdjian, E., Roberts, V. & Thomas, L., 1966. Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. *J Trauma,* Volume 6, pp. 600-604.

Guskiewicz, K. M. et al., 2007. Recurrent Concussion and Risk of Depression in Retired Professional Football Players. *Medicine & Science in Sports & Exercise.*

Guskiewicz, K. M. et al., 2003. Cumulative Effects Associated With Recurrent Concussion in Collegiate Football Players. *JAMA,* 290(19), pp. 2549-2555.

Guskiewicz, K. M. et al., 2007. Measurement of Head Impacts In Collegiate Football Players: Relationship Between Head Impact Biomechanics And Acute Clinical Outcome After Concussion. *Neurosurgery,* Volume 61, pp. 1244-1253.

Gwin, J. T., Chu, J. J., McAllister, T. W. & Greenwald, R. M., 2009. In situ Measures of Head Impact Acceleration in NCAA Division I Men's Ice Hockey: Implications for ASTM F1045 and Other Ice Hockey Helmet Standards. *Journal of ASTM International,* 6(6).

Gwin, J. T. et al., 2009. An Investigation of the NOCSAE Linear Impactor Test Method Based on In Vivo Measures of Head Impact Acceleration in American Football. *Journal of Biomechanical Engineering,* 132(1).

Halldin, P., 2015. *CEN/TC 158 Working Group 11 Rotational Test Methods-Proposal for a New Test Method Measuring the Kinematics in Angled Helmeted Impacts,* s.l.: s.n.

Halldin, P., Gilchrist, A. & Mills, N. J., 2001. A New Oblique Impact Test for Motorcycle Helmets. *International Journal of Crashworthiness,* 6(1), pp. 53-64.

Halldin, P. & Kleiven, S., 2013. *The Development of Next Generation Test Standards for Helmets.* London, s.n.

Herbst, B., Forrest, S. & Chang, D., 1998. *Fidelity of Anthropometric Test Dummy Necks in Rollover Accidents.* Windsor, ON, Canada, s.n.

Hitosugi, M. et al., 2014. Biomechanical analysis of acute subdural hematoma resulting from judo. *Biomedical Research (Tokyo),* 35(5), pp. 339-344.

Hodgson, V. & Thomas, L., 1971. Comparison of Head Acceleration Injury Indices in Cadaver Skull Fracture. *SAE Technical Paper 710854.*

Holburn, A. H. S., Edin, M. A. & Oxfd, D. P., 1943. Mechanics of Head Injuries. *The Lancet,* Volume 242, pp. 438-441.

Hootman, J. M., Dick, R. & Agel, J., 2007. Epidemiology of Collegiate Injuries for 15 Sports: Summary and Recommendations for Injury Prevention Initiatives. *Journal of Athletic Training,* 42(2), pp. 311-319.

Hoshizaki, B., Vassilyadi, M., Post, A. & Oeur, A., 2012. Performance analysis of winter activity protection headgear for young children. *J Neurosurg Pediatrics,* Volume 9, pp. 133-138.

Hoshizaki, T. et al., 2017. The development of a threshold curve for the understanding of concussion in sport. *Trauma,* 19(3), pp. 196-206.

Hutchison, M. G., Comper, P., Meeuwisse, W. H. & Echemendia, R. J., 2015. A systematic video analysis of National Hockey League (NHL) concussions, part I: who, when, where and what?. *Br J Sports Med,* Volume 49, pp. 547-551.

Ivancic, P. C., 2014. Cervical spine instability following axial compression injury: A biomechanical study. *Orthopaedics & Traumatology: Surgery & Research,* Volume 100, pp. 127-133.

Iverson, G. L., Gaetz, M., Lovell, M. R. & Collins, M. W., 2004. Cumulative Effects of Concussion in Amateur Athletes. *Brain Injury,* 18(5).

Izraelski, J., 2014. Concussions in the NHL: A narrative review of the literature. *J Can Chiropr Assoc,* 58(4), pp. 346-352.

Jadischke, R. et al., 2013. On the accuracy of the Head Impact Telemetry (HIT) System used in football helmets. *Journal of Biomechanics,* Volume 46, pp. 2310-2315.

Jadischke, R., Viano, D. C., McCarthy, J. & King, A. I., 2016. The effects of helmet weight on Hybrid III head and neck responses by comparing unhelmeted and helmeted impacts. *Journal of Biomechanical Engineering,* 138(10).

Ji, S., Li, Z., Zhao, W. & McAllister, T., 2014. Head Impact Accelerations for Brain Strain-Related Responses in Contact Sports: A Model-Based Investigation. *Biomechanics and Modeling in Mechanobiology,* Volume 13, pp. 1121-1136.

Kendall, M., Post, A. & Gilchrist, M. D., 2012. *A Comparison of dynamic impact response and brain deformation metrics within the cerebrum of head impact reconstructions representing three mechanisms of head injury in ice hockey.* s.l., IRCOBI Conference.

Kimpara, H. & Iwamoto, M., 2012. Mild Traumatic Brain Injury Predictors Based on Angular Accelerations During Impacts. *Annals of Biomedical Engineering,* 40(1), pp. 114-126.

King, A. I., Yang, K. H., Zhang, L. & Hardy, W., 2003. *Is Head Injury Caused by Linear or Angular Acceleration?.* Lisbon, Portugal, s.n.

Kleiven, S., 2006. Evaluation of head injury criteria using a finite element model validated against experiments on localized brain motion, intracerebral acceleration, and intracranial pressure. *IJCrash,* 11(1), pp. 65-79.

Kleiven, S., 2007. Predictors for Traumatic Brain Injuries Evaluated through Accident Reconstructions. *Stapp Car Crash Journal,* Volume 51.

Kleiven, S., 2007. Predictors for traumatic brain injuries eveluated through accident reconstructions. *Stapp Car Crash Journal,* Volume 51.

Kleiven, S., 2013. Why Most Traumatic Brain Injuries are Not Caused by Linear Acceleration but Skull Fractures are. *Frontiers in Bioengineering and Biotechnology,* Volume 1, pp. 1-5.

Kuramochi, R. et al., 2004. Anticipatory modulation of neck muscle reflex responses induced by mechanical perturbations of the human forehead. *Neuroscience letters,* Volume 366, pp. 206-210.

Levy, M. L. et al., 2004. Analysis and Evolution of Head Injury in Football. *Neurosurgery,* 55(3), pp. 649-655.

Lock, S., 2019. *https://www.statista.com/statistics/191658/participants-in-tackle-football-in-the-us-since-2006/.* [Online]
Available at: https://www.statista.com/statistics/191658/participants-in-tackle-football-in-the-us-since-2006/
[Accessed 08 07 2019].

MacAlister, A., 2013. Surrogare Head Forms for the Evaluation of Head Injury Risk. *Brain Injuries and Biomechanics.*

Martin, A. et al., 2018. Are head injuries to cyclists an important cause of death in roadtravel fatalities?. *Journal of Transport & Health,* Volume 10, pp. 178-185.

McCrea, M. et al., 2004. Unreported Concussion in High School Football Players. *Clin J Sport Med,* Volume 14, pp. 13-17.

McElhaney, J. H., Roberts, V. L., Hilyard, J. F. & Kenkyūjo, N. J., 1976. Properties of human tissues and components: nervous tissues. In: *Handbook of human tolerance.* Tokyo: Japan, p. 143.

McHenry, B. G., 2004. Head Injury Criterion and the ATB. *Proc. 2004 ATB Users' Conf.*

McIntosh, A., Dowdell, B. & Svensson, N., 1998. Pedal Cycle Helmet Effectiveness: A Field Study of Pedal Cycle Accidents. *Accident Analysis and Prevention,* 30(2), pp. 161-168.

McIntosh, A. S., Andersen, T. E. & Bahr, R., 2011. Sports Helmets Now and in the Future. *Br J Sports Med,* Volume 45, pp. 1258-1265.

McIntosh, A. S., Lai, A. & Schilter, E., 2013. Bicycle helmets: head impact dynamics in helmeted and unhelmeted oblique impact tests. *Traffic Injury Prevention,* 14(5), pp. 501-508.

McKee, A. C. et al., 2009. Chronic Traumatic Encephalopathy in Athletes: Progressive Tauopathy After Repetitive Head Injury. *J Neuropathol Exp Neurol,* 68(7), pp. 709-735.

McLean, A. & Anderson, R. W., 1997. Biomechanics of Closed Head Injury. In: Head Injury. London: Chapman & Hall.

Meaney, D. F. & Smith, D. H., 2011. Biomechanics of Concussion. *Clin Sports Med,* 30(1), pp. 19-vii.

Melo, N., Berg, R. J. & Inaba, K., 2014. Injuries sustained by bicyclists. *Trauma,* 16(3), pp. 183-188.

Mez, J. et al., 2017. Clinicopathological Evaluation of Chronic Traumatic Encephalopathy in Players of American Football. *JAMA,* 318(4), pp. 360-370.

Mihalik, J. P. et al., 2012. Head Impact Biomechanics in Youth Hockey: Comparisons Across Playing Position, Event Types, and Impact Locations. *Annals of Biomedical Engineering,* 40(1), pp. 141-149.

Mills, N. J., 2010. Critical evaluation of the SHARP motorcycle helmet rating. *International Journal of Crashworthiness,* 15(3), pp. 331-342.

Mills, N. J. & Gilchrist, A., 1991. The Effectiveness of Foams in Bicycle and Motorcycle Helmets. *Accident Analysis and Prevention,* 23(2/3), pp. 153-163.

Mills, N. J. & Gilchrist, A., 2006. Bicycle helmet design. *Proc. IMechE,* Volume 220.

Mollayeva, T., El-Khecken-Richandi, G. & Colantonio, A., 2018. Sex & Gender Considerations in Concussion Research. *Concussion,* 3(1), p. CNC51.

Moritz, A. R., 1943. Mechanism of Head Injury. *Annals of Surgery,* 117(4), pp. 562-575.

Myers, B. S. et al., 1989. *Response of the Human Cervical Spine to Torsion.* SAE Technical Paper 892437.

Nathanson, J. T. et al., 2016. Concussion Incidence in Professional Football - Position-Specific Analysis With Use of a Novel Metric. *The Orthopaedic Journal of Sports Medicine,* 4(1), pp. 1-6.

Naunheim, R. et al., 2002. Does the Use of Artificial Turf Contribute to Head Injuries?. *J Trauma,* Volume 53, pp. 691-694.

Newman, J. A., 1966. *A Generalized Acceleration Model for Brain Injury Threshold (GAMBIT),* Ottawa, Canada: s.n.

Newman, J. A., Shewchenko, N. & Welbourne, E., 2000. A Proposed New Biomechanical Head Injury Assessment Function - The Maximum Power Index. *Stapp Car Crash Journal,* Volume 44.

NFLPA, 2019. *NFL, NFLPA RELEASE 2019 HELMET LABORATORY TESTING PERFORMANCE RESULTS.* [Online]
Available at: https://www.nflpa.com/news/nfl-nflpa-release-2019-helmet-laboratory-testing-performance-results
[Accessed 05 08 2019].

Nightingale, R. W., McElhaney, J. H., Richardson, W. J. & Myers, B. S., 1996. Dynamic Responses of the Head and Cervical Spine to Axial Impact Loading. *J. Biomechanics,* 29(3), pp. 307-318.

Noble, J. M. & Hesdorffer, D. C., 2013. Sport-Related Concussions: A Review of Epidemiology, Challenges in Diagnosis, and Potential Risk Factors. *Neuropsychol Rev,* Volume 23, pp. 273-284.

NOCSAE, DOC, 2016. *Standard Performance Specification for Newly Manufactured Ice Hockey Helmets,* Overland Park: National Operating Committee on Standards For Athletic Equipment.

NOCSAE, DOC, 2017. *Standard Test Method and Equipment Used in Evaluating The perforrmance Characteristics of Headgear/Equipment,* Overland Park: National Operating Committee on Standards for Athletic Equipment.

NOCSAE, DOC, 2019. *Standard Performance Specification for Newly Manufactured Football Helmets,* Overland Park: National Operating Committee on Standards For Athletic Equipment.

NOCSAE, DOC, 2019. *Standard Pneumatic Ram Test Method and Equipment Used in Evaluating the Performance Characteristics of Protective Headgear and Face Guards,* Overland Park: National Operating Committee on Standards For Athletic Equipment.

Oeur, R. A., Zanetti, K. & Hoshizaki, T. B., 2014. *Angular Acceleration Responses of American Football, Lacrosse and Ice Hockey Helmets Subject to Low-Energy Impacts.* s.l., s.n.

Olivier, J. & Creighton, P., 2017. Bicycle injuries and helmet use: a systematic review and meta-analysis. *International Journal of Epidemiology,* pp. 278-292.

Ommaya, A. K. & Gennarelli, T. A., 1974. Cerebral Concussion and Traumatic Unconsciousness. *Brain,* Volume 97, pp. 633-654.

Ommaya, A. K., Hirsch, A. E., Yarnell, P. & Harris, E. H., 1967. *Scaling of Experimental Data on Cerebral Concussion in Sub-human Primates to Concussion Threshold for Man.* SAE Technical Paper 670906.

Ono, K. et al., 2003. *Biomechanical response of the head, neck, and torso to direct impact on the back of male and female volunteers.* Lisbon, s.n.

O'Riodain, K., Thomas, P., Phillips, J. & Gilchrist, M., 2003. Reconstruction of real world head injury accidents resulting from falls using multibody dynamics. *Clinical Biomechanics,* 18(7), pp. 590-600.

Padgaonkar, A. J., Krieger, K. W. & King, A. I., 1975. Measurement of angular acceleration of a rigid body using linear accelerometers. *J. Appl. Mech.,* 42(3), pp. 552-556.

Pang, T. Y. et al., 2011. Head and neck responses in oblique motorcycle helmet impacts: a novel laboratory test method. *International Journal of Crashworthiness,* 16(3), pp. 297-307.

Parizek, A. & Ferraro, R., 2015. *Concussion in Ice Hockey,* Grand Forks, ND, USA: The American College of Sports Medicine.

Patton, D. A., McIntosh, A. S., Kleiven, S. & Frechede, B., 2012. Injury data from unhelmeted football head impacts evaluated against critical strain tolerance curves. *J Sports Engineering and Technology,* 226(3/4), pp. 177-184.

Pellman, E. J. et al., 2003. Concussion in professional football: location and direction of helmet impacts - part 2. *Neurosurgery,* 5(6), pp. 1328-1341.

Pellman, E. J. et al., 2003. Concussion in professional football: reconstruction of game impacts and injuries - part 1. *Neurosurgery,* 53(4), pp. 799-814.

Pellman, E. J. et al., 2016. Concussion in professional football: helmet testing to assess impact performance - part 11. *Neurosurgery,* 58(1), pp. 78-95.

Peng, Y. et al., 2012. A Study of Pedestrian and Bicyclist Exposure to Head Injury in Passenger Car Collisitons Based on Accident Data and Simulations. *Safety Science,* Volume 50, pp. 1749-1759.

Petersen, P. G., 2018. The Effect of Surface Friction on Oblique Bicycle Helmet Impacts. In: *Masters Dissertation.* School of Mechanical and Material Engineering, Washington State Univ., Pullman: Washington State Univ.

Post, A. & Hoshizaki, T. B., 2012. Mechanism of brain impact injuries and their prediction: a review. *Trauma,* 0(0), pp. 1-23.

Post, A. & Hoshizaki, T. B., 2015. Rotational Acceleration, Brain Tissue Strain, and the Relationship to Concussion. *Journal of Biomechanical Engineering,* 137(3).

Post, A., Oeur, A., Hoshizaki, B. & Gilchrist, M. D., 2013. An examination of American football helmets using brain deformation metrics associated with concussion. *Materials and Design,* Volume 45, pp. 653-662.

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

Potvin, B. M. et al., 2019. A comparison of the magnitude and duration of linear and rotational head accelerations generated during hand-, elbow-, and shoulder-to-head checks delivered by hockey players. *Journal of Biomechanics,* Volume 91, pp. 43-50.

Prasad, P. & Mertz, H. J., 1985. The Position of the United States Delegation to the ISO Working Group 6 on the Use of HIC in the Automotive Environment. *SAE Technical Paper Series.*

Ramage-Morin, P. L., 2017. *Health Reports - Cycling in Canada,* Canada: Statistics Canada.

Roseveare, A. J., Plant, D. J. & Ghajari, M., 2016. *A new helmet-liner design for improved survivability.* Malaga, Spain, s.n.

Rossi, A. M., Claiborne, T. L., Thompson, G. B. & Todaro, S., 2016. The Influence of Friction Between Football Helmet and Jersey Materials on Force: A Consideration for Sport Safety. *Journal of Athletic Training,* 51(9), pp. 701-708.

Rowson, B., Rowson, S. & Duma, S., 2015. Hockey STAR: A Methodology for Assessing the Biomechanical Performance of Hockey Helmets. *Annals of Biomedical Engineering,* 10(2429-2443), p. 43.

Rowson, B., Terrel, E. J. & Rowson, S., 2018. Quantifying the effect of the facemask on helmet performance. *J Sports Engineering and Technology,* 232(2), pp. 94-101.

Rowson, S. et al., 2009. Linear and Angular Head Acceleration Measurements in Collegiate Football. *Journal of Biomechanical Engineering,* Volume 131.

Rowson, S., Daniel, R. W. & Duma, S. M., 2013. Biomechanical performance of leather and modern football helmets. *J Neurosurg,* Volume 119, p. 805–809.

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

Rowson, S. & Duma, S. M., 2013. Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration. *Annals of Biomedical Engineering,* 41(5), pp. 873-882.

Rowson, S. et al., 2012. Rotational Head Kinematics in Football Impacts: An Injury Risk Function for Concussion. *Annals of Biomedical Engineering,* 40(1), pp. 1-13.

Sanstreak Corp., 2019. *Edgertronic.* [Online]
Available at: https://edgertronic.com/home
[Accessed 05 05 2019].

Schuller, E. & Niemeyer, I., 2005. Mechanism and Tolerance Curves of Traumatic Diffuse Axonal Injury (DAI). In: *Gilchrist M.D. (eds) IUTAM Symposium on Impact Biomechanics: Fundamental Insights to Applications. Solid Mechanics and Its Applications, vol 124.* Dordrecht: Springer.

Sharp, D. J. & Jenkins, P. O., 2015. Concussion is confusing us all. *Pract Neurol,* Volume 15, pp. 172-186.

Siegkas, P., Sharp, D. J. & Ghajari, M., 2019. The traumatic brain injury mitigation effects of a new viscoelastic add-on liner. *Scientific Reports.*

Smith, T. A., Tees, D., Thom, D. R. & Hurt, Jr, H. H., 1994. Evaluation and Replication of Impact Damage to Bicycle Helmets. *Accident Analysis and Prevention,* 26(6), pp. 795-802.

Sone, J. Y., Kondziolka, D., Huang, J. H. & Uzma, S., 2016. Helmet efficacy against concussion and traumatic brain injury: a review. *Journal of Neurosurgery,* 126(3), pp. 1-14.

Sprecher, A. M., 2019. *Injury Data.* [Online]
Available at: https://www.playsmartplaysafe.com/newsroom/reports/injury-data/
[Accessed 30 August 2019].

Sproule, D. W., Campolettano, E. T. & Rowson, S., 2017. Football helmet impact standards in relation to on-field impacts. *Proc Inst Mech Eng P J Sport Eng Technol,* 231(4), pp. 317-323.

Svensson, M. Y. & Lovsund, P., 1992. *A Dummy for Rear-End Collisions - Development and Validation of a New Dummy-Neck.* Verona, Italy, IRCOBI Conference.

Takhounts, E. G. et al., 2011. *Kinematic Rotational Brain Injury Criterion (BRIC).* Washington, 22nd International Technical Conference on the Enhanced Safety of Vehicles (ESV).

Takhounts, E. G. et al., 2008. Investigation of Traumatic Brain Injuries using the Next Generation of Simulated Injury Monitor (SIMon) Finite Element Head Model. *Stapp Car Crash Journal,* p. 52.

Trotta, A. et al., 2018. Evaluarion of The Head-Helmet Sliding Properties in an Impact Test. *Journal of Biomechanics,* Volume 75, pp. 28-34.

Tyson, A. M. & Rowson, S., 2018. *Adult Football STAR Methodology,* Blacksburg: Virginia Tech Helmet Lab.

Ueno, K. & Melvin, J. W., 1995. Finite element model study of head impact based on Hybrid III head acceleration: the effects of rotational and translational acceleration. *J Biomech Eng,* 117(3), pp. 319-328.

UN ECE Regulation 22/05, 2002. *Uniform Provisions Concerning The Approval of Protective Helmets and Their Visors for Drivers and Passengers of Motor Cycles and Mopeds,* Geneva: United Nations.

Versace, J., 1971. A review of the severity index. *SAE Technical Paper 710881.*

Viano, D. C., 2005. Head Impact Biomechanics in Sport. In: *Mechanisms and Tolerance Curves of Traumatic Diffuse Axonal Injury (DAI). In: Gilchrist M.D. (eds) IUTAM Symposium on Impact Biomechanics: From Fundamental Insights to Applications. Solid Mechanics and Its Applications, vol 124.* Dordrecht: Springer.

Viano, D. C., Pellman, E. J., Withnall, C. & Shewchenko, N., 2006. Concussion in professional football: performance of newer helmets in reconstructed game impacts - part 13. *Neurosurgery,* 59(3), pp. 591-606.

VICIS Inc., 2019. *VICIS - Protect the Athlete, Elevate the Game.* [Online] Available at: https://vicis.com [Accessed 15 09 2019].

Walsh, E. S., Kendall, M., Hoshizaki, T. B. & Gilchrist, M. D., 2014. *Dynamic impact response and predicted brain tissue deformation comparison for an impacted hybrid III headform with and without a neckform and torso masses.* s.l., IRCOBI Conference.

Walsh, E. S., Rousseau, P. & Hoshizaki, T. B., 2011. The influence of impact location and angle on the dynamic impact response of a Hybrid III headform. *Sports Eng,* Volume 13, pp. 135-143.

Wennberg, R. A. & Tator, C. H., 2003. National Hockey League Reported Concussions, 1986-97 to 2001-02. *The Canadian Journal of Neurological Sciences,* Volume 30, pp. 206-209.

Whyte, T. et al., 2019. A Review of Impact Testing Methods for Headgear in Sports: Considerations for Improved Prevention of Head Injury Through Research and Standards. *Journal of Biomechanical Engineering,* 141(7).

Wilcox, B. J. et al., 2014. Head Impact Exposure in Male and Female Collegiate Ice Hockey Players. *J Biomech,* 47(1), pp. 109-114.

Williams, M., 1991. The Protective Performance of Bicyclists' Helmets in Accidents. *Accident Analysis and Prevention,* 23(2/3), pp. 119-131.

Willinger, R. & Baumgartner, D., 2003. Human head tolerance limits to specific injury mechanism. *IJCrash,* 8(6), pp. 605-617.

Willinger, R. et al., 2015. *Final report of Working Group 3: Impact Engineering,* Brussels, Belgium: COST Action TU1101/HOPE collaboration.

Withnall, C., Shewchenko, N., Gitttens, R. & Dvorak, J., 2005. Biomechanical investigation of head impacts in football. *Br J Sports Med,* 39(Suppl I), pp. i49-i57.

Yengo-Kahn, A. M., Johnson, D. J., Zuckerman, S. L. & Solomon, G. S., 2015. Concussions in the National Football League - A Current Concepts Review. *The American Journal of Sports Medicine,* 44(3), pp. 801-811.

Yu, W., Chen, C., Chiu, W. & Lin, M., 2011. Effectiveness of different types of motorcycle helmets and effects of their improper use on head injuries. *International Journal of Epidemiology,* Volume 40, pp. 794-803.

Zhang, L., Yang, K. H. & King, A. I., 2004. A proposed injury threshold for mild traumatic brain injury. *Journal of Biomedical Engineering,* Volume 126, pp. 226-236.

Appendix I.

Upcoming Publications

Abram, D., Wikarna, A., Wang, G. & Golnaraghi, F. A Novel Oblique Impact Test Rig.

Abram, D., Wikarna, A., Wang, G. & Golnaraghi, F. Predicting the Kinematic Response of a Helmet.

Wikarna, A., Abram, D., Wang, G. & Golnaraghi, F. A Kinematic Rating System for Football Helmet.

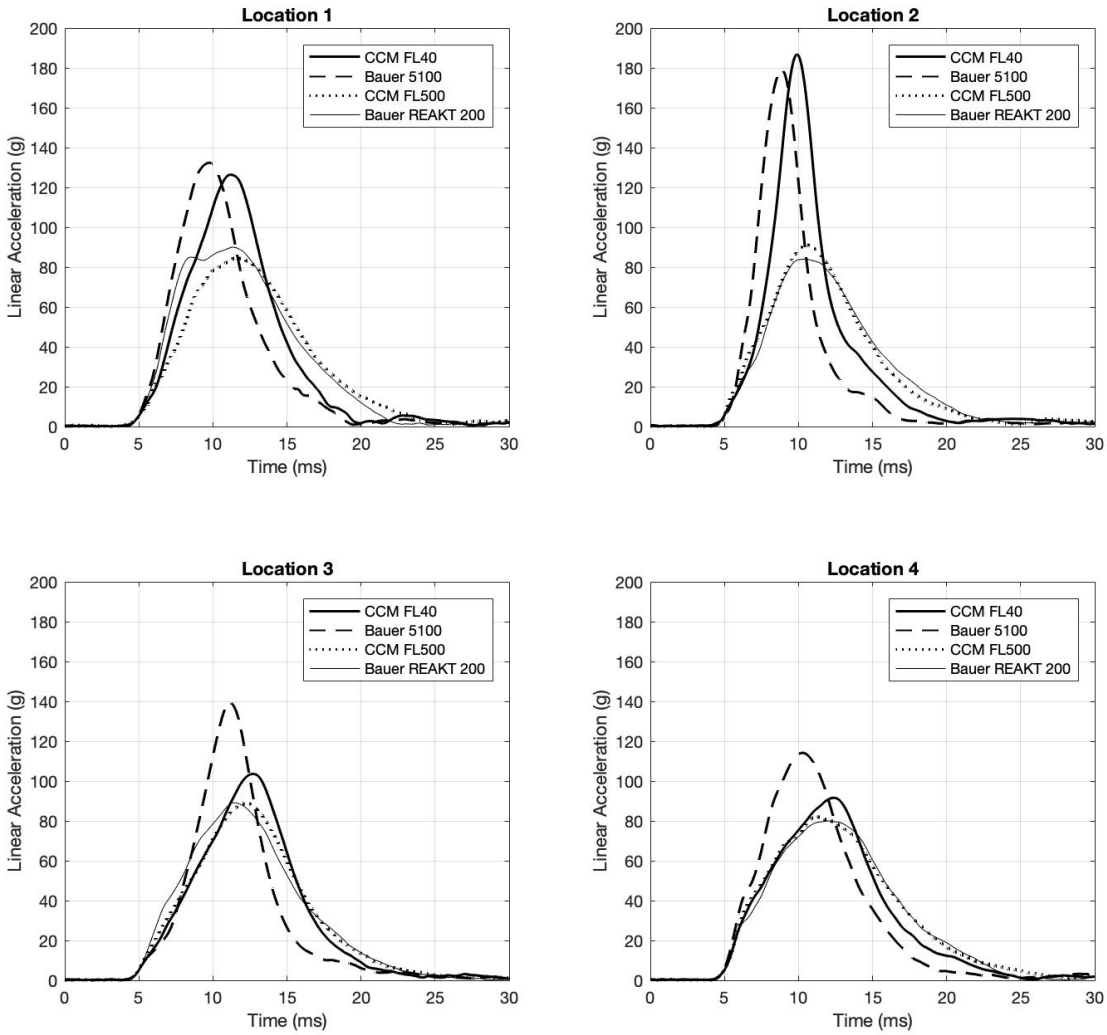
Wikarna, A., Abram, D., Wang, G. & Golnaraghi, F. A Kinematic Rating System for Hockey Helmet.

Wikarna, A., Abram, D., Wang, G. & Golnaraghi, F. A Kinematic Rating System for Cycling Helmet.

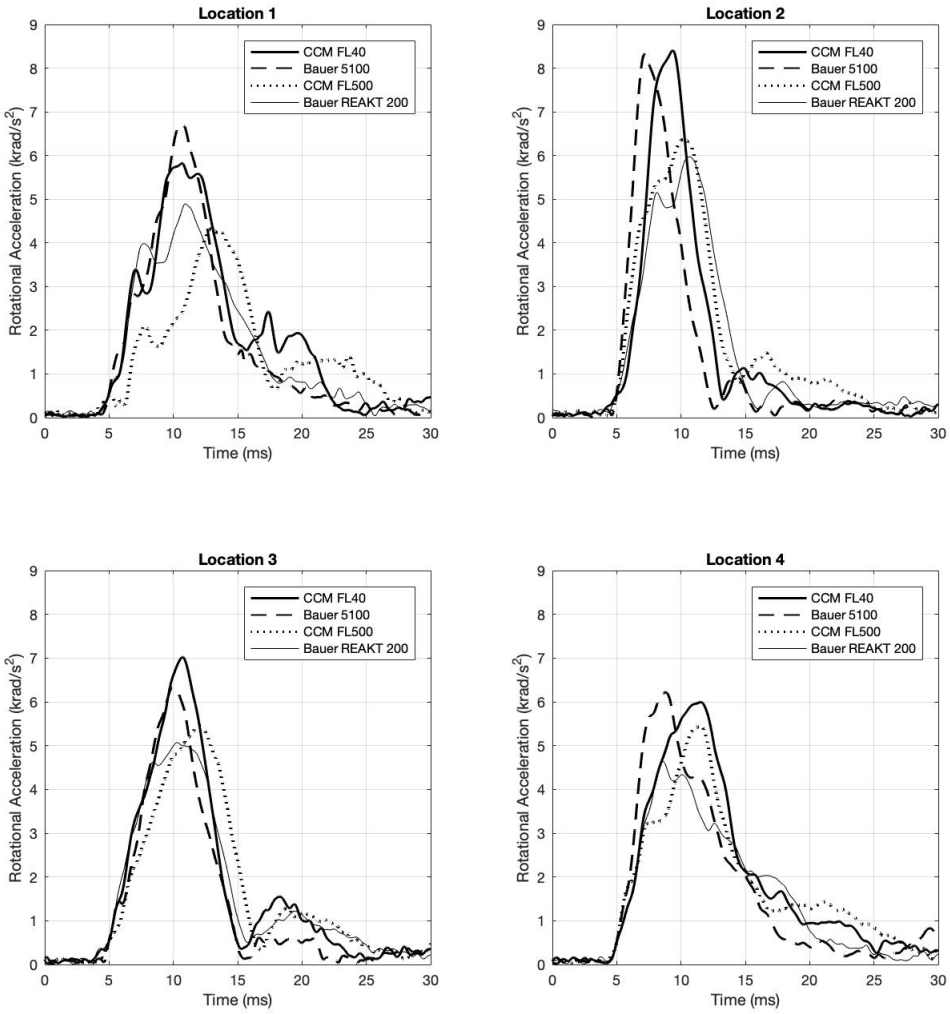
Appendix II.

Time-based Graphs for Results of Hockey and Cycling Helmets

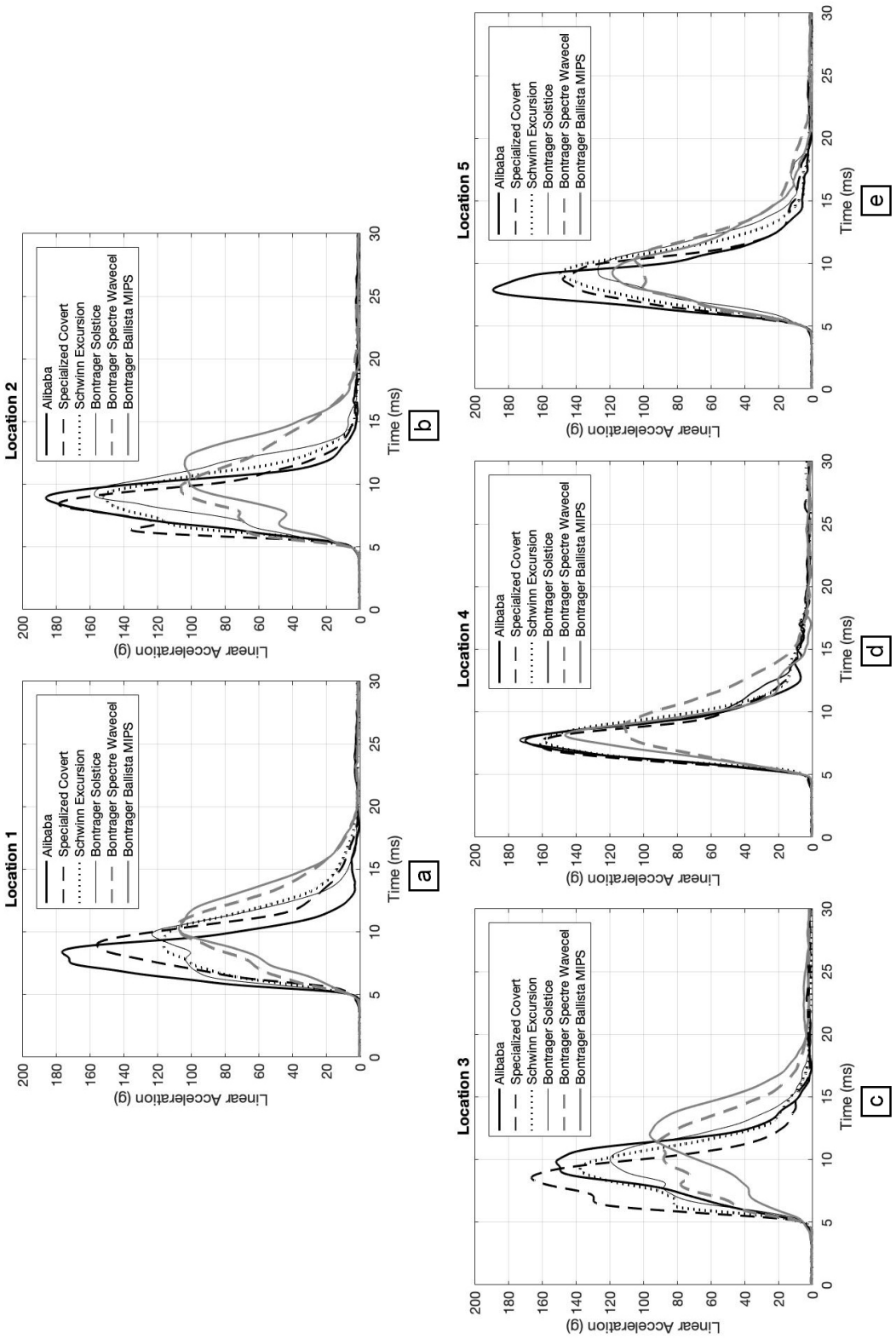
Hockey Helmets Average Linear Acceleration Results



Hockey Helmets Average Rotational Acceleration Results



Cycling Helmets Average Linear Acceleration Results



Cycling Helmets Average Rotational Acceleration Results