Biomechanical Assessment of Cycling Helmets: the Influence of Headform and Impact Velocity based on Bicycle Collisions associated with Injury Claims

by

Meagan J. Warnica

A thesis
presented to the University of Waterloo
in fulfillment of the
thesis requirement for the degree of

Master of Science

in

Kinesiology

Waterloo, Ontario, Canada, 2015

Author’s Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners. I understand that my thesis may be made electronically available to the public.

Abstract

The goal of my thesis was to fill some of the gaps in knowledge about cyclist/motor vehicle collisions and testing guidelines for cycling helmets. Cycling collisions with motor vehicles represent a problem for the Canadian health care system as they can cause severe injuries, especially to the head. Our current knowledge of the factors involved in cycling collisions in Southern Ontario is limited due to current injury reporting techniques. Furthermore, the effectiveness of cycling helmets for mitigating injury during high energy impacts is unknown as testing guidelines are designed for lower energy impacts, such as a sideways fall from a bicycle. Accordingly, my thesis was designed with two studies to address these limitations. The first study was novel as it was the first to characterize cycling collisions in Southern Ontario that resulted in injury claims and determine if relationships existed between injury circumstances (e.g. helmet use, impact surface) and injury outcomes. Data was collected from a unique database at a professional forensic engineering company. Using a subset of this data, a head impact velocity was determined to represent higher energy impacts of cyclist/motor vehicle collisions. The second study compared peak dynamic headform responses between three headforms (two biofidelic and the magnesium headform currently used in testing standards) and also assessed the mitigating capacity of three brands of cycling helmets when subjected to impact velocities of standard testing scenarios as well as higher energy impact velocities (determined in study one). It determined that the Hybrid III headform may be an appropriate tool for helmet testing. Furthermore, the helmets tested mitigated injury below injury thresholds at impact velocities used in current testing standards but *not* at an impact velocity representative of a higher energy scenario such as a cyclist/motor vehicle collision, as determined in study one. Injury risk reduction was affected by helmet brand with more expensive helmets not necessarily producing better results. These findings indicate the need for more work in the area of improving and understanding the biofidelity of our testing regimes. Finally, helmet manufacturers should be urged to be more transparent to consumers about the relative mitigating capacity of their helmet brands perhaps by creating a rating system for helmet safety.

Acknowledgements

I have been extremely fortunate to have the support and resources available to me throughout my time in graduate school. The Kinesiology graduate program is a thriving academic workplace with a challenging atmosphere yet supportive staff and students -- not to mention their very own MacGyver (Jeff Rice).

I would like to gratefully acknowledge my funding sources that I was so fortunate to receive throughout my program, the University of Waterloo, the Ontario Graduate Scholarship program, the Natural Sciences and Engineering Research Council Alexander Graham Bell Canada Graduate Scholarship, and the Toronto Rehabilitation Institute Scholarship for Students with Disabilities.

To my supervisor, Dr. Andrew Laing, you were the first to recognize my abilities and although you were likely unaware, you helped me through some of the most difficult decisions I have made in the past five years. You recognized my interests and never stopped trying to find a way to help me explore them making my graduate experience that much more rewarding. I was also extremely fortunate to meet and collaborate with Dr. Robert Parkinson and Giffin Koerth Smart Forensics who welcomed me into their workplace and made my thesis possible.

Last but not least, I would like to acknowledge my family, Carolyn, David, Josie, and William, who have been the closest to me throughout this journey and were always there with love, support, and the best barbequed meals I have ever had.

Table of Contents

List of Figures

List of Tables

Chapter 1: Thesis Overview

Motor vehicle collisions with cyclists represent the second largest number of non-fatal transport related injuries in Canada (SMARTRISK, 2009). Cycling as a sport, recreational activity and mode of transportation has increased in popularity over the past ten years (Pucher, Buehler, & Edward, 2011). Head injuries are among the most frequent injuries for cyclists with a reported incidence between 25-39% in epidemiological studies (Amoros, Chiron, Thélot, & Laumon, 2011; Haileyesus, Annest, & Dellinger, 2007; Rivara, Thompson, & Thompson, 1997). Helmets are designed and marketed to cyclists for head injury mitigation following a collision but compliance is generally a choice as the use of helmets is not mandatory for adults in Ontario (18+ years) (Government of Ontario, 2012). Head injuries represent a significant problem for the Canadian health care system as well as for the general quality of life of individuals. Therefore, it is important to understand the mechanisms of injury and consider the role of helmets and the injury mitigation they offer in cycling collisions.

Two general types of injury reporting exist in Canada. Information from police collision reports consist of general information about cycling/motor vehicle collisions such as characteristics of the cyclist and sometimes orientations of the vehicles upon impact. However, such incidents can be underreported and those that are reported have been shown to be inaccurate at times (Rosman & Knuiman, 1994). Hospital injury databases, such as the National Trauma Registry, contribute to a second method of collision injury reporting. These reports usually contain similar generalized collision information as well as specific information about injuries (CIHI, 2011). While these two types of injury reports can provide some useful information about characteristics of cycling collisions, there are still gaps in knowledge about more specific aspects of the collisions (such as cyclist and vehicle speed). Forensic engineering firms are sometimes hired to reconstruct collisions based on information from the police and hospitals as well as reported information from witnesses and involved parties. Databases from these firms provide a unique opportunity to reconstruct the cycling incidents and extract more specific information such as vehicle and cyclist speeds and orientations upon impact. Analyzing this data with respect to injury outcomes may provide more useful insights into collision characteristics that contribute to head injuries following cyclist/motor vehicle collisions.

Helmets are designed to mitigate head injuries to cyclists following impacts to the head. Therefore, it is important to consider mechanisms of these injuries in the design and testing of helmets. Although specific mechanisms of brain injury may not be addressed by helmets, evidence generally supports the use of helmets in decreasing the overall rate of head injury (Benson, Hamilton, Meeuwisse, McCrory, & Dvorak, 2009; Thompson, Nunn, Thompson, & Rivara, 1996). Helmet testing standards were introduced for cycling helmets in 1970 and it is now mandatory for helmets to pass these standards to be sold on the common market (American Society for Testing and Materials, 2004; Canadian Standards Association, 2009; Consumer Product Safety Commission, 1998; SNELL, 1995). Standards use a surrogate headform to test helmets at various impact energies. The headforms currently used in the standards are made of metal which improves repeatability of the impacts but decreases the biofidelity of the headforms. The National Operating Committee on Standards for Athletic Equipment (NOCSAE) has designed a biofidelic headform for their testing of athletic helmets (National Operating Committee on Standards for Athletic Equipment, 2009). The Hybrid III headform is another biofidelic headform currently used as part of a crash test dummy by the National Highway Traffic Safety Administration (NHTSA) in vehicle safety testing. Both these headforms may improve dynamic responses following impacts due to their biofidelic designs. A further limitation of current helmet testing standards is the ability of the impact energies used during testing to represent realistic common cycling collisions. Impact testing energies were chosen to represent a basic fall from a bicycle to the ground without any third party involvement (Walker, 2005). Therefore, drawing conclusions about the mitigating capacity of helmets during higher energy impacts such as collisions with motor vehicles cannot be done.

My thesis consisted of two main objectives to address the limitations of the current epidemiological literature and its link to helmet testing standards. The first was to characterize typical impact scenarios for cyclist collisions in Southern Ontario that resulted in injury claims and determine a head impact velocity more representative of cyclist collisions involving a motor vehicle. A database from a professional forensic engineering company, Giffin Koerth Smart Forensics, was searched and analyzed for a list of variables describing the circumstances of the collisions, cyclist characteristics, and injury outcomes. The specific goals of this first study were:

1) to describe the crash characteristics of high severity cycling collisions in Southern Ontario resulting in injury claims
2) to determine whether relationships exist between injury circumstances and resulting injury outcomes

Current helmet testing standards consist of several limitations including the biofidelity of the headforms used and the realism of test impact energies. Therefore, the second objective of my thesis was to compare peak dynamic headform responses between three surrogate headforms and to evaluate the mitigating capacity of cycling helmets during impacts at velocities currently used in testing standards as well as one more representative of collisions with motor vehicles.

The peak dynamic responses for three surrogate headforms were compared (two biofidelic headforms (NOCSAE and Hybrid III) and the metal headform used in testing standards (Canadian

Standards Association, 2009)) for three impact locations (front, back, and side impacts), with and without helmets. The study also evaluated the mitigating capacity of three cycling helmets during the impact velocity consistent with the CSA standards compared to an average impact velocity of high energy collisions. Mitigating capacity of the helmets was determined by comparing outcome variables to injury assessment reference values (IARV).

There were two main hypotheses for the second study:

1) The biofidelic headforms (NH and HIII) will have similar responses to impacts and will have different responses than the magnesium K1A headform for all impact orientations and for both unhelmeted and helmeted trials.

2) The helmets will adequately mitigate injuries (below an injury threshold) for the impacts completed at the standard velocity but not for impacts at the 'MV velocity' for all impact orientations.

Chapter 2: Literature Review

2.1 Cyclist Injuries from Collisions with Motor Vehicles

2.1.1 Scope of the problem

Cycling is a common recreational activity, sport and mode of transportation in which helmets are typically worn to mitigate head injuries in the event of an impact to the head. In Canada, 1.3% of commuters use bicycles as their mode of transportation to work and this number has risen in the past 19 years with more drastic changes seen in larger cities such as Toronto and Vancouver (Pucher et al., 2011). Bicycles are considered vehicles under the Highway Traffic Act and cycling collisions ranked second to motor vehicle incidents for non-fatal transport related injuries (SMARTRISK, 2009). According to Transport Canada (2012), about 2.7% of all traffic fatalities and 3.9% of serious injuries in traffic collisions in Canada involve bicyclists. Since these statistics are only from highway traffic act reportable scenarios, hospitalization records may give a more accurate account of injuries and costs to the health care system.

Cycling incidents accounted for 15% of hospitalizations, 21% of emergency room visits, 17% of permanent partial disability and 16% of permanent total disability from transport related collisions in 2004. This corresponded to $443 million in total costs to the Canadian health care system (SMARTRISK, 2009). In 2009-2010, Ontario saw the largest number of cycling incidents in Canada with ~1300 cycling injuries causing hospitalization (CIHI, 2011). More detailed information is hard to obtain from injury databases as they typically only report broad characteristics of impact scenarios. Risks to cyclists are greater in large urban areas where there are more commuters on bicycles and in motor vehicles (Amoros et al., 2011; Pucher et al., 2011). A report from the City of Toronto Police Services found 1315 total bicycle collisions in the City of Toronto in 2011, 87.6% of which caused injury or were fatal (City of Toronto, 2011).

Pucher et al. (2011) has suggested that cycling has become “safer” in Canada with trends of injuries decreasing slightly over the past 20 years. This could be due to multiple factors such as the increase in cycling awareness, increase in number of cycling paths and bike lanes or the introduction of mandatory helmet laws (Government of Ontario, 2012; Pucher et al., 2011). However, incidence of hospitalization due to cycling injury and cycling head injury has stayed relatively the same over the past ten years in Canada (CIHI, 2011) (**Error! Reference source not found.**) suggesting that it is still a major problem for our health care system.

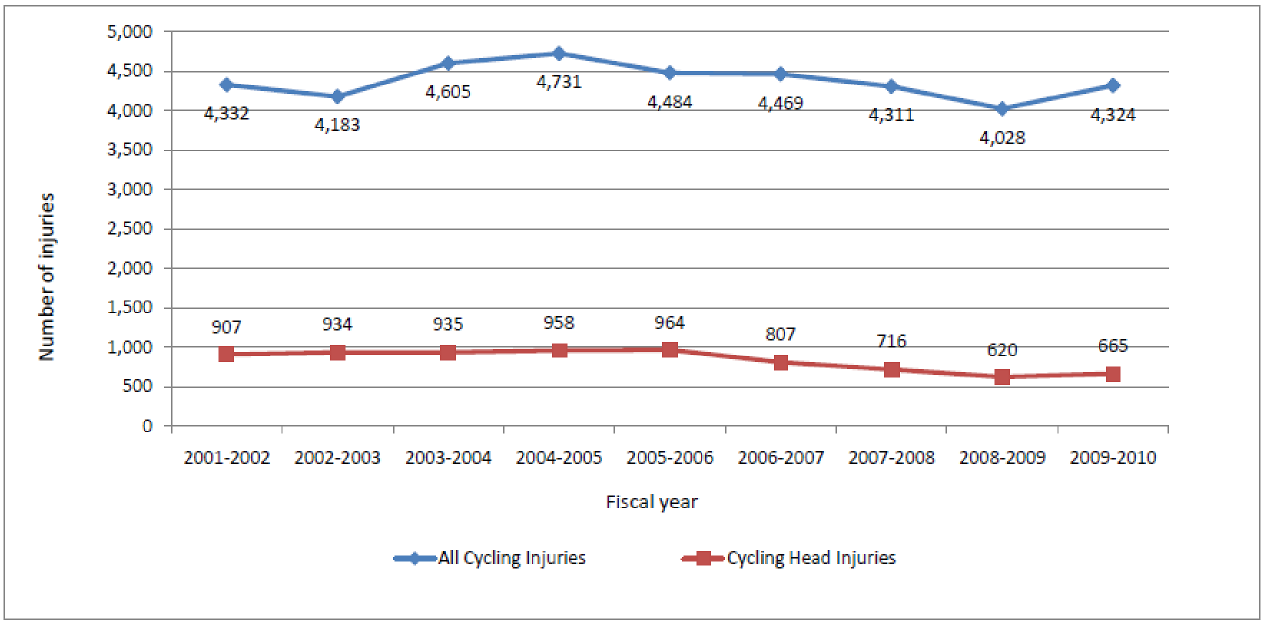


Figure 2-1: Number of Cycling Injuries and Cycling Head Injuries by Fiscal Year in Canada (CIHI, 2011)

2.1.2 Risk Factors for Cycling Collisions

A review Thompson & Rivara (2001) indicated some of the major risk factors for injuries related to cycling (Table 2-1). Many cyclists tend to be males as opposed to females which could explain the increased incidence of injuries in male cyclists (Pucher et al., 2011). This risk factor could also be due to the tendency of males to engage in more aggressive riding. Competitive as well as recreational mountain biking increases a cyclists risk of injury (Chow, Bracker, & Patrick, 1993). Injury outcomes are typically different among mountain bikers when compared to recreational cyclists with more injuries seen in the head, neck and upper limbs as opposed to the head and lower limbs in recreational cycling (Carmont, 2008; Chow et al., 1993). Due to these differences, specialized helmet design and protective equipment are recommended for mountain biking compared to recreational or commuter cycling (Chow, Corbett, & Farstad, 1995).

Cycling in the afternoon or early evening when the lighting conditions are typically changing causes an increase in risk of injury (City of Toronto, 2011) . Finally, riding while intoxicated causes an increased risk of injury according to Rivara et al. (1997). Although, this risk was shown to be insignificant and could also be due to poor rider choices such as a significant decrease in helmet compliance among intoxicated cyclists (Rivara et al., 1997; Thompson & Rivara, 2001).

Table 2-1: Risk Factors for Bicycle-Related Injuries (adapted from Thompson & Rivara, 2001)

Risk Factor
Cyclist is male.
Cyclist is nine to 14 years of age.
Cycling in the summer.
Cycling in late afternoon or early evening.
Cyclist does not wear helmet.
Motor vehicle involved.
Unsafe riding environment.
Cyclist is from an unstable family environment.
Cyclist has preexisting psychiatric condition.
Cyclist is intoxicated.
Cyclist is involved in competitive mountain-bike racing

Among the most detrimental risk factors to cyclists, and those further investigated in this thesis, are collisions with a motor vehicle and helmet use (Amoros et al., 2011; Haileyesus et al., 2007; Kim, Kim, Ulfarsson, & Porrello, 2007; Lustenberger et al., 2010; Rivara et al., 1997). The Canadian Hospital Injury Reporting and Prevention Program (2008) found that 6.5% of injuries were due to a motor vehicle collision. Although collisions with motor vehicles are not the top cause of cycling collisions, they tend to cause injuries that are much more severe (Haileyesus et al., 2007; Kim et al., 2007; Lustenberger et al., 2010; Rivara et al., 1997) with a 3.6 times increase in the severity of injury (as measured on an injury severity scale) when colliding with a motor vehicle (Rivara et al., 1997). Improving city infrastructure and increasing the number of bicycle lanes and paths has been proposed to address this problem (Pucher et al., 2011). However, designing infrastructure to completely eliminate this problem cannot be done with the number of motor vehicle users in urban areas. Thus, focusing on improving cyclist visibility and protection is also important to overall cycling safety.

2.1.3 Methods of Injury Reporting in Canada

Retrieving information about specific injuries in Canada can be a complicated task as there are many different bodies that report such information. Every two years, Statistics Canada sends out a Canadian Community Health Survey (CCHS) which is an overview of injuries to Canadians aged 12 and over (Stats Canada, 2013). The sample size for this survey is 65 000 people and is used to represent the typical injuries seen in all Canadians. The information obtained from this survey from 2009 was summarized and reported in an article "Injuries in Canada: Insights from the Canadian Community Health Survey" by Billette & Janz (2011). Injuries to cyclists and pedestrians from collisions with motor vehicles are excluded from this report making it impossible to understand the epidemiology of cyclist/motor vehicle collisions.

Two types of injury reports exist in Canada for cyclist/motor vehicle collisions. Cycling collisions with motor vehicles fall under the highway traffic act (HTA) (Government of Ontario, 2012). Therefore, police reports can be summarized to describe the types of vehicles involved and outcomes of the collisions. Transport Canada produces an annual report (Canadian Motor Vehicle Traffic Collision Statistics) from the Transport Canada National Collision Database (NCDB) which contains data on all reported motor vehicle collisions in Canada (Transport Canada, 2012). This report includes the incidence of cyclist collisions with motor vehicles and comments on fatalities and serious injuries to the cyclists. However, it does not provide specific information relating to the characteristics of the collision. More detailed information about collisions may be provided by individual police departments. However, it is not mandatory for these reports to be public. The Toronto Police Services have provided a public report characterizing age, gender, time of year and day, and helmet use for cyclist/motor vehicle collisions in 2011 (City of Toronto, 2011). Although this report is more detailed than those from Transport Canada, specific characteristics are not included such as injuries to the cyclist, speed of the vehicles, collision orientations, etc. Furthermore, some evidence has suggested that police reported data may be inaccurate and the number of collisions and injuries may be underreported (Rosman & Knuiman, 1994).

The second method of injury reporting in Canada is by summarizing databases of injury statistics generated by Canadian hospitals. The National Trauma Registry (NTR) includes demographic, diagnostic and procedural information from all admissions to acute care hospitals across Canada (CIHI, 2013). The Canadian Institute for Health Information uses this registry to produce reports on the injuries such as “Cycling Injury Hospitalizations in Canada, 2009-2010” (CIHI, 2011). This specific report includes categories of incidents as well as number of injuries and head injuries for each province/territory. The Canadian Hospitals Injury Reporting and Prevention Program (CHIRPP) is another surveillance program, run by the Public Health Agency of Canada, that reports on injuries from 14 hospitals in Canada (10 pediatric and 4 general). CHIRPP has previously released a report on injuries associated with bicycles with more specific injury information for cyclists (CHIRPP, 2008). The Economic Burden of Unintentional Injury in Canada (2009) uses hospital data from CIHI as well as information from the National Ambulatory

Care Reporting System (NACRS) to predict rates of injury in Canada and assess the economic burden to the health care system (SMARTRISK, 2009). However, like the police reports, more detailed information on cyclist/motor vehicle collisions is not available.

Epidemiological studies have been done of various populations in relation to cycling/motor vehicle collisions (Amoros et al., 2011; Depreitere et al., 2004; Eilert-Petersson & Schelp, 1997; Haileyesus et al., 2007; Kim et al., 2007; Linn, Smith, & Sheps, 1998; Lustenberger et al., 2010; Rivara et al., 1997). These studies are all based on samples from road trauma registries, police reported data, or hospital produced data. Epidemiological studies for specific injury and injury causation data are not common in Canada and have not been completed for the province of Ontario. Furthermore, although studies from other regions included more specific information about cyclist age, height, gender, helmet use, predicted head orientation etc., there was not one study that analyzed each incident in detail to produce impact speeds and orientations of the vehicles as well as injury outcomes. Analyses of detailed information on cycling collisions have mainly been done by simulating collisions using crash test dummies and cadavers (Simms & Wood, 2009). Analyzing regional bicycle/motor vehicle collisions would be beneficial for designing regional road infrastructure or promoting cycling health and safety specific to the population. Furthermore, more detailed knowledge of these incidents will add to our knowledge and understanding of testing scenarios for cycling safety.

2.2 Impact Mechanics and Head Injuries

2.2.1 Occurrence of Injuries

Injuries to cyclists following cyclist/motor vehicle collisions are generally among the most severe injuries due to the high impact speeds of the events (Kim et al., 2007; Rivara et al., 1997). In a study reviewing bicycle collisions in the United States, collisions with motor vehicles caused over 90% of cyclist fatalities (Lustenberger et al., 2010). The most common, non-fatal cyclist injuries following collisions with a motor vehicle are injuries to the lower extremities and head (Amoros et al., 2011; Eilert-Petersson & Schelp, 1997; Haileyesus et al., 2007; Rivara et al., 1997; Simms & Wood, 2009). The head was the primary body part injured in 38.6% of the hospitalized patients in Haileyesus, et al. (2007) and in 25.1% of the teens and adults injured in Amoros, et al. (2011). Furthermore, Amoros et al. (2011) found that two thirds of the most severe injuries as classified by the Abbreviated Injury Scale (AIS) were to the head and Thompson & Rivara (2001) found that injuries to the head caused the highest number of fatalities and long-term disabilities.

In order to better understand the problem of cyclist head injuries from collisions with motor vehicles, it is important to consider all aspects of collision circumstances involving the head. These circumstances include impact orientations of the head with respect to the surface in which it hits, impact velocity, helmet type and use, and the surface against which the head is hitting. Reported data on these important variables is not extensive. Often, injury databases and research studies use estimated information from drivers and cyclists which can have a high degree of error (Rosman & Knuiman, 1994). Therefore, kinematics of cyclist injuries following motor vehicle collisions have often been studied using crash test dummies or cadavers (Simms & Wood, 2009). While these can be useful to predict some aspects of the incident, estimations of impact speeds, orientations and trajectories of the bodies are often made in the design of the simulations which may take away from the realism of the impacts. The following sections summarize the main factors needed for modeling head injuries from cyclist/motor vehicle collisions.

2.2.1.1 Impact Surfaces

In collisions with motor vehicle collisions there are many obstacles that pose a risk for impacting a cyclists head. During collisions, one study found that the ground was the first impact site for 31% of the injuries (Simms & Wood, 2009). Of the cyclists that impacted the motor vehicle, the windshield was the most common impact site for cyclists accounting for 14% of the injuries (Simms & Wood, 2009). There is limited data available summarizing impact sites of cyclists on motor vehicles and their links to specific injuries as larger injury databases do not include impact sites on vehicles as an outcome and studies specifically investigating vehicle impact sites typically have small sample sizes.

2.2.1.2 Impact Orientations

Cyclists may impact vehicles and/or the ground in multiple locations and orientations. The projection of the cyclist depends on the mass of the cyclist and motor vehicle, the height of the cyclist and bicycle, the height and size of the motor vehicle and the speeds and positions in which they hit. The mass of the bicycle is generally quite low and does not play as large of a role in the impact orientations of cyclist/motor vehicle collisions (Simms & Wood, 2009). Therefore, the most important factor influencing cyclist motion following the impact is the initial position of the cyclist with respect to the motor vehicle (Simms & Wood, 2009).

Simms & Wood (2009) analyzed collision data from several countries and found that cyclists are most frequently hit from the side by the front of the vehicle and are usually hit obliquely because of the elongated shape of the bicycles. A study of Japanese collision data divided the side impacts into three categories (Figure 2-2). Head injury occurred 38% of the time in category one and 69% of the time in category two (Simms & Wood, 2009). Oblique impacts can cause a “wrapping” phenomenon of the cyclist over the hood of the car depending on the cyclist’s centre of gravity and the speed of the vehicle (Simms & Wood, 2009). Taller cyclists therefore would be more likely to hit their head on the windshield and A-pillar, which are made of stiffer materials than the bonnet of the vehicle.

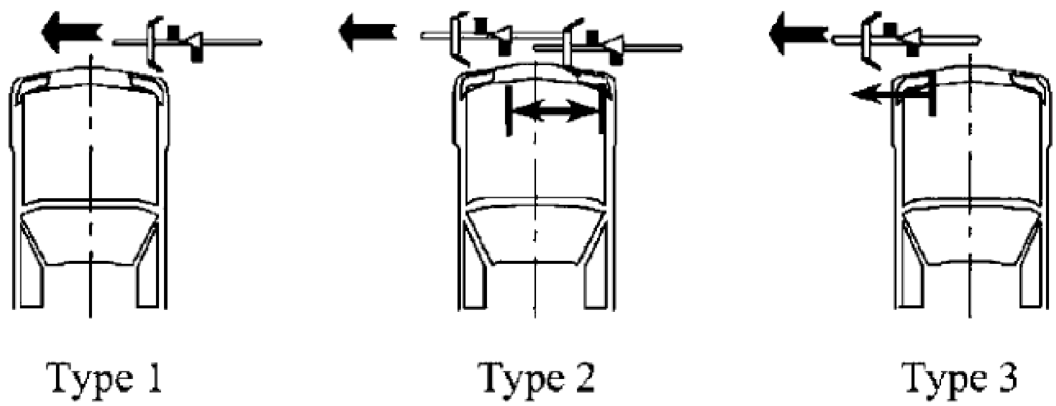


Figure 2-2: Three categories of side struck cyclists (adapted from Simms & Wood, 2009)

Studies involving crash test dummies and cadavers can be useful in determining initial impact orientations of the cyclists with respect to the motor vehicles and can help estimate body projections following impact. However, it is difficult to predict exact head impact locations based on these models because they may not be representative of realistic scenarios. Depreitere, et al. (2004) tried to deduce head impact location of 86 cycling incidents based on the location of swelling on computed tomography (CT) scans. Of the 33 identifiable head impact locations from collisions with motor vehicles, 23 had a single impact location, 8 had a double, and 2 had a triple impact (Depreitere et al., 2004). Most patients impacted their forehead or side of the head. However, in the motor vehicle group, a larger number of cyclists seemed to impact the back of the head as well (Figure 2-3) (Depreitere et al., 2004). These results suggest that cyclists most often hit these three locations following severe crashes.

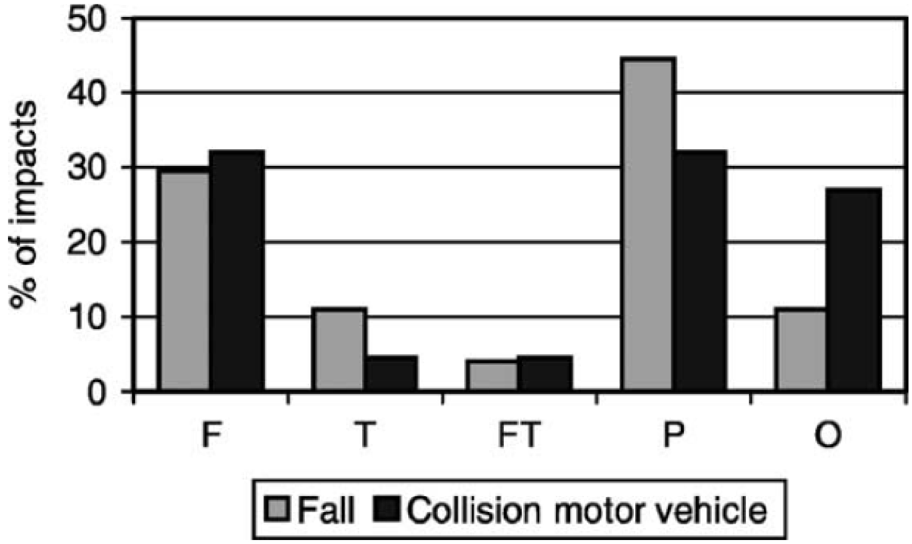


Figure 2-3: Impact sites for cyclists that sustained a single head impact from falls or collisions with a motor vehicle (N=49) (F - Frontal, T - Temporal, FT - Frontotemporal, P - Parietal, O - Occipital) (adapted from Depreitere, et al. (2004)

2.2.1.3 Impact Velocities

Estimating impact velocities for cycling collisions is difficult due to the reporting techniques of the incidents. German collision reports estimated that 50% of cars struck cyclists below 35 km/h and 76% struck them below 50 km/h (Simms & Wood, 2009). Although impact speeds of the vehicles can be estimated post-collision, estimating impacts of specific body parts of the cyclist (e.g. the head) is much more complicated depending on the direction of the cyclists fall post-impact. To achieve this, researchers have tried to model cyclist-motor vehicle impacts using software such as PC Crash (Datentechnik Group, Linz, Austria) and MADYMO (Mathematical Dynamical Model) (TASS International, Netherlands) or custom finite element programs. The models are generally validated with experimental studies using post-mortem human subjects (PMHS) (Fanta, Boucek, Hadraba, & Jelen, 2013). In impact configurations not involving a motor vehicle, Bourdet, Deck, Carreira, & Willinger (2012) found an average head impact velocity of 5.5 m/s which is a similar value to those used in helmet testing standards (American Society for Testing and Materials, 2012). In a recent model of cycling collisions with motor vehicles using MADYMO, Fanta et al. (2013) found head impact velocities from 4.6-13.5 m/s depending on the vehicle and bicycle size and type. This is likely due to the role of hood height in the rotation of the body following impact influencing the acceleration of the head (Fanta et al., 2013; Maki & Kajzer, 2001). The values resulting from this study are generally higher than velocities used in current testing standards (which range from ~5.4-6.6 m/s).

An advantage of reconstructing collisions using modeling software is the ability to approximate the velocity of the head just prior to impact. This information is not attainable from reported collision data from the police or hospitals. It is also less expensive than reconstructing collisions using PMHS data. The use of MADYMO and other finite element modeling programs to simulate cycling/motor vehicle collisions is a relatively new concept. Therefore, although head impact velocities can be calculated with these programs, validation with experimental data is still required to decrease the error with these approximated values.

2.2.2 Mechanisms of Head Injury in Cyclists

Many types of head injuries are seen following cyclist crashes due to the large variation in injury scenarios putting the cyclist at risk. Depreitere, et al. (2004) found no specific pattern in the type of head injury but did find that some injuries occur more frequently. Brain swelling, skull fractures, and cerebral contusions occurred in over 70% of head injured cyclists, of which the majority were not wearing helmets (Depreitere et al., 2004).

Mechanisms of cycling head injuries are difficult to analyze because injuries are often combined (Depreitere et al., 2004). Ommaya & Hirsch (1971) suggested that rotational acceleration could account for 50% of the potential for brain injury but that the remainder was attributed to direct loading or translational acceleration. Cyclist head injuries from collisions with motor vehicles are all dynamic in nature and typically occur at high speeds. Due to the lack of protection and restraint of the cyclist, all injuries result in impact, with the ground and/or vehicle, although, many are combined with other injuries that may be caused by inertial forces (Depreitere et al., 2004; Schmitt, Niederer, Cronin, Muser, & Walz, 2014; Simms & Wood, 2009).

Among the injuries seen in the head injured cyclists studied by Depreitere et al. (2004) were skull fractures, cerebral contusions, diffuse axonal injuries, extradural haematomas, subdural haematomas and intraventricular haemorrhaging. Figure 2-4 demonstrates typical injury mechanisms for head injuries. Skull fractures and some contusions are caused by direct contact loading to the skull from an external object (Figure 2-4). Extradural and subdural haematomas are usually secondary to these injuries (Depreitere et al., 2004; Schmitt et al., 2014). There has been some evidence indicating a slight tendency for the severity of skull fracture to increase as the severity of brain injury increases, however, the tendency was not found to be significant (Nahum & Melvin, 2002). Direct contact and deformation of the skull at one area can also cause stress waves to propagate in the brain leading to a pressure gradient. Positive pressure is associated with the site of impact (coup) and negative pressure is associated with the opposite side of the impact (contre-coup) (Schmitt et al., 2014; Simms & Wood, 2009). Although it is still unclear as to which gradient causes the most internal damage, pressure gradients can either cause tensile loading or shear strains within the brain tissue causing contusions and subdural hematomas (Schmitt et al., 2014). Finite element models and live animal testing have predicted a pressure threshold for minor brain injuries at 173 kPa (Simms & Wood, 2009).

Diffuse axonal injuries and contusions of the frontal and temporal lobes can be caused by inertial loading due to the motion of the brain within the skull (Figure 2-4) (Depreitere et al., 2004; Schmitt et al., 2014; Simms & Wood, 2009). Translational acceleration produces similar focal injuries as direct contact loading and rotational acceleration can produce diffuse axonal injuries (Figure 2-4)(Simms & Wood, 2009).

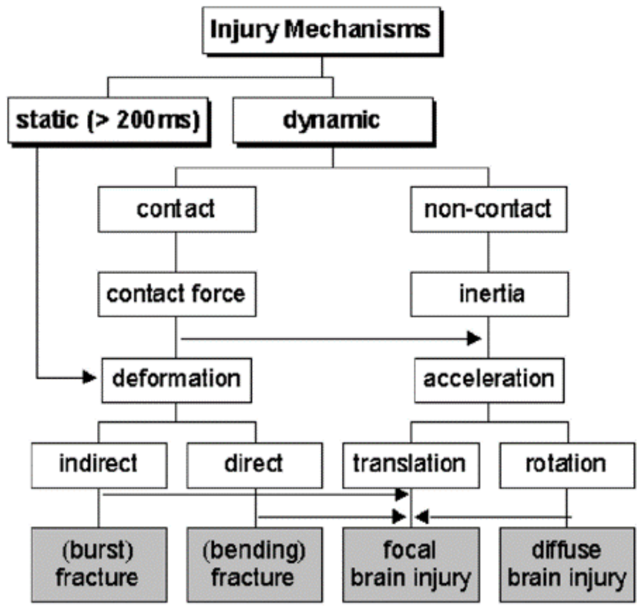


Figure 2-4: Possible mechanisms for head injury (Schmitt, et al., 2014)

Tolerance levels of the tissues involved are important to understand when designing prevention strategies for these injuries (e.g. protective devices) and testing guidelines for protective devices. Since surrogate headforms in testing guidelines currently are not biofidelic, there is no way to measure variables at the level of the brain. Therefore, although it is useful to understand the tolerances of specific brain tissue, we cannot currently predict injuries beneath the skull. Fracture force of the skull has been measured (Table 2-2) and it differs across areas of the head (Schmitt et al., 2014). These results would be more relevant to our current level of impact testing with surrogate headforms.

Table 2-2: Peak force for fracture at different areas of the skull (Schmitt, et al., 2014)

Impact Area	Force (kN)
Frontal	4.0-6.2
Lateral	2.0-5.2
Occipital	12.5

2.2.3 Head Injury Prediction

With our current level of understanding of head injuries, specific injury tolerances cannot be used in many situations such as helmet testing. Therefore, head injury prediction tools have been designed to help bridge the link between measured variables such as forces and accelerations and injury outcomes. Due to the large variation in types of head injuries and lack of knowledge of exact mechanisms that cause them, there is no one specific parameter that can predict all types of head injury and lesion-specific criteria do not exist (Depreitere et al., 2004; Simms & Wood, 2009). The following sections describe the most common head injury prediction tools.

2.2.3.1 Wayne State Tolerance Curve (WSTC)

The Wayne State Tolerance Curve (WSTC) was originally developed based on cadaver tests focusing on head acceleration (Schmitt et al., 2014). It indicates a relationship between the linear acceleration required for skull fracture and the impact duration (Schmitt et al., 2014; Simms & Wood, 2009). Since skull fracture was correlated with moderate concussion, the WSTC was proposed as an appropriate criterion for the prediction of head injury and became the first model to assess the tolerance of the human head to impact (Nahum & Melvin, 2002).

The WSTC was initially developed only using six data points. This has contributed to some criticism along with its questionable instrumentation techniques, lack of documentation on the data used to extend the curve, and the uncertainty of the definition of acceleration levels (Nahum & Melvin, 2002; Schmitt et al., 2014). The WSTC assumes that translational acceleration causes pressure gradients in the brainstem which results in shear-strain injuries making the curve incapable of accounting for rotational accelerations (Nahum & Melvin, 2002).

2.2.3.2 Gadd Severity Index (SI)

The Gadd Severity Index (SI) (Equation 2.1) was developed by Gadd et al. (1966) based on the finding that the WSTC becomes a straight line in a logarithmic scale with a weighting factor of -2.5 (Schmitt et al., 2014). The threshold value proposed to represent the limit for the probability of sustaining a life-threatening brain injury to be zero is 1000 (Shorten & Himmelsbach, 2003).

$$GSI = \left[\int_0^{\tau} a(t)\right]^{2.5} dt \qquad \textbf{(2.1)}$$

Along with the criticisms from the WSTC, the SI was criticized for predicting high scores for long duration, low intensity impacts which were deemed to be unrealistic (Shorten & Himmelsbach, 2003). Another major criticism that still remains to be verified is the direct link between skull fracture and brain injury (Schmitt et al., 2014).

2.2.3.3 Head Injury Criterion Score (HIC)

In 1972, the National Highway Traffic Safety Administration (NHTSA) proposed a modification to the SI which is currently known as the head injury criterion (HIC). It is currently used to assess the potential for head injury in automobile crash test dummies (Marjoux, Baumgartner, Deck, & Willinger, 2008; Nahum & Melvin, 2002). The HIC is based on equation 2.2. It calculates a weighted time impulse using two arbitrary time points during the acceleration-time curve (t_1 and t_2) which provide the greatest HIC value. The relationship between the probability of head injury and HIC scores are provided by the Prasad-Mertz curves (Prasad & Mertz, 1985) (Figure 2-5). A score of 1000 corresponds with an 18% probability of a severe (AIS

4) head injury, a 55% probability of a serious (AIS 3) head injury, and a 90% probability of a moderate (AIS 2) head injury (Mackay, 2007).

$$HIC_{15} = max[t_2 - t_1]\left(\left[\frac{1}{[t_2 - t_1]}\right]\int_{t_1}^{t_2}[a]dt\right)^{2.5} \; where \; (t_2 - t_1) \leq 15 \, ms \qquad \textbf{(2.2)}$$

Limitations of the HIC are commonly known and include the lack of inclusion of the direction of impact and angular accelerations (Marjoux et al., 2008). Furthermore, how the probabilities of a head injury and their severities shift across populations is unknown (Mackay, 2007).

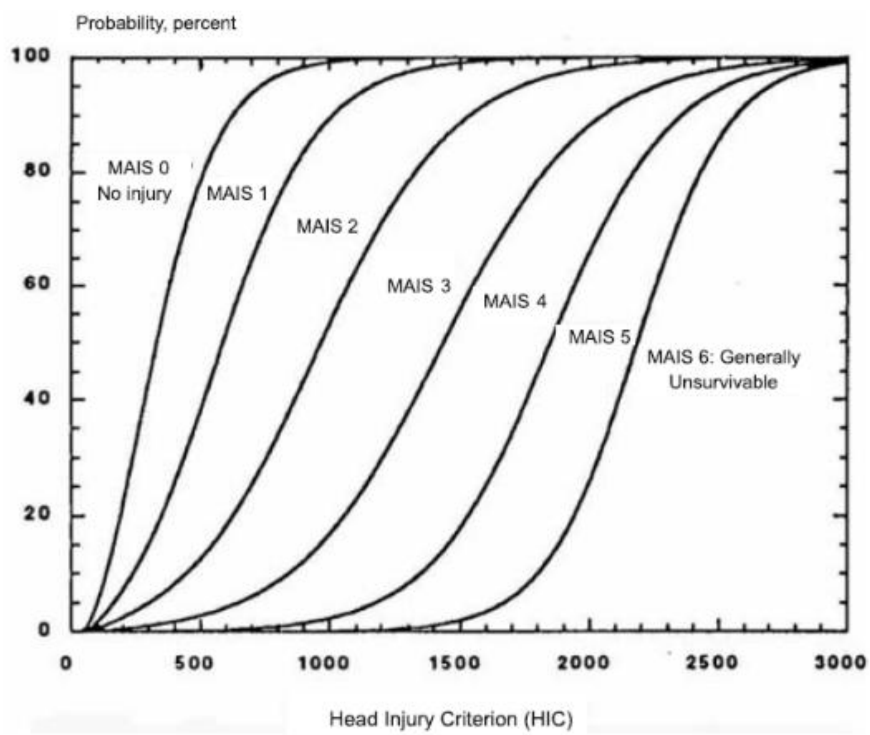


Figure 2-5: The probability of head injury of different severities for given HIC scores (from Mackay, 2007)

2.2.3.4 Newer Predictive Models of Head Injury

Due to the limitations of the previous and current head injury prediction tools (WSTC, SI, and HIC), Newman (1986) proposed the Generalized Acceleration Model for Brain Injury Threshold (GAMBIT). The GAMBIT criteria (Equation 2.3) takes into account both translational and rotational acceleration following the assumption that both loading scenarios contribute to the final injury (Schmitt et al., 2014). The GAMBIT criterion was never properly validated and therefore has hardly been used nor has been included in any regulations (Schmitt et al., 2014).

$$GAMBIT = \left[\left(\frac{a(t)}{250}\right)^{2.5} + \left(\frac{\ddot{\phi}(t)}{25}\right)^{2.5}\right]^{\frac{1}{2.5}} \qquad \textbf{(2.3)}$$

Newman, Shewchenko, & Welbourne (2000) also proposed the head injury power (HIP) criterion which also predicts head injury based on the linear and angular acceleration of the centre of the head (see Equation 2.4). The c_i coefficients are the mass and mass moments of inertia of the head. The HIP includes rotational acceleration as well as translational acceleration making it more applicable to realistic scenarios. The HIP can also account for differences in directional sensitivities of the head to loading if this is known (Newman et al., 2000) perhaps giving it a stronger predictive capacity than the HIC. However, removing the rotational components produced a similar curve. Therefore, the improvements of the HIP to the predictive ability of concussion have not proven to be stronger than the HIC and it is not currently used in any testing standards (Newman et al., 2000).

$$HIP = c_1 a_x \int a_x dt + c_2 a_y \int a_y dt + c_3 a_z \int a_z dt \qquad \textbf{(2.4)}$$

$$+c_4 \alpha_x \int \alpha_x dt + c_5 \alpha_y \int \alpha_y dt + c_6 \alpha_z \int \alpha_z dt$$

The HIC and HIP criteria model the head as a rigid mass without any deformation. Due to the advances in technology and abilities to model the head with finite element models, deformation of the skull and internal components of the head can now be achieved using finite element head models (FEHM). Two recent models using this type of technology are the ULP FEHM, developed at Strasbourg Louis Pasteur University, and the simulated injury monitor (SIMon) FEHM, developed by Takhounts et al. (2003) (Marjoux et al., 2008; Takhounts et al., 2003). However, these methods are very time consuming and costly and therefore are not currently appropriate as regulation head injury prediction tools.

2.2.4 Injury Reference Tools

Since head injury cannot be predicted directly from measuring applied loads, other approaches have to be taken to estimate the probability of injury. Injury risk prediction curves using the HIC score were developed by Prasad & Mertz (1985). Injury risk curves were later developed relating fracture risk to peak acceleration values (Mertz, Prasad, & Nusholtz, 1996). Federal motor vehicle safety standards (FMVSS) use injury assessment reference values (IARV) created by Mertz, Irwin, & Prasad (2003) in their design and testing relating to motor vehicle safety. A proposed tolerance of 700 for HIC represents a 5% chance of obtaining a head injury with an abbreviated injury score (AIS) ≥4 (representing a severe injury) (Mertz et al., 2003). A tolerance level of 180g was proposed based on its injury risk curve as this level has a 5% chance of a skull fracture for a 50th percentile male (Mertz et al., 2003). These IARVs allow for an estimate of head injury probability following impacts using variables that are easy to measure using surrogate headforms in testing situations.

2.3 Helmets and Helmet Testing Standards

Helmets were introduced around the 1970s initially as leather straps worn around the head (Swart, 2003). They did not protect against impacts but did help with sliding along the pavement following a fall or collision. In order to design better helmets for cyclists, it’s important to understand the role of helmets in reducing impact forces and preventing injury. It is also important to consider testing scenarios to create standards which helmets have to pass to certify that they offer the best protection available.

2.3.1 Epidemiology of Helmet Use

Helmet use is an important factor in determining risk of injury to cyclists (Table 2-1). Helmet compliance differs among populations and ranges from 11-82% (Billette & Janz, 2011;

CHIRPP, 2008; City of Toronto, 2011; Linn et al., 1998; Maimaris, Summers, Browning, & Palmer, 1994; Robinson, 2006). It is mandatory in the province of Ontario for riders under the age of 18 to wear a helmet while cycling (Government of Ontario, 2012). Helmet legislation varies across provinces and states and is slowly being introduced worldwide.

Helmets are designed to mitigate head injury and skull fracture which are severe injuries that are costly to the health care system. In 2009-2010, 2.4% of the Canadian population aged 12 and over sustained a head injury (Billette & Janz, 2011). Head injuries can also significantly affect the quality of life of the individual and cause loss of independence representing a significant problem for the individual and their family (Ribbers, 2013).

2.3.2 Helmet Design

Following the initial helmet design of leather straps (otherwise known as a “hairnet”), the “Bell Biker” helmet was introduced and was the first bicycle helmet to contain crushable expanded polystyrene (EPS) with a stiff polycarbonate shell (Swart, 2003). Helmet manufacturers continued to improve upon this concept with current helmet designs typically using slightly different processes to create the EPS foam interior and coating the outside with an ABS plastic outer shell (Swart, 2003). Hard shell helmets have been shown to significantly reduce the incidence of head injury over foam helmets without the hard outer shell (Hansen, Engesæter, & Viste, 2003). Although Snell and the American National Standards Institute (ANSI) had previously published bicycle helmet testing standards, the Consumer Product Safety Commission (CPSC) did not create their first standard until 1999 (Swart, 2003). It was after the release of this standard when most stores would not sell helmets that did not pass a published standard.

Helmets absorb energy through two main mechanisms. The energy absorbed depends on the shape, thickness, and material of the helmet at the impact point. The first mechanism is the foam below the contact area which yields in response to the impact and the second mechanism of energy absorption is the elastic shell deformation surrounding the areas of un-crushed foam (Mills, 1990). The earliest cycling helmets were typically designed for a fall from a bicycle without any third party involvement (Walker, 2005) and current helmets have only built on these initial designs rather than considering specific higher loading scenarios such as cyclist/motor vehicle collisions. Thus, helmets may mitigate injuries during falls or collisions with flat surfaces at lower impact velocities, but the ability to mitigate during higher velocity impacts is unknown (Mills, 1990).

2.3.3 Helmet Efficacy

There have been opposing arguments regarding the efficacy of bicycle helmets as protection during crashes. Early studies on the effectiveness of helmets reported risk reductions in the range of 37-88% for head injury (Attewell, Glase, & McFadden, 2001; McDermott, Lane, Brazenor, & Debney, 1993; Persaud, Coleman, Zwolakowski, Lauwers, & Cass, 2012; Thompson et al., 1996). The Cochrane Collaboration review was criticized by Curnow (Curnow, 2005), who stated that the studies included in the review did not provide scientific evidence that bicycle helmets reduce brain injury since they did not include any knowledge of specific types or mechanisms of brain injury. Other arguments by Curnow (2005) included the possible decline in the number of cyclists following mandatory helmet legislation as well as the possibility that cyclists would engage in riskier behavior if wearing a helmet. The latter arguments have been addressed by multiple studies indicating little relationship between helmet legislation and number of cyclists and no increase in risky behaviour with helmet use (Hagel & Barry Pless, 2006). Although specific mechanisms of brain injury may not be addressed by the design of current helmets (Curnow, 2005), more recent evidence has still supported the efficacy of helmets in decreasing the overall rate of head injury (Benson et al., 2009).

Bicycle helmets are generally designed to mitigate injury after a free fall from approximately 1.5 m above the ground (Mills & Gilchrist, 2006; Walker, 2005). While helmets currently on the market all pass some sort of standard that tests them for this type of injury mitigation, it is also a common occurrence in worst case cycling collisions for a cyclist to impact a car windshield and project into the air before striking the ground. This not only increases the drop height of the cyclist but also can increase the kinetic energy of the collision if the cyclists head impacts the windshield (Mills & Gilchrist, 2006). Therefore, there are still questions regarding the effectiveness of helmets in high energy impact scenarios such as motor vehicle collisions.

2.3.4 Helmet Testing Standards

Although the first bicycle helmet standard was introduced in 1970 by the Snell Memorial Foundation, the common market did not sell helmets that met standards until the mid-1980's (Swart, 2003). Currently, bicycle helmets have to meet one of the following standards to be sold on the market; American Society of Testing and Materials 1446 – Standard Testing in Protective Headgear, Consumer Product Safety Commission 16 CFR Part 1203 – Safety Standard for Bicycle Helmets – Final Rule, Snell B95 – Bicycle Helmet Standards, British Safety Institute EN 1078 – Helmets for pedal cyclists and for users of skateboards and roller skates, Canadian Standard Association D113.2-1989 – Cycling Helmets. The CSA provides the Canadian standard for cycling helmets, although, other standards (listed above) may act as interim standards. Typically, companies will use the CPSC or European standard since these markets are where the helmets are initially manufactured.

All the helmet standards have essentially the same approach to testing the impact performance of bicycle helmets. Each standard uses an artificial headform to which the helmets are fitted as indicated by the manufacturers. The headform is dropped to achieve a certain impact speed chosen to represent the specific application of the helmet. Impact speed, environmental factors and impact surfaces are all considered during this impact. Finally, acceleration of the headform is monitored throughout the impact (Nahum & Melvin, 2002).

Differences in the standards are outlined in Table 2-3. The major differences between the standards are the number of anvils and the drop heights used for testing (Table 2-3). The CSA standard only uses two anvils whereas the others use three. The drop heights range from 1.5 m in the European standards to 2.2 m in the Snell B-95 standard. Since drop heights differ across the standards, choosing one over another could change the outcome of the helmet performance testing. Furthermore, the drop heights chosen in the standards typically represent a fall from a bicycle and thus are indicating the performance of the helmets in this scenario (Walker, 2005). Therefore, they cannot be extended to higher energy impacts such as some cyclist/motor vehicle collisions.

2.4 Mechanical Testing Systems

2.4.1 Drop Towers

The standard method for impact testing of helmets is to use a surrogate headform which is dropped onto a surface in a guided free fall (American Society for Testing and Materials, 2012; Consumer Product Safety Commission, 1998). Two commonly used drop towers to produce this guided free fall include a monorail or a twin guide wire system (Figure 2-6). Both hold the headform rigid during the impact and typically use a uniaxial accelerometer to measure acceleration in the direction of impact (Thorn, Hurt Jr, & Smith, 1998). The European standards use a “guided free fall” system with an unrestrained headform equipped with a triaxial accelerometer (EN, 1997). A comparison of impact tests with motorcycle helmets between the two types of drop towers was done by Thorn, et al. (1998) who found that the monorail system consistently gave higher peak accelerations and suggested that the European system represented a less severe test.

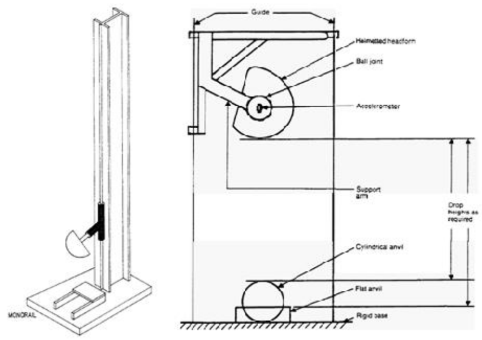


Figure 2-6: Monorail (left) and Twin Guide wire (right) Testing Systems (adapted from ASTM, 2004 and CSA D113.2-M89)

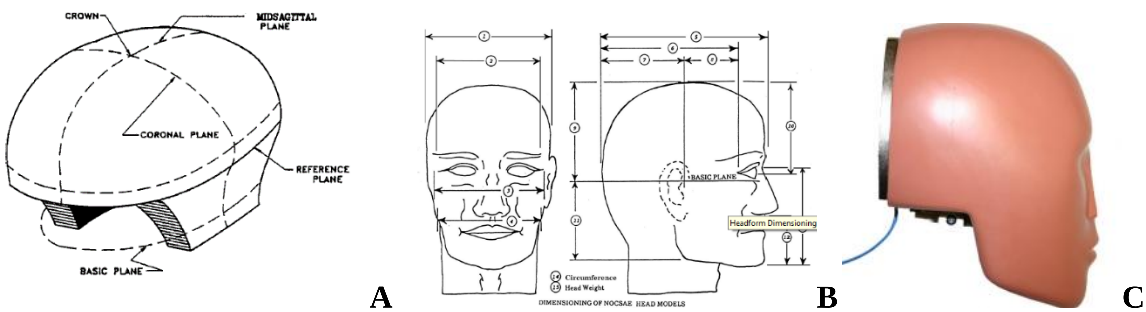


Figure 2-7: Surrogate headforms for head impacts (a - ISO K1A Magnesium b - NOCSAE headform c - Hybrid III headform)

2.4.2 Headforms

Helmet test systems use surrogate headforms to represent the human head in helmet testing. Typical testing standards use metal headforms such as the ISO K1A Magnesium headforms (Figure 2-7a) (American Society for Testing and Materials, 2012). These solid, spherical headforms were designed to produce a more reliable response to impacts and this was done at the expense of replicating the natural response of the head during impact (Hoshizaki & Brien, 2004; Kendall, Walsh, & Hoshizaki, 2012).

The Hybrid III headform (HIII) (Figure 2-7c) was developed as part of a General Motors project to build a crash test dummy for simulating humans in car collisions (Kendall et al., 2012). It was designed to be able to withstand high impact scenarios without breaking but still replicate some of the data found in cadaveric head impact studies (Kendall et al., 2012). The NOCSAE biofidelic headform (Figure 2-7b) is constructed with a high durometer, urethane skull covered by a lower durometer urethane that forms anatomical features such as the ears and lips. This headform also has a glycerin-filled brain cavity to simulate the behaviour of the brain during impact (National Operating Committee on Standards for Athletic Equipment, 2009). The NOCSAE headform is currently used in NOCSAE testing standards (e.g. for football, baseball, equestrian helmets) (Kendall et al., 2012; National Operating Committee on Standards for Athletic Equipment, 2009). It is considered to have more human-like characteristics based on its gel-filled brain cavity and human anthropometry. However, it is typically only used with a single accelerometer which does not allow calculations of three-dimensional impact responses making it difficult to justify the use of the NOCSAE headform with three-dimensional finite element models of the brain (Kendall et al., 2012).

Table 2-3: Summary of Testing Standards

Standard	ASTM 1446/1447	CSA D113.2-M89	SNELL B-95	CPSC – 16 CFR Part 1203	BSI – EN 1078
Date last updated	2012	1989 (reaffirmed 2009)	1995	2012	2012
Type of Headform	ISO K1A-F Mg	ISO K1A-F Mg (can also use Al alloy)	Metal headforms	ISO K1A-F Mg	Metal headforms – low resonance frequency
Type of Drop Tower	Guided free fall using monorail or twin wire	Guided free fall with twin wire	Guided free fall using monorail or twin wire	Guided free fall using monorail or twin wire	Guided free fall using twin or triple wire guide system
Drop Assembly Weight	5 ± 0.1 kg	5 ± 0.1 kg	5 ± 0.1 kg	5 ± 0.1 kg	3.1-6.1 kg
Drop Heights	Flat anvil - 2.0 m	Flat anvil – 1.66m	Flat anvil – 2.2m	Flat anvil – 2.0m	Flat anvil – 1.5m
	Curbstone – 1.2 m	Cylindrical – 1.13m	Curbstone – 1.3m	Curbstone – 1.2m	Curbstone – 1.5m
	Hemispherical – 1.2 m		Hemispherical – 1.3m	Hemispherical – 1.2m	
Impact Energy	54 J	55J	Flat - 110J	Flat – 90J	Not specified
			Curbstone – 72J	Curbstone – 56J	
			Hemispherical – 72J	Hemispherical – 56J	
Head Orientations	4 sites (front, back, side, one choice) – at least 1/5 of circumference away from each other	4 sites (front, back, side, one choice) – at least 1/4 of circumference away from each other	4 sites (front, back, side, one choice) – at least 120 mm of circumference away from each other	4 sites (front, back, side, one choice) – at least 120mm of circumference away from each other	Not specified
Number of samples	8 samples of each helmet (2 for each conditioning environment)	8 samples of each helmet (2 for each conditioning environment)	5 samples destroyed in testing - a 6th to compare to	8 samples of each helmet (2 for each conditioning environment)	10 samples of each helmet (2 impacts to each)
Conditioning Environments	"Normal", hot, cold, wet	"Normal", hot, cold, wet	"Normal", hot, cold, wet	"Normal", hot, cold, wet	"Normal", hot, cold, aging
Impact Surface	3 anvils (made of steel, flat, hemispherical and curbstone)	2 anvils (made of steel - flat, cylindrical)	3 anvils (made of steel, flat, hemispherical and curbstone)	3 anvils (made of steel, flat, hemispherical and curbstone)	2 anvils (made of steel, flat and curbstone)
Instrumentation	Uniaxial accelerometer (capable of withstanding 1000 g)	Uniaxial or triaxial accelerometer (capable of withstanding 1000 g)	Uniaxial accelerometer (capable of withstanding 1000 g)	Uniaxial accelerometer (capable of withstanding 1000 g)	Triaxial accelerometer (capable of withstanding 2000 g)
Outcome Measures	Peak acceleration	Peak acceleration	Peak acceleration	Peak acceleration	Peak acceleration
Tolerance Level (g)	300	Flat – 200	300	300	250
		Cylindrical - 150			
Testing Environment	Temperature: 17-23°C	Temperature: 20±5°C	Temperature: 17-23°C	Temperature: 17-23°C	Not specified
	Humidity: 25-75%	Humidity: 60±5%	Humidity: 25-75%	Humidity: 25-75%	
	Pressure: 75-110 kPa		Pressure: 75-110 kPa	Pressure: 75-110 kPa	
Velocity Measured	Last 40 mm of free fall (±3% of velocity)	Last 40mm of free fall (±3% of velocity)	Last 40mm of free fall (±1% of velocity)	Last 40mm of free fall (±3% of velocity)	Last 60mm of free fall (±1% of velocity)

Two studies have compared dynamic responses of these headforms. Kendall et al. (2012) compared peak linear and angular accelerations of the NOCSAE and HIII headforms impacts to the front and side of the head. Both headforms produced linear and repeatable data. However, significant differences were found for both linear and angular acceleration between the two headforms with the NOCSAE headform having higher peak acceleration values than the HIII headform suggesting that differences in biofidelity (Kendall et al., 2012). This is an important concern when considering use of these headforms in standardized testing. Kendall, et al. (2012) limited their study to include only two impact orientations. They also used drop heights that were less than half the heights used in the current testing standards which are not representative of cycling collisions (Kendall et al., 2012; Walker, 2005). Finally, only peak linear and angular accelerations were measured. Other outcome variables for estimating head injury were not included. Therefore, there are still questions regarding the use of these headforms in bicycle helmet standards.

Stuart, Cripton, Dressler, Dennison & Richards (2013) compared the HIII headform to the standard K1A magnesium headform for drop heights spanning the current helmet testing standards and also included three impacts from heights above the current standards (up to 3 m). They found that the magnesium headform had higher peak acceleration and HIC values than the HIII headform and that the HII headform produced repeatable results within ±5%. Although this study provided some insight into the usefulness of the HIII headform in bicycle helmet testing, this study only used frontal impacts and also only included unhelmeted trials (Stuart et al., 2013). Therefore, there are still questions regarding the efficacy of the HIII headform for side and back of the head impacts and while wearing a bicycle helmet.

2.5 Literature Review Summary

Cycling falls and collisions represent an ongoing problem for the Canadian health care system specifically for costs involved with injuries to the head which are one of the most common types of injuries sustained by cyclists. In order to quantify risks associated with cycling, various methods of injury reporting techniques (e.g. police reports and hospital databases) have been used to publish epidemiological literature in Canada involving cycling. However, these reporting techniques have been shown to be inaccurate or incomplete at times creating difficulty for researchers to understand all factors or mechanisms behind cyclist falls and collisions (Rosman & Knuiman, 1994). Forensic engineering companies provide a unique opportunity to obtain information about all factors in a cycling incident (crash circumstances, cyclist and driver characteristics, and injury characteristics). Despite these unique datasets, to date, databases of forensic engineering companies have not been used in epidemiological studies involving cyclists. Furthermore, Southern Ontario is lacking in regional data concerning detailed cyclist characteristics. Regional data would be beneficial when considering infrastructure design and safety campaigns.

In previous epidemiological studies summarizing cycling risk factors, it has been reported that two of the most detrimental factors include cycling without a helmet and colliding with a motor vehicle (Rivara et al., 1997). Helmets are designed to mitigate injuries to the head. However, current testing standards have limitations that may limit their application to real-life scenarios. Canada has its own cycling helmet safety standard (D113.2-1989; (Canadian Standards Association, 2009); although, due to overlaps in market distribution, the Canadian standard allows the use of alternative testing standards (e.g. American or European) in place of their own. Standards impact helmets using a magnesium headform which was initially incorporated into the methods because it produced highly repeatable and reliable results. However, two other headforms (by NOCSAE and General Motors) have been developed that may represent a human head more closely (Kendall et al., 2012). Two studies compared responses of different combinations of the three headforms. However, to date, no study has compared impacts to all headforms, at all impact velocities, with and without the use of a helmet.

Cycling collisions with motor vehicles may increase injury severity up to 3.6 times, likely due to the increased energy involved in such collisions (Rivara et al., 1997). Currently, cycling helmet test standards use impact velocities of ~5.4-6.6 m/s. These impact velocities represent the lower end of velocities determined in a previous study investigating head impact velocities following hypothetical cyclist/motor vehicle collisions (Fanta et al., 2013). Therefore, it is unknown if cycling helmets mitigate injury at impact velocities more representative of higher energy scenarios such as collisions with motor vehicles.

My thesis was designed to address these gaps in literature. First, it summarized cyclist incidents that were collected from a database at a forensic engineering company located in Southern Ontario. Using a subset of these case files, models were created to approximate head impact velocities during cyclist/motor vehicle collisions in the region. Second, my thesis compared responses from three common surrogate headforms for helmeted and unhelmeted impacts. Finally, the mitigating capacity of three brands of cycling helmets were evaluated during impacts at velocities currently used in testing standards as well as one more representative of collisions with motor vehicles (as determined in study one).

These studies will provide regional data for cycling collisions in Southern Ontario as well as investigate aspects of cycling helmet testing standards that may not be representative of real-life cycling scenarios. Finally, it will contribute to our understanding of helmet mitigating capacity and provide useful information for future cycling helmet design.

Chapter 3: Analysis of an Expert Database on High Severity Cycling Collisions

3.1 Background

Cycling is a common recreational activity, sport and mode of transportation for which helmets are designed and marketed for head injury mitigation. The City of Toronto ranked among nine large North American cities for higher trends of workers commuting by bicycle (Pucher et al., 2011). Furthermore, a 42% increase was found in the number of Canadian commuters using bicycles between the years 1996-2006 (Pucher et al., 2011). Due to the number of commuters on bicycles, related injuries have become an important public health issue. Injuries following cycling collisions ranked second to those due to motor vehicle incidents for non-fatal transport related injuries in 2009. As a consequence, these injuries were associated with $443 million to the Canadian health care system (SMARTRISK, 2009). Various countries/regions have previously examined cyclist road risk in their respective populations including France, Sweden, USA, and British Columbia, Canada (Amoros et al., 2011; Haileyesus et al., 2007; Isaksson-Hellman, 2012; Linn et al., 1998; Lustenberger et al., 2010). While it is helpful to incorporate the findings from these studies into the development of programs and improving cycling infrastructure in Ontario, the results may not fully represent the cyclist road risk in Ontarian cities.

Risks to cyclists increase in urban areas where there are more commuters on bicycles and traffic is more dense (Amoros et al., 2011; Pucher et al., 2011). This is reinforced in Canadian statistics with Ontario reporting the largest number of cycling incidents causing hospitalization in 2011 (~1300 incidents) (CIHI, 2011). Other risk factors for cycling incidents include gender, competitive riding, riding in poor lighting conditions, motor vehicle involvement, and non-helmet use, with the latter two as the most detrimental risk factors (Amoros et al., 2011; Chow et al., 1993; City of Toronto, 2011).

Two main types of injury reporting exist in Canada. The first is through police services who report on all collisions that fall under the Highway Traffic Act (HTA). Information from these reports, such as time of day, age, gender and helmet use, can be helpful to transport safety professionals for designing appropriate city infrastructure and indicating at-risk groups for cycling incidents. However, police reports of cycling incidents have been shown to be inaccurate, incomplete, or underreported (Rosman & Knuiman, 1994). The second method of injury reporting is completed through analyzing hospital databases (e.g. National Trauma Registry) (CIHI, 2013). These reports may describe similar details as police services, such as cyclist age, gender, time of day, and helmet use but they also typically contain specific information about cyclist injuries. While both sources provide some background about cycling/motor vehicle collisions, all factors about the collisions (e.g.cyclist and vehicle speed at impact, collision orientations etc.) are not usually available from these sources leaving much of the current epidemiological literature incomplete (Elvik & Mysen, 1999).

Finally, epidemiological literature summarizing cycling collisions, specifically in Southern Ontario, is scarce. As such, transport safety professionals have little information to design infrastructure and prevention controls specific to the area. In order to fully understand the characteristics of cycling collisions with motor vehicles, reports containing all factors of realistic impact scenarios must be obtained.

Forensic engineering companies are often hired on behalf of individuals involved in cyclist collisions. These companies have the ability to obtain information from both police and hospital reports as well as have access to information provided by the parties (such as witness statements, examinations, etc.). This gives them the unique opportunity to match reported crash circumstances with injury outcomes and reconstruct the incidents to estimate more detailed collision characteristics such as vehicle/cyclist impact speeds and orientations. These firms have rarely reported their unique collection of data on cyclist collision characteristics in the literature. Furthermore, links between injury outcome and collision characteristics such as speed and orientation have not often been drawn from realistic cyclist/motor vehicle collisions causing injury claims such as those recorded by these companies.

Estimating head impact velocities during real-life scenarios involving a motor vehicle is complex due to the variety of impact scenarios (e.g. orientations of the vehicles at impact, size of the vehicle and cyclist). Multi-body modeling software programs such as PC Crash or its add-on, MADYMO (Mathematical Dynamical Model) (Datentechnik Group, Linz, Austria) or custom finite element programs have previously been used to estimate head velocities of cyclists prior to impact (Bourdet et al., 2012; Fanta et al., 2013; Ito, Yamada, Oida, & Mizuno, 2014). The use of multi-body modeling software to simulate cycling/motor vehicle collisions is a relatively new concept that allows investigators to reconstruct collisions and approximate velocities of specific body regions such as the head which is not available from hospital or police-reported collision data. Bourdet, Deck, Carreira, & Willinger (2012) used MADYMO and estimated an average head impact velocity of 5.5 m/s in impact configurations not involving a motor vehicle. However, in two recent models of cyclist/motor vehicle collisions also using MADYMO, Fanta et al. (2013) found head impact velocities ranging from 4.6-13.5 m/s and Ito, Yamada, Oida & Mizuno (2014) found a head impact velocity of 11.6 m/s, both of which are significantly higher than the 5.7 m/s used in current testing standards. However, both of these models were simulations of hypothetical rather than real-life scenarios.

3.2 Purpose and Hypotheses

The current study was split into two experiments. The goal of the first experiment was to characterize typical impact scenarios of cyclist collisions/incidents (with and without motor vehicles) resulting in injury claims in Southern Ontario. Specifically, the study involved analyzing a database from a professional forensic engineering firm for a list of variables describing the circumstances of the crash (e.g. orientation of the vehicles upon impact, road conditions, velocities of the vehicles, etc.), characteristics of the cyclist and driver (age and gender), and nature of the cyclists injuries (helmet/no helmet, severity of injury, type of injury, etc.). The output variables (Table 3-1) were chosen based on factors influencing collision outcomes that would be relevant to the automotive industry, the helmet industry as well as for biomechanical research interests. For the remainder of the document, the cases involving a motor vehicle will be referred to as 'MV cases' and those without motor vehicle involvement will be referred to as 'nMV cases'.

The primary objective (descriptive analysis) of this experiment was to describe the characteristics associated with cycling collisions causing injury claims in Southern Ontario (persons involved, collision details, injuries). The secondary objective (inferential analysis) of the experiment was to determine whether relationships exist between injury circumstances and resulting injury outcomes (e.g. impact location and Abbreviated Injury Score (AIS)). Three main hypotheses were chosen for this objective based off of previous findings in literature. However, the data set was also analyzed to see if any other relationships existed between injury circumstances and outcomes.

Specific hypotheses for the second objective were:

1) ***Impacts to stiffer materials*** *such as the vehicle frame will be associated with* ***higher AIS scores*** *compared with impacts to less stiff materials such as the hood or body of the vehicle*. Stiffer materials will not absorb energy as well as materials that are more compliant. Therefore, more energy will be transferred to the cyclist during these impacts creating a higher chance of obtaining a more severe injury.
2) ***Helmet use*** *will be associated with* ***lower head-specific AIS scores*** *for* ***a) all cases (MV and nMV) and b) for MV cases alone***. Although there has been debate as to the efficacy of helmets, risk reductions of 37-88% have been estimated in previous studies for the effect of helmets on reducing head injury (Attewell et al., 2001; McDermott et al., 1993; Persaud et al., 2012).
3) ***Higher vehicle speeds at impact*** *will be associated with* ***higher AIS scores***. Higher vehicle speeds have previously been correlated with an increased risk of injury to the cyclist (Simms & Wood, 2009).

The purpose of experiment two was to model a subset of cases in experiment one (only those involving a cyclist/motor vehicle collision resulting in head impacts) to estimate a head impact velocity that is more relevant of higher energy cycling head impacts. Hereafter, the higher energy impact velocity will be referred to as the 'MV velocity'.

3.3 Methods

3.3.1 Study design

Giffin Koerth Smart Forensics is a professional forensic engineering firm based in Toronto, Ontario, Canada. They are typically hired by lawyers and insurance companies to answer questions related to personal injury cases and give an expert opinion on potential contributors to the injuries (Giffin Koerth Forensic Engineering, 2013). Each case has a paper file associated with it and is entered into an internal database. The database is searchable by fields such as the start/finish dates, number of hours, budget, supervising engineer and project description.

3.3.1.1 Experiment one: collision characteristics in Southern Ontario

The database of Giffin Koerth (hereafter referred to as the "GK database") was searched in its entirety (encompassing 14 years (2000-2013)) using a "match any" search strategy with the terms "helmet", "cyclist", "bike", and "cycle" on November 5, 2013. Specifically, the project descriptions for each case file were searched to find descriptions containing any of the search terms. The search returned 301 results which were consolidated and reviewed to remove any files that were not relevant to the study (e.g. files involving motorcycle collisions). The 139 remaining files were reviewed for output variables describing the circumstances of the crash, characteristics of the cyclist and driver, and nature of the cyclists' injuries (Table 3-1). Each file was documented using the project number to keep driver and cyclist information confidential. For each case, detailed collision information was coded using reconstruction reports completed by Giffin Koerth. When a report was not written, information was collected from available police and hospital reports. Finally, if needed, supplemental information was recorded from witness statements. Of the 139 files, 18 files did not focus on the cyclist or their injuries, 4 cases had two files associated with them (as such, the files were combined into one case), and 7 files were not available during the collection period. These files were removed from the study leaving 110 files. One special case had two cyclists involved, each with injuries. This case was divided into two separate cases creating a final collection of files of 111 which were then sorted into two groups depending on the circumstances of the collision, with (N = 86) and without (N = 25) motor vehicle involvement.

Injuries to the cyclist were documented and rated by the primary investigator on the Abbreviated Injury Scale (AIS) developed by the American Association for Automotive Medicine (Gennarelli & Wodzin, 2006). The AIS has six levels: 1 (minor), 2 (moderate), 3 (serious), 4 (severe), 5 (critical), 6 (beyond treatment) (Amoros et al., 2011). Injuries were separated into the following regions of the body before rating: upper and lower extremity, head, face, trunk, pelvis, internal injuries, spinal cord, spine, and fatal. Fatal injuries were classified as a score of AIS 6 (Amoros et al., 2011). In order to measure the whole-body severity of the injuries, the maximum AIS (MAIS) score was also documented (Amoros et al., 2011). To relate factors to injury risk during the inferential portion of the study, the injuries were collapsed into two categories, AIS <3 (mild and moderate injuries) and AIS 3+ (severe or worse injuries).

The final collection of data was reviewed using cross-tabulations and then sorted into more general categories based on the spread of information and from groupings previously used in the literature. The categories of each variable along with their respective references can be found in Appendix A.

Table 3-1: Output variables for the search within the GK database

Output Variables from Giffin & Koerth Project Files	
Date of Loss	Specific Cyclist Injuries (AIS Score)
Location of Incident	Primary Cyclist Impact Site
Driver Age	Secondary Cyclist Impact Site
Driver Gender	Head Impact Occurrence (Yes/No)
Cyclist Age	Orientation of Head at Impact (if applicable)
Cyclist Gender	Helmet Used (Yes/No)
Cyclist Height	Type of Helmet (if worn)
Cyclist Weight	Damage to Helmet
Month	Cyclist Speed at Impact
Time of Day	Vehicle Speed at Impact
Road Condition	Location of Incident
Type of Bicycle	Direction of Vehicles at Impact
Colour of Bicycle	Primary Vehicle Impact Site
Reflective Lights Used (Yes/No)	Secondary Vehicle Impact Site
Location of Reflective Lights (if used)	Throw Distance of Cyclist
Type of Vehicle Involved	Source of Injury (e.g. ground vs. vehicle)
Cyclist Injury (MAIS Score)	Type of Collision (e.g. cyclist/MV, cyclist/no MV)

3.3.1.2 Experiment two: determining realistic head impact velocities for high severity cyclist/motor vehicle collisions

The 86 cases from experiment one were searched for all files that fit the following inclusion criteria: (1) the files must involve a cyclist impact with a motor vehicle, (2) the cyclist must have incurred a head impact from either contact with the motor vehicle or the ground, and (3) there must be available injury information for the cyclist

Of the cases reviewed, 48 met the inclusion criteria. The 48 files were then searched and were selected for reconstruction if: (1) the files included vehicle and cyclist speeds at impact, (2) the impact locations between the cyclist and the vehicle were clearly identified in the file, and (3) only one cyclist and vehicle were involved in the collision. Out of the 48 files, 30 met the requirements and were selected for reconstruction.

The 30 files were modeled using the software PC Crash (Version 9, Datentechnik Group, Linz, Austria). PC Crash is a three dimensional software that contains a multi-body component to include pedestrians or cyclists. Motor vehicles from each respective case were modeled using vehicle specification data from the Expert AutoStats database (Version 5.1.1, La Mesa, CA, USA). This database provided the investigator with vehicle dimensions that were used as inputs for the vehicle type in each model. The bicycle model was the same for all cases. It was a predesigned model that used ellipsoids for the wheels, saddle, seat stay, and the frame. Joints connected the wheels to the frame and were ‘unlocked’ to allow movement at the wheels.

Cyclists were modeled using the multi-body component which incorporates 20 ellipsoids connected by 19 six-degree of freedom joints (Figure 3-1). The mass and geometry of each ellipsoid can be determined by the user but there are no scaling factors available (e.g. for a 50^{th} % male). The mass of the segments were changed for each cyclist based on their individual anthropometric values (McDowell & National Center for Health Statistics (US), 2008). The geometry of the ellipsoids were left the same for each model as changing the geometry significantly increased the complexity and time to create each model. Joint stiffnesses were not changed for any of the ball and socket joints to simplify the model and because the models were only created to examine a short duration impact period. A fixed head/neck joint is the default setting in the multi-body model. This setting was turned off for all models to allow the head to rotate about the neck upon impact.

Figure 3-1: Average male cyclist and bicycle in PC Crash 9.0

The orientation and position of the cyclists with respect to the vehicle at impact were approximated based on the collision reconstruction reports completed by GK. Vehicle and cyclist speeds were assigned based on information from the files (Appendix B). All models were designed so the cyclist and motor vehicle were oriented to the vehicle directions just prior to impact using the available data in the case files and reconstruction reports. The braking power of the vehicles was always set to 100% in the software; thus, assuming that the drivers were aware they had struck a cyclist. The velocity ($Head_{vel}$) and contact force of the head segment were tracked throughout the simulation and $Head_{vel}$ was recorded just prior to the initiation of contact force between the head and the respective surfaces for each case (e.g. ground, windshield). Appendix C shows examples of time-varying head impact force and head velocity traces for one reconstructed collision. The head impact velocity ($Head_{vel}$) was averaged for the 30 files and used as an input into study two (Chapter 4). The average value was chosen to capture a representative head impact velocity of all the modeled cases.

Towards assessing the general validity of the head velocities estimated through the reconstruction process, correlation coefficients were calculated between the model output ($Head_{vel}$) and head injury severity score (head AIS score) obtained from each claims file. The assumption was higher head AIS scores would be associated with higher estimated head impact velocities ($Head_{vel}$).

3.3.2 Statistical analysis

For the descriptive portion of experiment one, means, standard deviations, and ranges were calculated for cyclist ages. Frequency distributions were calculated for all categories of variables. For the inferential portion of experiment one, χ^2 tests were completed by comparing the ratio of 'serious' (AIS 3+) injuries over the total number of injuries for the following variables:, source of injury, helmet/no helmet status, vehicle speed, cyclist speed and age, gender, and type of vehicle. When χ^2 tests showed significance for variables that included more than two categories, χ^2 residuals were reviewed to assess the likely contribution of each category to the outcome of the test.

For experiment two, a correlation coefficient was calculated between the head impact velocities estimated through the collision reconstruction process and their associated head injury severity score (AIS score). Specifically, a Pearson product moment correlation was completed between the output of the model (head impact velocity ($Head_{vel}$)) and head AIS score.

Descriptive analyses were completed in Microsoft Excel 2010 (Microsoft, Redmond, WA, USA) while χ^2 tests and correlations were performed using SPSS statistical software package ($\alpha < .05$) (Version 19.0, SPSS Inc., Chicago, IL, USA).

3.4 Results – Experiment One - Descriptive Analysis

The 111 cases occurred between the years 1998-2013.

3.4.1 Cyclist Characteristics

For MV and nMV files, the mean (SD) ages of the cyclists were 30.1(18.2) and 39.0(14.2) years, respectively. Males dominated the gender distribution for cyclists for both MV and nMV cases. Of the 86 MV cases, 68 (79.2%) of the cyclists were male and most cyclists fell in the age group 16-34 years (Table 3-2). For nMV cases, 64.0% of the cyclists were male with the majority of cyclists (36.0%) falling in the age group 35-54 (Table 3-2).

Table 3-2: Cyclist characteristics

		MV Collisions (N=86)	**nMV Collisions (N=25)**
		Count (%)	Count (%)
Age (years)	<16	19(22.1)	1(4.0)
	16-34	31(36.0)	6(24.0)
	35-54	19(22.1)	9(36.0)
	55+	9(10.5)	2(8.0)
	NA	8(9.3)	7(28.0)
Gender	Male	68(79.0)	16(64.0)
	Female	17(19.8)	8(32.0)
	NA	1(1.2)	1(4.0)

3.4.2 Crash Characteristics

All collisions occurred in Ontario with the majority taking place in the Greater Toronto Area (GTA). Figure 3-2 illustrates the locations of MV and nMV collisions on a map of Ontario. Most collisions occurred during August and September (Figure 3-3) with many of them (36.9%) falling on Fridays and Saturdays (Figure 3-4).

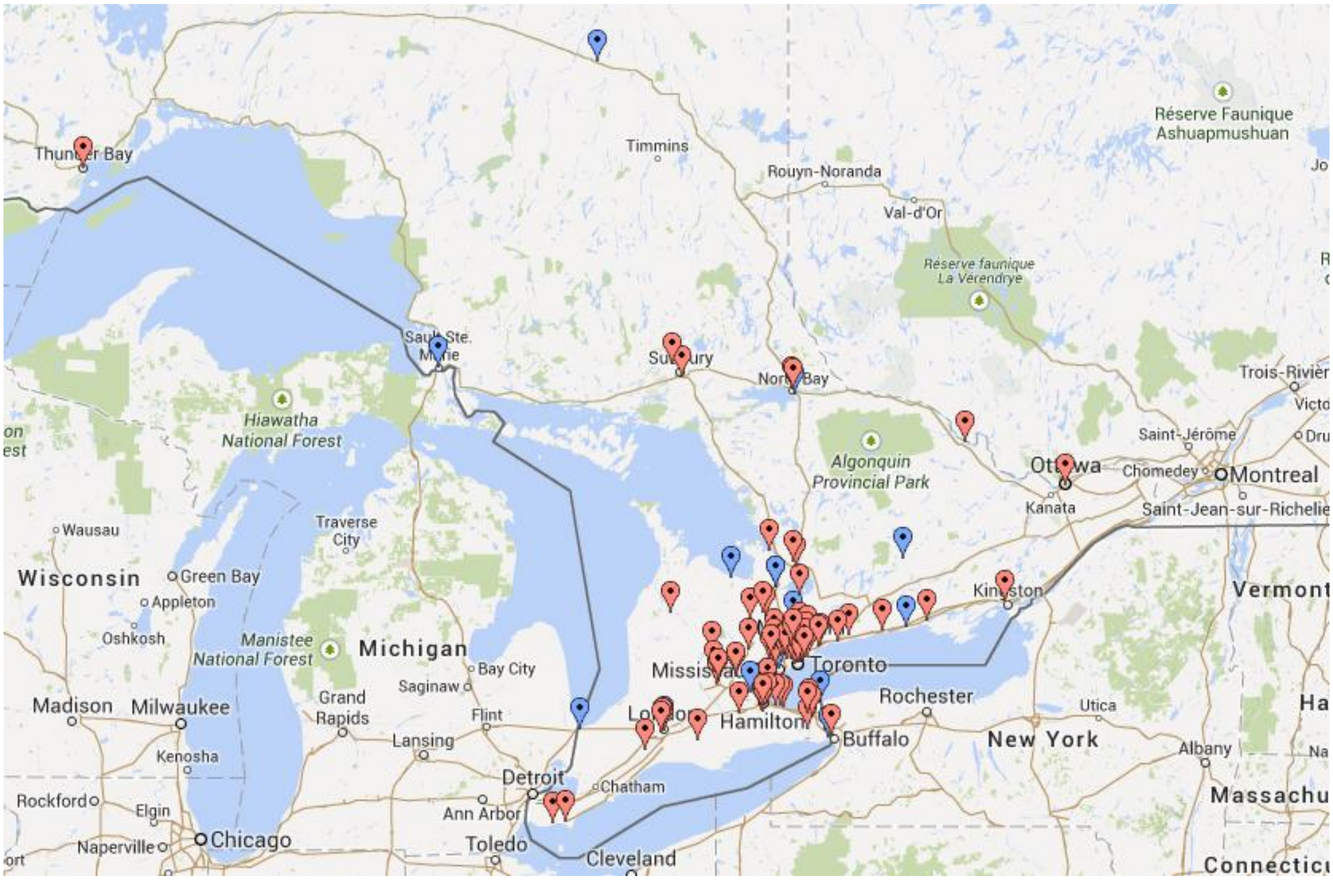


Figure 3-2: Locations of all MV (red markers) and nMV (blue markers) Collisions

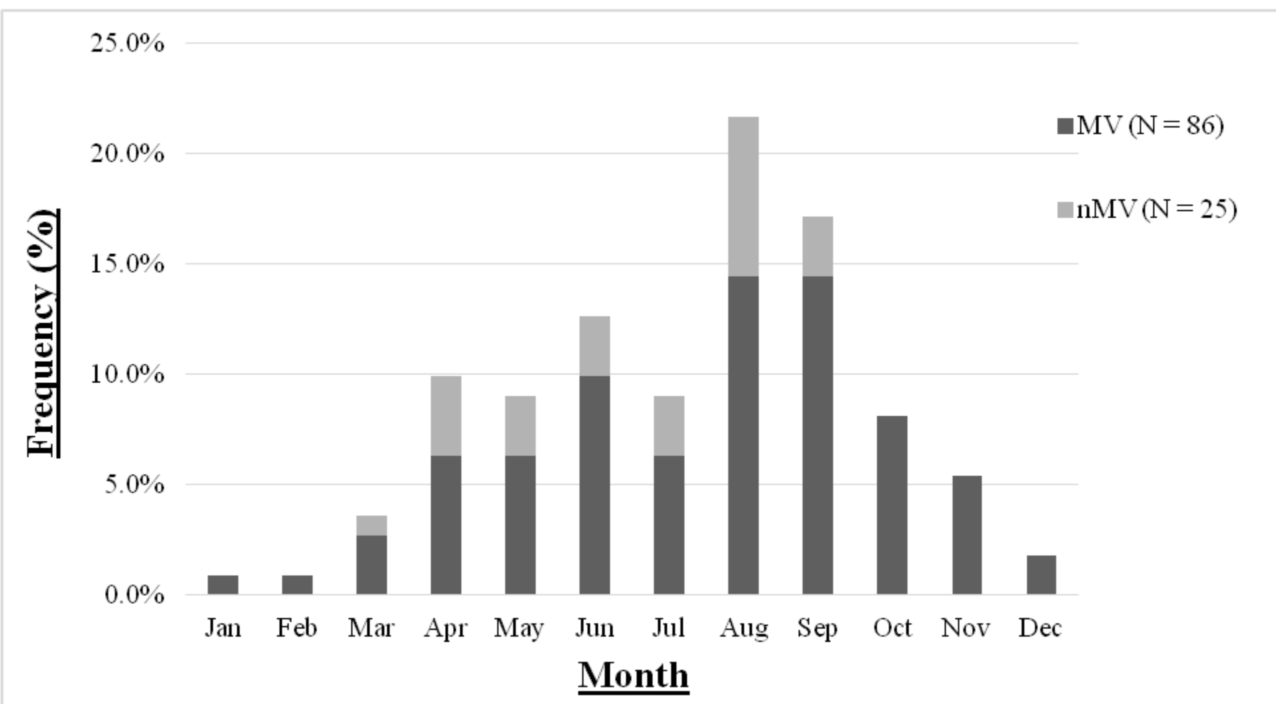


Figure 3-3: Crash distribution over the year

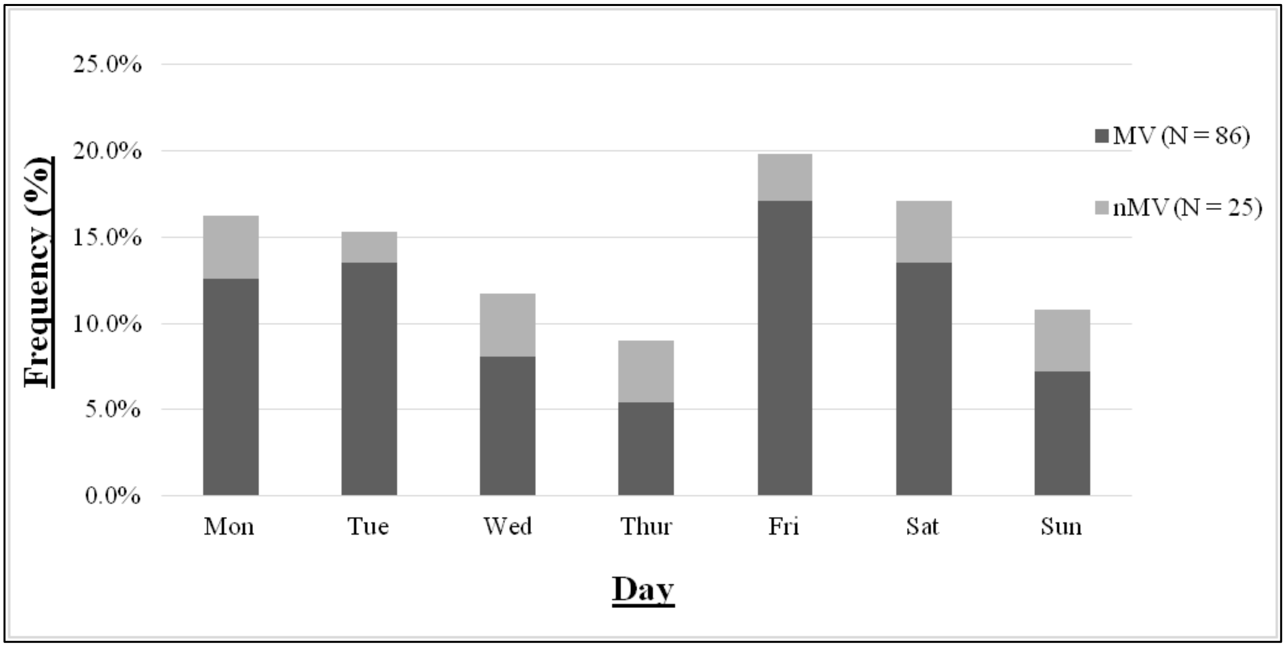


Figure 3-4: Crash distribution over the week

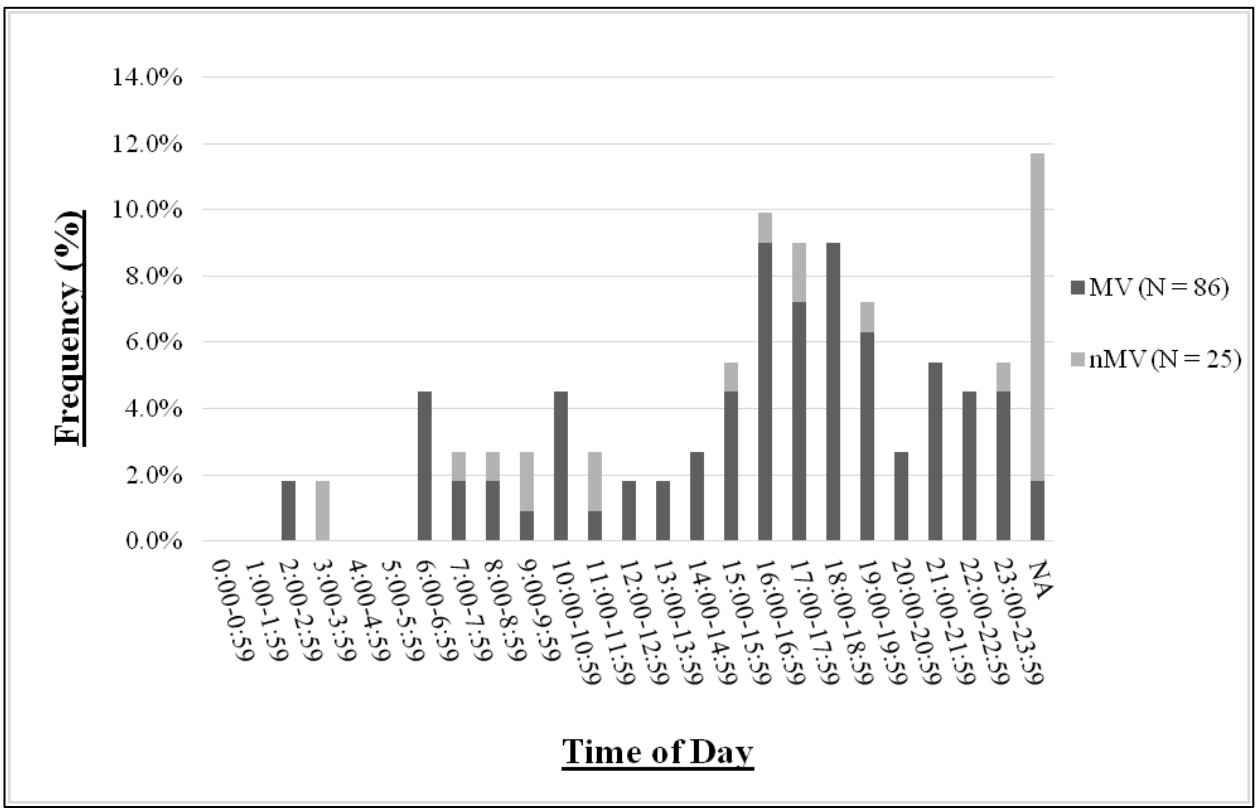


Figure 3-5: Crash distribution over the course of the day (NA indicates that the data was not available in the file)

The crash distribution over the day shows that most collisions (35.1%) occurred between the hours of 4-8 pm (Figure 3-5). Most collisions occurred during daylight hours (45.0%) and while the road surface was dry (56.8%, Table 3-3). Mountain and road type bicycles were the most frequent types of bicycles with 42.3% and 18.9%, respectively. Sedans, SUVs, and pickups were the most common vehicle types in the MV collisions, contributing to 73.3% of all vehicles (Table 3-3).

The majority of cyclist speeds (61.3%) were not determined from reviewing the files. Of the known speeds, most cyclist speeds fell within 11-20 km/h (Table 3-3). The majority of the vehicles were travelling below 60 km/h at impact with 22.1% and 20.9% travelling between 21-40 and 41-60 km/h, respectively (Table 3-3).

Many of the MV and nMV collisions occurred on urban, municipal, or residential roads with only 28.8% of incidents occurring in rural areas, service roads, or highways (Table 3-3). With respect to the direction of vehicles at impact in the MV conditions, more than three quarters of the cyclists and motor vehicles were oriented obliquely at impact (typically the bicycle was hit on the side by the front of the vehicle) (Table 3-3). Of the MV impacts, many occurred in pedestrian crosswalks and in the curb lane of the roadway at 29.1% and 41.8%, respectively.

Relating to the head impacts and helmet use among the cyclists, the majority of cyclists did not wear a helmet with less than one quarter of the cyclists wearing a helmet in both MV and nMV conditions combined (Table 3-4). Despite this, 65 of 111 cyclists experienced one or more head impacts during the collision. Of these 65 cyclists, the orientation of the head upon impact was fairly well distributed across the front, back, side, and face with the majority of the impacts occurring at the side of the head (20.0% for nMV+MV cases, Table 3-4).

Table 3-3: Crash characteristics of all cases

		MV Cases (N = 86)	MV + nMV Cases (N = 111)
	Categories	**Count (%)**	**Count (%)**
Lighting Conditions	Dark	20(23.3)	23(20.7)
	Daylight	38(44.2)	50(45.0)
	Dusk/Dawn	9(10.5)	9(8.1)
	NA*	19(22.1)	29(26.1)
Road Conditions	Dry	54(62.8)	63(56.8)
	Wet	13(15.1)	13(11.7)
	Icy	1(1.2)	1(0.9)
	Gravel	0(0.0)	1(0.9)
	NA*	18(20.9)	33(29.7)
Bicycle Type	Road	15(17.4)	21(18.9)
	Mountain	37(43.0)	47(42.3)
	Hybrid	10(11.6)	14(12.6)
	BMX	10(11.6)	10(9.0)
	Child	2(2.4)	3(2.7)
	Electric	1(1.2)	2(1.8)
	NA*	11(12.8)	14(12.6)
Vehicle Type	Sedan	41(47.7)	NA
	Station Wagon	2(2.3)	NA
	SUV	11(12.8)	NA
	Pickup	11(12.8)	NA
	Minivan	5(5.8)	NA
	Van	3(3.5)	NA
	Bus	2(2.3)	NA
	Large Truck	8(9.3)	NA
	SUV with trailer	1(1.2)	NA
	NA*	2(2.3)	NA
Cyclist Speed (km/h)	0-10	6(7.0)	6(5.4)
	11-20	23(26.7)	23(20.7)
	21-30	10(11.6)	12(10.8)
	>30	1(1.2)	2(1.8)
	NA*	46(53.5)	68(61.3)
Vehicle Speed (km/h)	0-20	13(15.1)	NA
	21-40	19(22.1)	NA
	41-60	18(20.9)	NA
	61-80	5(5.8)	NA
	>80	3(3.5)	NA
	NA*	28(32.6)	NA

* NA indicates the data was not available in the file

	Categories	Count (%)	Count (%)
Road Type	Urban	12(13.9)	16(14.4)
	Municipal	33(38.4)	35(31.5)
	Residential	21(24.4)	25(22.5)
	Rural	13(15.1)	18(16.2)
	Highway	3(3.5)	3(2.7)
	Service	3(3.5)	3(2.7)
	Private	1(1.2)	2(1.8)
	Recreational	0(0.0)	6(5.4)
	NA*	0(0.0)	3(2.7)
Direction of Vehicles (at impact)	Oblique	66(76.7)	NA
	Head on	1(1.2)	NA
	From behind	11(12.8)	NA
	Sideswipe	2(2.3)	NA
	Door	1(1.2)	NA
	NA*	5(5.8)	NA
Area of Impact	Pedestrian crosswalk	25(29.1)	25(22.5)
	Sidewalk	2(2.3)	4(3.6)
	Curb lane - side	18(20.9)	25(22.5)
	Curb lane - middle	18(20.9)	19(17.1)
	Passing lane	4(4.7)	4(3.6)
	Intersection – turning lane	6(7.0)	6(5.4)
	Bicycle lane	2(2.3)	2(1.8)
	Highway On-ramp	1(1.2)	1(0.9)
	Recreational bike path	0(0.0)	4(3.6)
	Construction zone	0(0.0)	1(0.9)
	Parking lot	0(0.0)	1(0.9)
	Railway tracks	0(0.0)	1(0.9)
	NA*	10(11.6)	18(16.2)

* NA indicates the data was not available in the file

Table 3-4: Head impact characteristics and helmet use for the cyclists

	Categories	MV Cases (N = 86) Count (%)	MV + nMV Cases (N = 111) Count (%)
Helmet Use	Y	18(20.9)	22(19.8)
	N	61(70.9)	75(67.6)
	NA*	7(8.2)	14(12.6)
Head Impact	Y	53(61.6)	65(58.6)
	N	17(19.8)	23(20.7)
	NA*	16(18.6)	23(20.7)
Head Orientation (for impacts to the head) (MV: N = 53) (MV+ nMV: N = 65)	Front	6(11.3)	8(12.3)
	Rear	9(17.0)	11(16.9)
	Side	12(22.6)	13(20.0)
	Face	7(13.2)	10(15.4)
	Top	0(0.0)	1(1.5)
	NA*	19(35.8)	22(33.8)

* NA indicates the data was not available in the file

3.4.3 Injury Characteristics

Injury data was available for 71 of the 86 MV cases and 20 of 25 nMV cases. Of the injuries sustained by the cyclists, 20.0% (14 cyclists) were fatal and all occurred in the MV conditions. The frequency of mild injuries according to the Maximum Abbreviated Injury Scale (MAIS) score was 12.9% (MAIS1), moderate injuries 30.0% (MAIS2), serious injuries 17.1% (MAIS3), severe injuries 14.3% (MAIS4) and critical injuries 5.7% (MAIS5) (Figure 3-6).

Some cyclists sustained multiple injuries producing 163 total injuries amongst the 91 injured cyclists. Table 3-5 describes the injury characteristics of the 163 injuries sustained by the cyclists in both the MV and nMV conditions. The frequency of injuries were approximately split between those with an AIS score less than 3 (AIS <3) and those with a serious or greater score (AIS 3+) (Table 3-5). In the nMV condition, the ground was the source for all the injuries. In the MV condition, the ground also was the source of the most injuries (27.8%) with the frame of the vehicle and the body of the vehicle also causing 25.6% and 19.5% of injuries, respectively. The majority of the injuries were to the head and face (41.1%) when combining all the cases (Table 3-5). Injuries to the trunk/pelvis/internal and to the lower extremities were also quite high at 19.6% and 14.7%, respectively (Table 3-5).

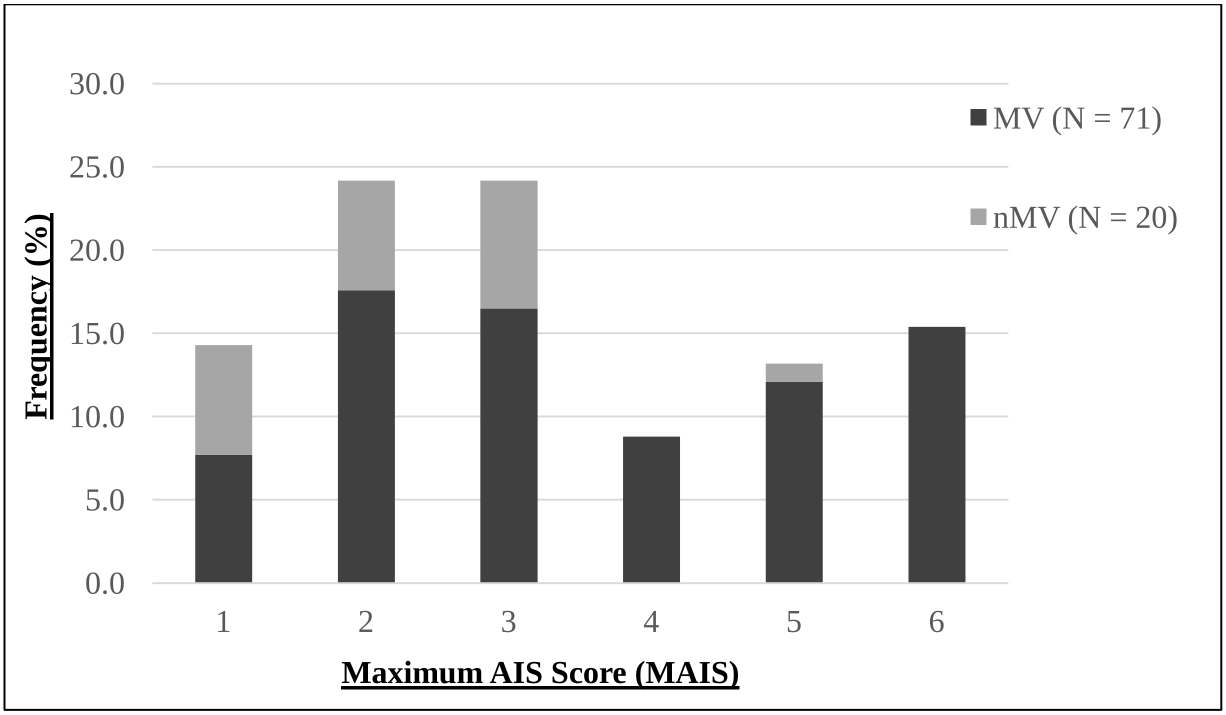


Figure 3-6: Maximum Abbreviated Injury Scale (MAIS) scores for all cyclists with available injury information (N = 91)

3.5 Results – Experiment one - Inferential Analysis

3.5.1 Factors Influencing Injury Risk

Relationships between injury severity and external impact material and vehicle speed were tested for MV conditions using χ^2 tests with resulting *p values* of *.002* and *.035*, respectively (Table 3-8). Injury severity tended to be higher when the cyclist hit stiffer materials with 60.6% of the injuries rated as an AIS 3+ as opposed to the body of the vehicle for which only 30.8% of the injuries were rated AIS 3+ (Table 3-6). Vehicle speed was significantly associated with increased injury severity (*p=.035*, Table 3-8). Residuals were larger for the 0-20 km/h category (-17.5, Table 3-8). Thus, the number of serious (AIS 3+) injuries in this category was much smaller than at higher speeds.

χ^2 tests were completed to assess the relationship between helmet use and injury severity for MV cases alone and nMV/MV cases combined (Table 3-8). Helmet use was not significantly associated with increased injury severity for the combined cases (*p=.149*, Table 3-8). However, it was significantly associated with increased injury severity for MV cases alone (*p=.001,* Table 3-8).

A significant relationship between cyclist age and injury severity existed for MV collisions alone and when MV and nMV collisions were combined (*p=.006* and *p=.003,* respectively, Table 3-8). The frequency of serious injuries decreased as the age of the cyclist increased with 69.0% of cyclists under the age of 16 acquiring an AIS 3+ injury and only 36.8% of cyclists over the age of 54 acquiring an AIS 3+ injury (Table 3-7). Residuals in both the MV and MV/nMV combined groups supported this with decreasing residuals as the age of the cyclist increased (Table 3-8). The distribution of injury severity was not significantly different between genders for either the MV group alone or the combined (MV/nMV) group (*p=.375, p=.571,* respectively, Table 3-8). There was no significant relationship between injury severity and the size of the vehicle (*p=.265,* Table 3-8).

Table 3-5: Injury characteristics of the cyclists

	Categories (collapsed)	MV Cases (count(%)) (N = 133)	MV + nMV Cases (count(%)) (N = 163)	Categories (expanded)	MV Cases (count(%)) (N = 133)	MV + nMV Cases (count(%)) (N = 163)
AIS Score	AIS <3	61(45.9)	83(50.9)	1	14(10.5)	27(16.6)
				2	47(35.3)	56(34.4)
	AIS 3+	72(54.1)	80(49.1)	3	28(21.1)	35(21.5)
				4	13(9.8)	13(8.0)
				5	17(12.8)	18(11.0)
				6 (fatal)	14(10.5)	14(8.6)
Source of Injury	Vehicle: Frame/ Windows	34(25.6)	34(20.9)	A pillar	7(5.3)	7(4.3)
				Windshield	23(17.3)	23(14.1)
				Window	4(3.0)	4(2.5)
	Vehicle: Body	26(19.5)	26(16.0)	Bumper	13(9.8)	13(8.0)
				Fender	4(3.0)	4(2.5)
				Hood	6(4.5)	6(3.7)
				Roof	3(2.3)	3(1.8)
	Ground	37(27.8)	67(41.1)	Ground	37(27.8)	67(41.1)
	Crush	20(15.0)	20(12.3)	Crush	20(15.0)	20(12.3)
	NA*	16(12.0)	16(9.8)	NA*	16(12.0)	16(9.8)
Injury Location	Upper Extremity	10(7.5)	15(9.2)	Upper Extremity	10(7.5)	15(9.2)
	Lower Extremity	21(15.8)	24(14.7)	Lower Extremity	21(15.8)	24(14.7)
	Head/Face	53(39.8)	67(41.1)	Head	47(35.3)	52(51.9)
	Trunk/Pelvis/ Internal	29(21.8)	32(19.6)	Face	6(4.5)	15(9.2)
				Trunk	14(10.5)	17(10.4)
				Pelvis	4(3.0)	4(2.5)
				Internal	11(8.3)	11(6.7)
	Spine/Spinal Cord	6(4.5)	10(6.1)	Spinal Cord	1(0.8)	1(0.6)
				Spine	5(3.8)	9(5.5)
	Fatal	14(10.5)	14(8.6)	Fatal	14(10.5)	14(8.6)
	NA*	0(0.0)	1(0.6)	NA*	0(0.0)	1(0.6)

Table 3-6: Factors influencing injury risk for MV conditions associated with main hypotheses

	Categories (collapsed)	Count (%) AIS <3	Count (%) AIS 3+
External Impact Material (N = 133)**	Stiff: Frame/Ground	28(39.4)	43(60.6)
	Not Stiff: Vehicle Body	18(69.2)	8(30.8)
Vehicle Speed (km/h) (N = 133)**	0-20	13(65.0)	7(35.0)
	21-40	13(38.2)	21(61.8)
	41-60	15(46.9)	17(53.1)
	>60	6(40.0)	9(60.0)
Helmet Use (Y/N) (N = 67)#	Y	9(60.0)	6(40.0)
	N	24(42.9)	28(57.1)
Helmet Use (Y/N) (N = 53)**	Y	8(36.4)	4(63.6)
	N	14(12.9)	27(87.1)

**only MV conditions included
both conditions (MV and nMV) included

Table 3-7: Other factors (cyclist age, gender, and vehicle type) potentially influencing injury risk

	Categories (collapsed)	Count (%) AIS <3	Count (%) AIS 3+
Cyclist Age (years) (N = 163)#	<16	9(31.0)	20(69.0)
	16-34	35(50.0)	35(50.0)
	35-54	23(62.2)	14(37.8)
	55+	12(63.2)	7(36.8)
Cyclist Age (years) (N = 133)**	<16	9(31.0)	20(69.0)
	16-34	25(43.9)	32(56.1)
	35-54	15(57.7)	11(42.3)
	55+	12(70.6)	5(29.4)
Cyclist Gender (N = 163)#	M	65(53.3)	57(46.7)
	F	18(43.9)	23(56.1)
Cyclist Gender (N = 133)**	M	47(47.5)	52(52.5)
	F	14(41.2)	20(58.8)
Vehicle Type (N = 133)**	Large Truck/Bus	8(42.1)	11(57.9)
	Van/Pickup	8(32.0)	17(68.0)
	Minivan/SUV	13(44.8)	16(55.2)
	Sedan/Station Wagon	32(53.3)	28(46.7)

**only MV conditions included
both conditions (MV and nMV) included

Table 3-8: χ^2 results between injury severity and categorical variables describing the collisions

Main Hypotheses Injury severity related to:	**Categories (collapsed)**	**Number of AIS 3+ Scores/Total Number of Injuries per category (%)**	**χ^2 p value**	**Residuals**
External impact material (N = 133)**	Stiff (Frame/Ground)	60.6	.002*	
	Not Stiff (Vehicle Body)	30.8		
Vehicle speed (km/h) (N = 133)**	0-20	35.0	.035*	-17.5
	21-40	61.8		9.5
	41-60	53.1		0.5
	>60	60.0		7.5
Helmet use (N = 67)#	Y	40.0	.149	
	N	53.8		
Helmet use (N = 53)**	Y	33.3	.001*	
	N	65.9		
Secondary Relationships Injury severity related to:				
Cyclist age (years) (N = 163)#	<16	69.0	.003*	20.5
	16-34	50.0		1.5
	35-54	37.8		-10.5
	55+	36.8		-11.5
Cyclist age (years) (N = 133)**	<16	69.0	.006*	18.0
	16-34	56.1		5.0
	35-54	42.3		-9.0
	55+	36.8		-14.0
Cyclist Gender (N = 163)#	M	46.7	.375	
	F	56.1		
Cyclist Gender (N = 133)**	M	52.5	.571	
	F	58.8		
Vehicle Type (N = 130)&**	Large Truck/Bus	57.9	.265	
	Van/Pickup	68.0		
	Minivan/SUV	55.2		
	Sedan/Station Wagon	46.7		

*indicates significance for $\alpha<.05$
**only MV conditions included
both conditions (MV and nMV) included
& sample size reduced because a unique case was removed for the statistical test

3.6 Results - Experiment two - determining 'MV Velocity'

Model inputs and outputs from the 30 simulations can be found in Appendix B: PC Crash model inputs and outputs. Vehicle speed at impact ranged from 15-80 km/h and cyclist speed at impact ranged from 3-30 km/h. Nineteen out of 30 vehicle/cyclist collisions were oriented perpendicularly at impact and the majority of vehicles involved were sedans. All cases included a head impact. The majority of impacts were to the side of the head and impacts occurred often to the stiffer areas of the vehicle (window/windshield/A pillar or the ground, Appendix B: PC Crash model inputs and outputs). Only 10 of 30 cyclists were wearing a helmet during the impact (Figure 3-7). Pearson product-moment correlations indicated that when all cases were included (helmeted and unhelmeted), AIS scores were weakly, positively correlated with head impact velocity (r=.347, n=30, *p=.030*). When helmeted impacts were removed, the correlation between head AIS score and head impact velocity was slightly stronger (r=.440, n=20, *p=.026*).

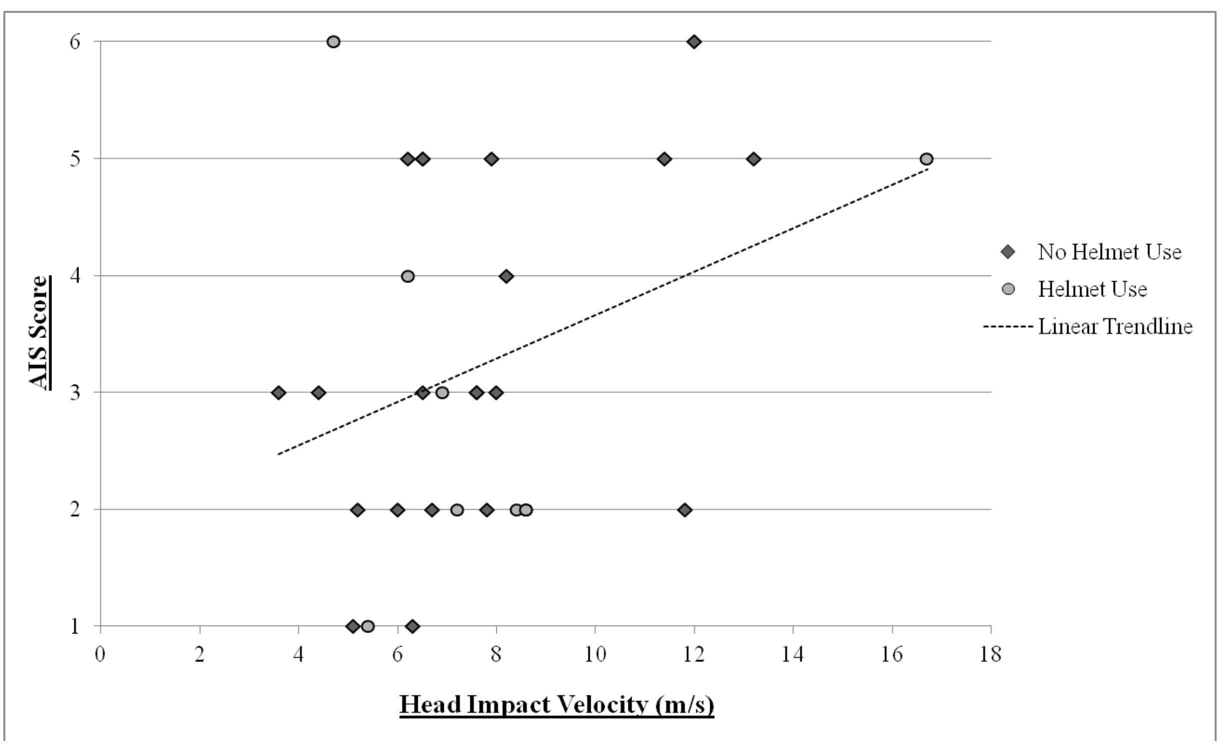


Figure 3-7: Head impact velocity vs. head AIS score for the 30 simulated cases

3.7 Discussion

Current reports of cyclist/motor vehicle collisions can be incomplete (Elvik & Mysen, 1999). Furthermore, epidemiological literature summarizing cycling collisions in Southern Ontario is limited. To address these current limitations in the literature, the current study analyzed cyclist/motor vehicle collisions resulting in injury claims in Southern Ontario using data from a professional forensic engineering company. These companies typically have access to police and hospital reports as well as other useful information provided by clients (e.g. continuing medical reports, witness statements, etc.), giving them the opportunity to summarize more details about cyclist, collision and injury characteristics than police services or hospital records can provide. The GK database contained 86 cyclist/motor vehicle collisions and 25 cyclist incidents without motor vehicle involvement that caused a total of 163 injuries to the cyclists.

The first objective of experiment one in the current study was to describe the crash characteristics associated with cycling collisions that caused injury claims. Most of the collisions occurred during the months of August and September. Over the course of a day, the majority of collisions occurred in the evening between the hours of 4-7 p.m. These results are similar to those of Isaksson-Hellman et al. (2012), who analyzed data from Swedish insurance companies, and the Toronto Police Services (City of Toronto, 2011). Haileysus et al., (2007) also recorded a larger percentage of collisions in the summer months from May-September in the United States. Since Canada has cold, snowy winter months, it is very likely the population chooses bicycle riding as a mode of transportation or recreational activity during the warmer summer months. The time of day coincides with when people may be cycling home from work or when traffic density is the highest (Isaksson-Hellman, 2012). Interestingly, a spike was not seen in this data during the 7-8 a.m. commute period as was seen in Isaksson-Hellman et al. (2012) and the

Toronto Police Cycling Collision Summary (2011). This may be due to the smaller sample size of the current study or the type of data (i.e. cyclists in the current dataset may not have been commuters).

The majority of cyclists in this data set did not wear a helmet (2/3 of the cyclists, Table 3-4) which aligns with a previous epidemiological study investigating cyclist head injuries (Linn et al, 1998). Furthermore, 55 of 86 cyclists experienced a head impact during one or more parts of the collision. Of these 55 cyclists, the orientation of the head upon impact was fairly distributed across the front, back, side, and face (Table 3-4). This aligns with findings from Depreitere et al. (2004) who characterized head injuries to cyclists and also found most head injuries to cyclists involved in a motor vehicle collision were located mostly at the forehead or on the side of the head, with occipital impacts (back of the head) closely behind.

Injuries to the head and face represented the majority of injuries to cyclists in this data set (41.1%, Table 3-5) which is higher than studies reporting statistics from the United States, France and Germany (head injury frequency ranged from 22.0-31.7%; (Amoros et al., 2011; Haileyesus et al., 2007; Lustenberger et al., 2010; Rodarius, Mordaka, & Versmissen, 2008). However, the sample sizes in previous studies are much larger (N = 2500-12000 compared to N = 163). Furthermore, datasets from these studies are typically taken from hospital databases whereas the current dataset is from a firm representing cyclists involved in litigation indicating that perhaps head injuries may contribute to individuals starting the legal process more often than injuries to other areas.

Injuries to the pelvis and lower extremities were the second most common types of injuries to cyclists in the current data set (19.6% and 14.7%, respectively, Table 3-5). Again, these results aligned with other studies with lower extremities representing 13.0-27.7% of injuries (Amoros et al., 2011; Haileyesus et al., 2007; Lustenberger et al., 2010; Rodarius et al., 2008). The higher incidence of these injuries seen in MV collisions is typically because the lower extremity is often the first contact site for many of the cyclists (Simms & Wood, 2009). The majority of cyclists and motor vehicles in the current dataset were oriented obliquely (76.7%, Table 3-3), further supporting the likelihood of the lower extremity as the first point of contact on the cyclist. Furthermore, the high incidence of oblique impacts and collisions occurring on pedestrian crosswalks may indicate that the cyclists in the current dataset may have been more representative of a recreational cycling population as opposed to road cyclists who would likely be oriented in the same direction of traffic during a collision with a motor vehicle (likely hit from behind).

The secondary objective of experiment one in the current study was to determine whether relationships existed between injury circumstances and resulting injury outcomes in the collected dataset. The results supported hypothesis one and found a significant relationship between injury severity and external impact material (Table 3-8). Injury severity tended to increase (to injuries rated AIS 3+) when the cyclist hit stiffer materials such as the A-pillar or frame of the windshield/window as opposed to hitting materials that are less stiff such as the bumper or hood of the vehicle. This result aligns with previous findings which found that impacts to the windshield caused more severe injuries when compared with other regions of the vehicle such as the hood and when only looking at the windshield, impacts to areas closest to the frame seemed to increase injury severity (Rodarius et al., 2008).

The Enhanced European Vehicle Safety Committee (EEVC) and the International Harmonization Research Activity (IHRA) have proposed impact tests to improve vehicle front-end design as a provision for pedestrian safety. In response to this, vehicle manufacturers have started designing front-ends with external airbags that deploy to lift the hood of the vehicle away from the stiffer engine block or create a u-shaped airbag around the bottom and sides of the windshield (Millward, 2009). Impact tests for vehicle design have been designed based on studies of pedestrian safety and generally hoped that they will also improve the safety of cyclists (Simms & Wood, 2009).However, due to pedestrians lower centre of gravity, further research likely needs to be done into vehicle front-end design with respect to vulnerable road users with a higher centre of gravity such as cyclists. The current results indicate that future vehicle design should continue to focus and build on protecting the cyclists from the windshield and surrounding areas (the stiffer areas).

The second hypothesis was also supported but only for the MV conditions (Table 3-8). Specifically, injury risk increased when a helmet was not worn (Table 3-8). A Cochrane review of bicycle helmet efficacy strongly supports the use of helmets when cycling indicating that helmets reduce the effects of head injury by 85%, brain injury by up to 88%, severe brain injury by 75%, and even facial injuries of the upper to mid facial regions by up to 65% (Thompson, Rivara, & Thompson, 1999). Some cycling advocates have argued that the use of helmets may change the riding behavior of cyclists so that they feel more comfortable putting themselves in riskier situations (Hillman, 1993). However, there are no objective studies to support these claims. The current dataset supports the use of helmets in scenarios that may produce higher energy impacts such as collisions with motor vehicles. Although not statistically significant, when nMV and MV collisions were combined, the trend was still the same with helmet users sustaining less severe injuries compared with non-helmet users (Table 3-8). Helmet users represented a small group in the total (nMV and MV combined) sample (N=15). An increased sample size with more cases of helmet compliance would be beneficial when comparing these results to previous findings in support of helmet efficacy. It may also be possible that helmet users do not acquire severe injuries in general and thus are not as widely represented in cases involving injury claims such as those used in the current dataset.

Increased injury severity was significantly associated with vehicle speed (*p=.035,* hypothesis three, Table 3-8) supporting correlations found in previous studies (Simms & Wood, 2009). The trend between the category of vehicle speed and injury severity increased positively with a higher ratio of serious (AIS 3+) injuries seen when the vehicles were travelling over 20 km/h compared to when they were travelling less than 20 km/h (Table 3-6). An insignificant relationship was found between injury severity and vehicle type (Table 3-8) indicating that perhaps the role of the vehicle shape may be less important to the final injury outcome compared to the speed of the vehicle or the surface of the impact. The height of the cyclist would likely influence the relationship between vehicle shape (front-end height) and injury location (and severity) for the cyclist. Unfortunately, the sample size of the current dataset was too small for any inferential statistical tests between these factors.

An interesting relationship was found in the current dataset between cyclist age and injury severity for both MV cases alone and combined (nMV and MV) cases (*p=.006, p=.003,* respectively; Table 3-8). Trends in injury severity decreased as the age of the cyclists increased (residuals were largely positive for cyclists in age group <16 and largely negative for the age group 55+). Other epidemiological studies have found opposite trends with older cyclists sustaining more severe injuries than younger cyclists (Kloss, Tuli, Haechl, & Gassner, 2006; Malczyk, Bauer, Juhra, Schick, & Münster, 2014). A contribution to this difference may be due to the types of injuries seen in the age group <16 years in the current dataset. Almost half (15 of 39) of the children included in the current dataset were not wearing a helmet at the time of the collision and sustained a significant head injury (AIS 3+). Injury claims are likely in these circumstances because of the significant influence of severe head injuries on quality of life. Furthermore, younger cyclists are likely to be shorter and have a lower centre of gravity than older cyclists. Thus, depending on the vehicle shape, the trajectory pattern of younger cyclists will change compared to adult cyclists. This may explain the increased injury severity seen in younger cyclists of the current dataset. Younger cyclists would likely be impacted at the mid-body region and experience a forward projection or may even sustain crush injuries rather than wrapping around the vehicle front-end like taller cyclist and impacting the top of the vehicle (Simms & Wood, 2009). To support this explanation in the current dataset, the injuries of the younger cyclists were examined. Of the injuries to the younger cyclists, only 13.8% occurred at the lower extremity whereas the remaining injuries were to the trunk, spinal cord, internal structures, head and face. These findings tend to support differences in the mechanisms of injury to younger cyclists with the trunk or head likely being a main area of impact. To further investigate these theories, future epidemiological studies should focus on detailed impact sites and injury locations and compare these with cyclist ages and/or heights and vehicle size and shape.

The goal of experiment two of the current study was to determine an impact velocity that was more realistic of cyclist/motor vehicle collisions in Southern Ontario ('MV velocity'). The velocity obtained from modeling 30 collisions, of 7.7 m/s, was on the lower end of ranges seen in previous studies using mathematical models to simulate similar collisions (Carter & Neal-Sturgess, 2009; Fanta et al., 2013; Ito et al., 2014). However, these previous studies chose to model one specific, rather severe collision (Carter & Neal-Sturgess, 2009; Ito et al., 2014) or categories of collisions based on validation studies with post mortem human subject data (Fanta et al., 2013). The current study modeled 30 collisions with each of their independent characteristics based on their respective case files from a professional forensic engineering company. Injury outcomes ranged from minor (AIS 1) injuries to critical and fatal injuries (AIS 5/6). Therefore, it is possible that the differences in head impact velocities occurred because the currently used collisions spanned a larger range of collision severities compared to the three previous studies.

The correlation between head impact velocity and head AIS score was weakly, positively correlated when unhelmeted and helmeted impacts were combined (Figure 3-7). In a similar attempt to correlate the severity of head injuries (e.g. AIS score) following motorcycle collisions, it was found that the majority of low velocity injuries were associated with minor head injury while higher speed injuries seemed to cause more critical or fatal head injuries (Chinn, Doyle, Otte, & Schuller, 1999). However, a weak, positive correlation was also found by these authors as even at higher speeds, head impacts weren't always associated with more severe injuries (Figure 3-8). Chinn et al. (1999) only observed cases involving a motorcyclist wearing a helmet; thus, outcomes may have been critically influenced by individual helmet properties, such as geometry, mass, shape, and materials. The weak, positive correlation in the current study aligned with the findings from Chinn et al. (1999).

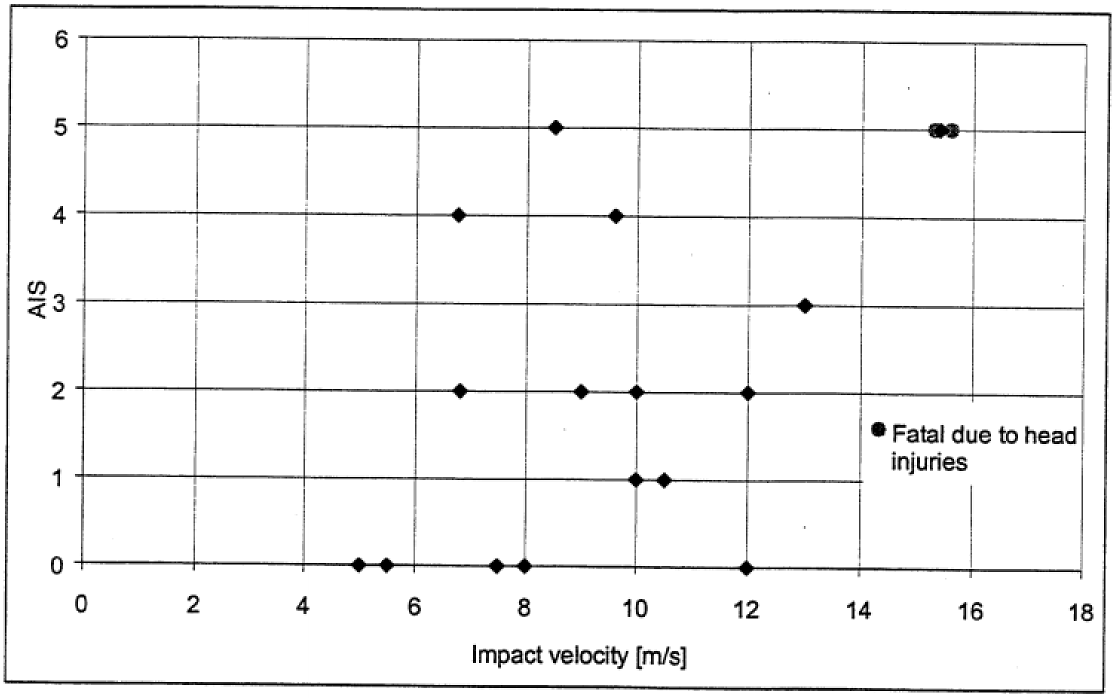


Figure 3-8: Impact velocity vs. head AIS score for motorcycle collisions (from Chinn et al., 1999)

The current study had several limitations. First, while the dataset in the current study begins to provide a characterization of cyclist collisions in Southern Ontario, it is not clear how the sample relates to the general population. Generally, cases are litigated if the injuries severely affect an individual's quality of life and further financial support is warranted beyond what is covered by insurance companies. Thus, the current dataset may represent cyclist collisions that are more severe than the population of cycling collisions with motor vehicles. Secondly, the dataset is quite small (N=111). To address these limitations, data from other forensic companies could be collected and compared to collisions described from other sources. Insurance companies may also be able to provide a useful dataset for these types of collisions (Isaksson-Hellman, 2012). Increasing the size of the dataset would also allow for the development of more advanced predictive models between injury circumstances and injury outcomes. Thirdly, modeling realistic collisions using information from reports has inherent limitations. Only a small sample of collisions was modeled due to gaps in reported information to the forensic company. Therefore, while correlations were attempted between head impact velocity and head AIS score, not enough cases were involved to add in other factors to validate the model outputs (e.g. impact surface, head orientation). In the future, companies with access to such information at the time of the crash should be encouraged to document and perhaps share relevant information for modeling purposes. This would increase the sample sizes of realistic cyclist/motor vehicle collision models and allow for more pragmatic analyses of current injury tools/criteria and realistic injury outcomes from such scenarios. Finally, PC Crash is a simplified modeling tool for multi-body cases such as cyclist/motor vehicle collisions. Segments (bodies) are designed as ellipsoids and while segments are joined with ball and socket joints. Although the joint stiffness's can be restricted by the user, adding stiffness coefficients significantly increases the complexity of the mathematical model and for lengthier impacts, the multi-body model breaks down (Moser, Steffan, & Kasanický, 1999). For the purposes of the current study, PC Crash was used for simpler cases with short impact periods to begin to estimate the velocities seen in cyclist/motor vehicle collisions so that the third part of the study could be completed comparing such velocities to those currently used in testing standards. However, in the future the use of a more robust tool such as MADYMO may provide a more accurate and detailed picture of these types of collisions. The mean value of the 30 reconstructed cases was used as an input into study two, experiment two (Chapter 4). A median or peak impact velocity may have been equally as relevant to the goals of relating the impact velocities of the collision reconstructions to experiment three. Despite the limitations of PC Crash and the restrictions of the current models, the outputs of the reconstructions in this dataset provided a useful range of head impact velocities that may be used as inputs into experimental studies investigating cycling collisions and head impacts.

Cyclist collisions have yet to be characterized for the province of Ontario. Furthermore, datasets from professional forensic engineering companies have not previously been used to summarize collision characteristics and injury information in cycling epidemiological literature. These companies offer a unique opportunity to analyze collision characteristics in detail and provide links to injury outcomes. Thus, the current dataset is novel and provides details about risk factors of cycling collisions whereas previous epidemiological studies may have been missing important elements (Rosman & Knuiman, 1994). This study is the first to characterize cycling collisions in Southern Ontario and is in general agreement with common characteristics reported in previous epidemiological literature. The collected information will be useful for designing road infrastructure in this region. Furthermore, it will contribute to the development of injury prevention methods for vehicle and cycling equipment manufacturers, and basic knowledge of collision characteristics for biomechanists designing experiments in these areas.

Chapter 4: Headform Responses to Helmet Testing Standards

4.1 Background

Head injuries are among the most severe injuries following a cyclist/motor vehicle collision accounting for approximately two thirds of hospital admissions for cyclists (Amoros et al., 2011; Thompson & Rivara, 2001). Helmets were introduced for cyclists in the late 1900's to mitigate head injuries following falls and collisions (Swart, 2003). Current helmet designs consist of a foam liner, typically expanded polypropylene (EPS), surrounded by a hard outer shell made from polycarbonate or another type of plastic. During an impact, the foam below the area of contact yields in response to an impact and the shell deforms around the areas of uncrushed foam to absorb energy (Mills, 1990). The efficacy of helmets has been debated in research. However, currently there are no objective studies against helmet use while cycling and reductions in head injury frequency have previously been reported in literature in the ranges of 37-88% (McDermott et al., 1993; Persaud et al., 2012; Thompson et al., 1996). Furthermore, study one investigated the link between head injury severity and helmet use in a sample of cyclists involved in injury claims in Southern Ontario. This study also supported the use of helmets with decreased head injury severity significantly associated with helmet use for cyclists involved in collisions with motor vehicles (Section 3.5). Although these studies report a decrease in head injury risk with helmet use, head injuries continue to occur indicating the need to look further into the design and mitigating capacity of helmets.

In order to properly certify helmets and (presumably) increase market share, manufacturers must ensure their bicycle helmets comply with current standards. In Canada, the Canadian Standards Association holds the current national standard (CSA D113.2_1989;

(Canadian Standards Association, 2009). However, this standard also states that helmets sold on the Canadian market can be certified through alternative test methods including the American (Consumer Product and Safety Commission (CPSC) and American Society of Testing Materials (ASTM)) or European standards (British Standards Institute (BSI). All helmet standards have essentially the same approach to testing the impact performance of bicycle helmets. An artificial headform (usually the ISO Magnesium K1A) is dropped while wearing the helmet to achieve an impact at a certain energy level chosen to represent the specific application of the helmet. Helmets are typically tested in four conditioning environments (e.g. normal, cold, hot, wet) on two to three impact surfaces. Finally, an accelerometer is mounted at the centre of mass of the headform to monitor acceleration and predict injury outcome (Nahum & Melvin, 2002).

Several limitations exist with the current standards for bicycle helmets. First, the current surrogate headforms are made of a magnesium alloy and are only shaped like the top half of the human head. Therefore, they are not completely representative of the geometry of a realistic human head. Second, the current standards use drop heights between 1.5 and 2.2 m (American Society for Testing and Materials, 2012; Canadian Standards Association, 2009; Consumer Product Safety Commission, 1998; EN, 1997; SNELL, 1995). These drop heights typically represent a fall from a bicycle and do not necessarily simulate higher energy scenarios such as cyclist/motor vehicle collisions which were classified in study one for the region of Southern Ontario (Walker, 2005). Third, current standards only use the Gadd Severity Index (GSI) (peak acceleration) as an outcome variable. The GSI was based on the Wayne State Tolerance Curve (WSTC) which was originally developed to relate linear acceleration to skull fracture and impact duration (Schmitt et al., 2014; Simms & Wood, 2009). A direct link has not been established between skull fracture and brain injury (Schmitt et al., 2014). The GSI has also been criticized for predicting high scores for long duration, low intensity impacts which may not be realistic during cycling collisions (Shorten & Himmelsbach, 2003). Finally, current standards typically use a monorail drop tower attached to the surrogate metal headform with a rigid ball-arm. This set up improves the repeatability and reliability of the results, but may not represent realistic injury scenarios during cycling collisions. This project addresses the limitations of surrogate headform selection and drop heights of the current helmet testing standards.

Several biofidelic headforms currently exist in other types of head impact testing. The Hybrid III headform (HIII) was developed as part of a General Motors project to build a crash test dummy for simulating humans in car collisions (Kendall et al., 2012). It was designed to withstand high impact scenarios without breaking but still replicate some of the data found in cadaveric head impact studies (Kendall et al., 2012). The second headform was produced by the National Operating Committee on Standards for Athletic Equipment (NOCSAE). It is constructed with a high durometer, urethane skull covered by a lower durometer urethane that forms anatomical features such as the ears and lips. This headform also has a glycerin-filled brain cavity to optimally simulate the behavior of the brain during impact (National Operating Committee on Standards for Athletic Equipment, 2009). It is considered to have more human-like characteristics based on its gel-filled brain cavity and human anthropometry (Kendall et al., 2012). Both biofidelic headforms are not currently used in cycling helmet testing. However, presuming they produce reliable data, they may be better representations of a real human head and thus provide more realistic impact scenarios and accurate predictions of the mitigating capacity of bicycle helmets.

Cycling helmets were initially designed to mitigate injuries following a collision where a cyclist falls to the ground without any vehicle involvement (Walker, 2005). The current standard drop heights of 1.5-2.2 m were chosen to replicate this scenario and create a final impact velocity prior to head impact of around 5.7 m/s. Study one modeled 30 of the cyclist/motor vehicle collisions retrieved from the database at Giffin Koerth Forensic Engineering that involved a head impact to estimate head impact velocity during higher energy impacts (Chapter 3). The average of the 30 cases was 7.7 m/s which aligns with ranges seen in previous literature modeling hypothetical collisions (Fanta et al., 2013; Ito et al., 2014).

4.2 Purpose and Hypotheses

There were two main purposes of the second study of my thesis. The first was to compare headform responses between three surrogate headforms for three impact locations, with and without the use of a helmet and assess the reliability of the different surrogate headforms to produce similar results across trials. This part of the study was completed at an impact velocity of 1 m/s for unhelmeted conditions and at the CSA standard impact velocity of 5.7 m/s for helmeted conditions. The lower impact velocity was chosen for unhelmeted conditions to protect the mechanical integrity of the headforms from high impact energies. The second purpose of this study was to compare the mitigating capacity of three models of cycling helmets (at three price points) to proposed injury assessment reference values (IARV) using: i) the impact velocity used in the CSA standards (hereafter referred to as “standard velocity”) and ii) an impact velocity more realistic of cyclist/motor vehicle collisions seen in Southern Ontario (referred to as “MV velocity”). The MV velocity was determined in study one (Chapter 3) by simulating a subset of the impacts using the software PC Crash (Version 9, Datentechnik Group, Linz, Austria).

The specific hypotheses for this study were split into two sections.

1) *The* ***biofidelic headforms*** *(NH and HIII) will have* ***similar responses*** *to impacts and will have* ***different responses*** *than the* ***magnesium K1A headform*** *for all impact orientations and for both unhelmeted and helmeted trials. Specifically:*
    a. K1A headform will have **higher peak accelerations, peak forces, and HIC scores** than the HIII and NH headforms. Previously, a comparison between the HIII and K1A headforms for some impact conditions involving only the front of the head found higher peak accelerations for the K1A headform by 5% (Stuart et al., 2013).
    b. Impact orientation will **not affect** the differences in headform responses from hypothesis 1a.
    c. Helmet use will **not affect** the differences in headform responses from hypothesis 1a.
    d. All three headforms (K1A, HIII, NH) will produce **repeatable data (within ±5%)** for both force and acceleration responses. Stuart, et al. (2013) found the HIII headform to have repeatable impacts within ±5%. However, these impacts were only for the front of the head.
2) *The helmets* ***will adequately mitigate injury (below an injury threshold)*** *for the impacts completed at the* ***standard velocity*** *but* ***not*** *for impacts at the* ***'MV velocity'*** *for all impact orientations and this will be true for all brands of helmets.*
    a. Outcome variables (*Gmax, HICmax*) will **fall below injury assessment risk value (IARV) levels** for impacts at the **standard velocity**. The helmets chosen for this study passed the CSA standards which is an impact velocity of 5.7 m/s. Therefore, it's expected that the helmets will adequately mitigate injury at the standard velocity (Canadian Standards Association, 2009).

b. Outcome variables (*Gmax, HICmax*) will **exceed IARV levels** for impacts at the '**MV velocity**'. This is based on data which suggests that impacts to the front of the head exceeded a 50% probability of head injury according to a curve produced by Mertz, Irwin & Prasad (2003) (Stuart et al., 2013). Again, this study only looked at front of the head impacts.
c. Hypotheses 2a and 2b will be **true for all impact orientations**.

4.3 Methods

4.3.1 Study design

This study was split into two experiments. The experimental protocol for all impacts in the study was adapted from the methods set out in the CSA Standards D113.2 (Canadian Standards Association, 2009). A monorail mechanical drop tower (Dixon & Brodie, 1993) was used to drop each headform onto a flat, steel anvil (Figure 4-3) (Canadian Standards Association, 2009). The headforms were attached with the desired impact orientation (front, back, or side) to the mechanical drop tower and raised to the required drop height where they were held by an electromagnet (custom model, AEC Magnetics, Cincinnati, OH, USA) until released. All impacts were recorded using a high speed video camera (High Speed Imaging Inc., Mississauga, ON, Canada) to measure impact velocity during the last 40 mm of free fall (Canadian Standards Association, 2009). A triaxial accelerometer (Model 2707A, frequency range: 0 – 2000 Hz; Endevco Corporation, San Juan Capistrano, CA, USA) was mounted at the centre of mass of each head form to record impact accelerations while a uniaxial load cell (Model 925M113, Kistler Instrument Corporation, Amherst, NY, USA) was mounted below the impact surface to record impact forces (Figure 4-3). Force and acceleration data were sampled at 20 kHz (American Society for Testing and Materials, 2004).

4.3.1.1 Experiment One – Comparing Dynamic Headform Responses

Experiment one compared dynamic responses of three surrogate headforms currently used in equipment and safety testing to helmeted and unhelmeted impacts. The three headforms tested included the K1A Magnesium headform (hereafter referred to as “K1A”, ISO size J half-headform, Cadex Inc., QC, Canada) which is used in the majority of cycling helmet testing standards (American Society for Testing and Materials, 2012; Canadian Standards Association, 2009; Consumer Product Safety Commission, 1998), the Hybrid III headform from the anthropomorphic test device (ATD) (hereafter referred to as “HIII”, Humanetics ATD Manufacturing Inc., OH, USA) and the NOCSAE Hodgson-WSU headform (hereafter referred to as “NH”, NOCSAE, KS, USA). Details of the headform specifications and their comparison to cadaver head specifications are outlined in Table 4-1.

To limit the potential of headform damage (most specifically NH and K1A), the unhelmeted impact velocity for impacts in this experiment was 1 m/s. The three headforms were impacted three times at three different orientations (front, back, and side of the head) (Figure 4-1). The headform orientations chosen represent the most commonly injured areas on the head for cyclists (Depreitere et al., 2004) and common impact sites for other helmet testing found in previous studies (Kendall et al., 2012). All drops within a headform were completed consecutively to limit error introduced from changing headforms and removing the accelerometer. However, headform order and orientation were randomized in accordance with the CSA standards (Canadian Standards Association, 2009).

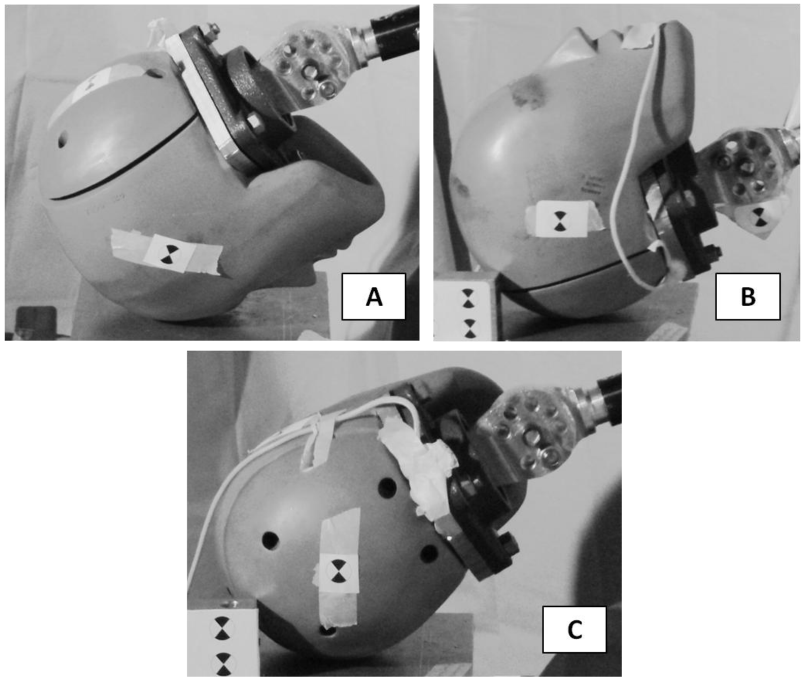


Figure 4-1: Headform orientations: A - Front, B - Back, C - Side

The three headforms (K1A, HIII, and NH) were then impacted using the same protocol at an impact velocity from the CSA standard (5.7 m/s) while wearing three brands of cycling helmets purchased from Mountain Equipment Co-op (MEC) (Figure 4-2) (Mountain equipment coop - MEC.2014). The three helmets represented best sellers in their respective price segments according to sales information provided by MEC (Vancouver, BC, Canada). Specific helmet brand names, material, and sales information can be found in (Table 4-2)

. Helmets were all certified to the European standard which has a lower drop height than the CSA standard but acts as an alternative to which helmets may be certified to on the Canadian market. Helmets were fitted onto the surrogate headforms according to the manufacturer's recommendations. Straps were tensioned at the same amount (175 N) for each helmet and measured using an in-series strain gauge. Each helmet experienced one impact at each impact location (for a total of 3 impacts per helmet). Again, all impacts for each headform were completed consecutively and orders of helmet and impact orientation were randomized. In total this experiment impacted the headforms 54 times (27 unhelmeted impacts at 1 m/s and 27 helmeted impacts at 5.7 m/s) and used 9 of each brand of helmet (27 helmets total). Impacts were completed a minimum of three minutes apart (Canadian Standards Association, 2009).

Table 4-1: Headform specifications (Cadex, 2013; Kendall, et al., 2012)

Headform	Weight (kg)	Circum. (m)	Materials		
			Shell	Cover	Interior
K1A (size J)	3.11	5.70×10^{-1}	NA	NA	Magnesium K1A Alloy
NHWSU	4.85	5.78×10^{-1}	Nylon	Urethane	Glycerin
HIII	4.54 ± 0.01	5.72×10^{-1}	Aluminum	Vinyl	None
Cadaver	5.60	5.72×10^{-1}	Cortical Bone	Scalp	Brain

Figure 4-2: a) MET Xilo Helmet b) MET Crossover Helmet c) Kask Mojito Cycling Helmet (Mountain equipment coop - MEC.2014)

<u>Table 4-2: Helmet specifications</u>

	MET Xilo	MET Crossover	Kask Mojito
Price (CAD)	$34	$59	$209
Weight (g)	280	290	220
Materials – Inner Lining	EPS (expanded polystyrene)	EPS (expanded polystyrene)	EPS (expanded polystyrene)
Materials – Outer Shell	Polycarbonate	Polycarbonate	Polycarbonate
Intended Use	Cycling (commuting, touring, road)	Cycling (commuting, touring)	Cycling (road)
Construction	Molded in the shell	Molded in the shell	Molded in the shell
Fit System	Adjustable dial and interchangeable foam pads	Adjustable dial	Adjustable dial
Approved Standard	EN 1078	EN 1078	EN 1078

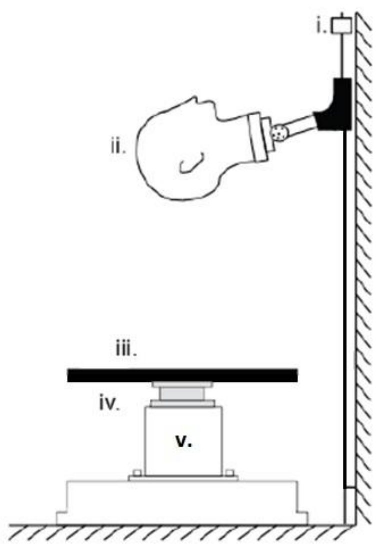


Figure 4-3: Schematic of the mechanical drop tower (adapted from Wright (2011)). Hardware elements include: i. electromagnetic release, ii. surrogate headform, iii. flat anvil, iv. load cell, v. concrete base.

4.3.1.2 Experiment two: mitigating capacity of cycling helmets based on "standard" and "MV" velocities

In the second experiment of the study, the mitigating capacity of the three helmets purchased from MEC (Table 4-2, Figure 4-2) were evaluated for two impact velocities at three impact orientations using the Hybrid III headform (HIII). The second impact condition employed a higher impact velocity than has been previously used in testing standards. Therefore, the HIII headform was chosen to preserve the mechanical integrity of the test system as it was designed for higher energy impacts. The first velocity ("standard velocity") was chosen based off the drop height used in the CSA standards (5.7 m/s ± 3%). The data for the "standard velocity" trials were collected as part of experiment one. The second velocity (motor vehicle or "MV velocity") was selected as the $Head_{vel}$ calculated in experiment two (7.7 m/s), which represents a velocity more representative of the cyclist/motor vehicle collisions that we observed in the database characterized in study one (Chapter 3). The HIII headform was used to impact three helmets at three impact orientations. Each helmet was impacted once at each orientation corresponding to three impacts for each helmet. Dependent variables *Gmax* and *HICmax* were compared with IARVs from the skull fracture and head injury risk curves developed by Mertz, et al. (2003).

4.3.2 Data analysis

Accelerometer data was processed according to the SAE J211/1 guidelines (SAE, 2003) using a custom program written in MatLab (vR2007b, Mathworks, Natick, MA, USA). A 2nd-order, dual-pass, low-pass Butterworth filter (1650 cutoff) was used to filter the acceleration data in the three orthogonal axes before the resultant acceleration was calculated using Equation 4.1.

$$a_r(t) = \sqrt{a_x(t)^2 + a_y(t)^2 + a_z(t)^2} \qquad \textbf{(4.1)}$$

Gmax was recorded as the single largest value from the resultant acceleration-time curve. The *HIC* score was also calculated from the acceleration data using Equation 4.2. Force data was not filtered to preserve peak values and the data appeared to be clear when plotting the raw data vs. time (Figure 4-4).

$$HIC_{15} = max[t_2 - t_1]\left(\left[\frac{1}{[t_2-t_1]}\right]\int_{t_1}^{t_2}[a]dt\right)^{2.5} \; where \; (t_2 - t_1) \leq 15\,ms \qquad \textbf{(4.2)}$$

4.3.3 Statistical analysis

The following tests were completed to address hypotheses 1a-1c. A two-way ANOVA was used to assess the influence of headform and impact orientation on *Fmax, Gmax,* and *HICmax* during unhelmeted impacts. When significant main effects were observed, least significant difference (LSD) post-hoc tests were conducted to compare between headforms.

Helmeted impacts were assessed using a three-way ANOVA with factors including headform, impact orientation, and helmet type. Adjusted R^2 scores were analyzed to determine the influence of each factor on the outcome variables. Again, least significant difference (LSD) post-hoc tests were performed when significant main effects occurred to compare outcome variables between headforms.

The repeatability of the headform responses (hypothesis 1d) was assessed two ways. The first used the average differences of the dependent variables (*Fmax, Gmax,* and *HICmax*) from their respective means within ±5%. The second used an intra-class correlation coefficient (ICC) (two-way mixed for absolute agreement, average measures) for the three dependent variables (*Fmax, Gmax,* and *HICmax*). An ICC > 0.8 represents an excellent reliability between the outcome values.

Hypothesis two was assessed using a descriptive analysis. Specifically, the dependent variables (*Gmax and HICmax*) were compared to IARVs for peak acceleration and HIC score for both the standard velocity and the MV velocity.

All analyses were performed using statistical analysis software (SPSS Version 19.0, SPSS Inc., Chicago, IL) with an alpha of 0.05.

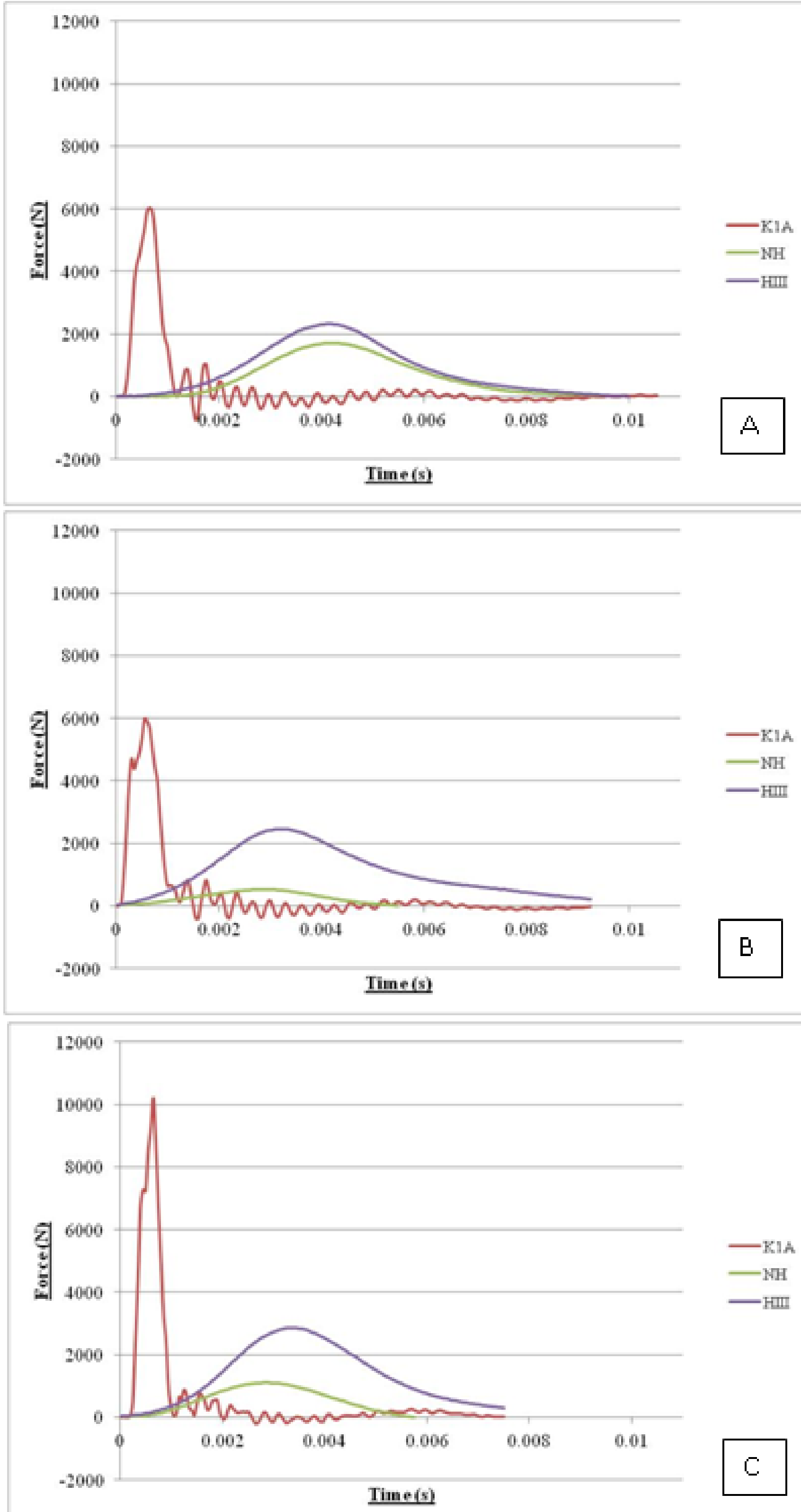


Figure 4-4: Sample force traces for all three headforms at three impact orientations A) Front B) Back and C) Side

4.4 Results

4.4.1 Headform differences during unhelmeted trials

A two-way ANOVA indicated a significant interaction between headform and impact orientation for *Fmax* (*p<.001*, Table 4-3) but no significant interactions for the variables *Gmax* and *HICmax* (Table 4-3). Main effects of headform were significant for *Gmax* and *HICmax* (*p<.001*, Table 4-3) and one-way ANOVAs indicated significant main effects for *Fmax* at all impact orientations ($F_{2,6}$=31.3, 92.5, and 985.9, *p<.001-.001*, for the front, back, and side of the headform). Least significant difference (LSD) post-hoc tests revealed that the magnesium headform (K1A) had significantly higher outcome variables than both the NOCSAE (NH) and Hybrid III (HIII) headforms (by 2.1–15.6 times, Figure 4-5). The NH and HIII headforms were not significantly different from each other (*p>.05*).

Table 4-3: ANOVA results of headform and impact orientation with no helmet use for *Fmax, Gmax, and HICmax*

	Headform		**Impact Orientation**		**Headform*Impact Orientation**	
	F(df)	**p**	**F(df)**	**P**	**F(df)**	**p**
Fmax	266.6(2)	<.001*	14.4(2)	<.001*	9.8(4)	<.001*
Gmax	283.3(2)	<.001*	1.7(2)	.213	1.3(4)	.313
HICmax	165.8(2)	<.001*	1.5(2)	.245	1.3(4)	.291

***** indicates significance at an alpha level of 0.05

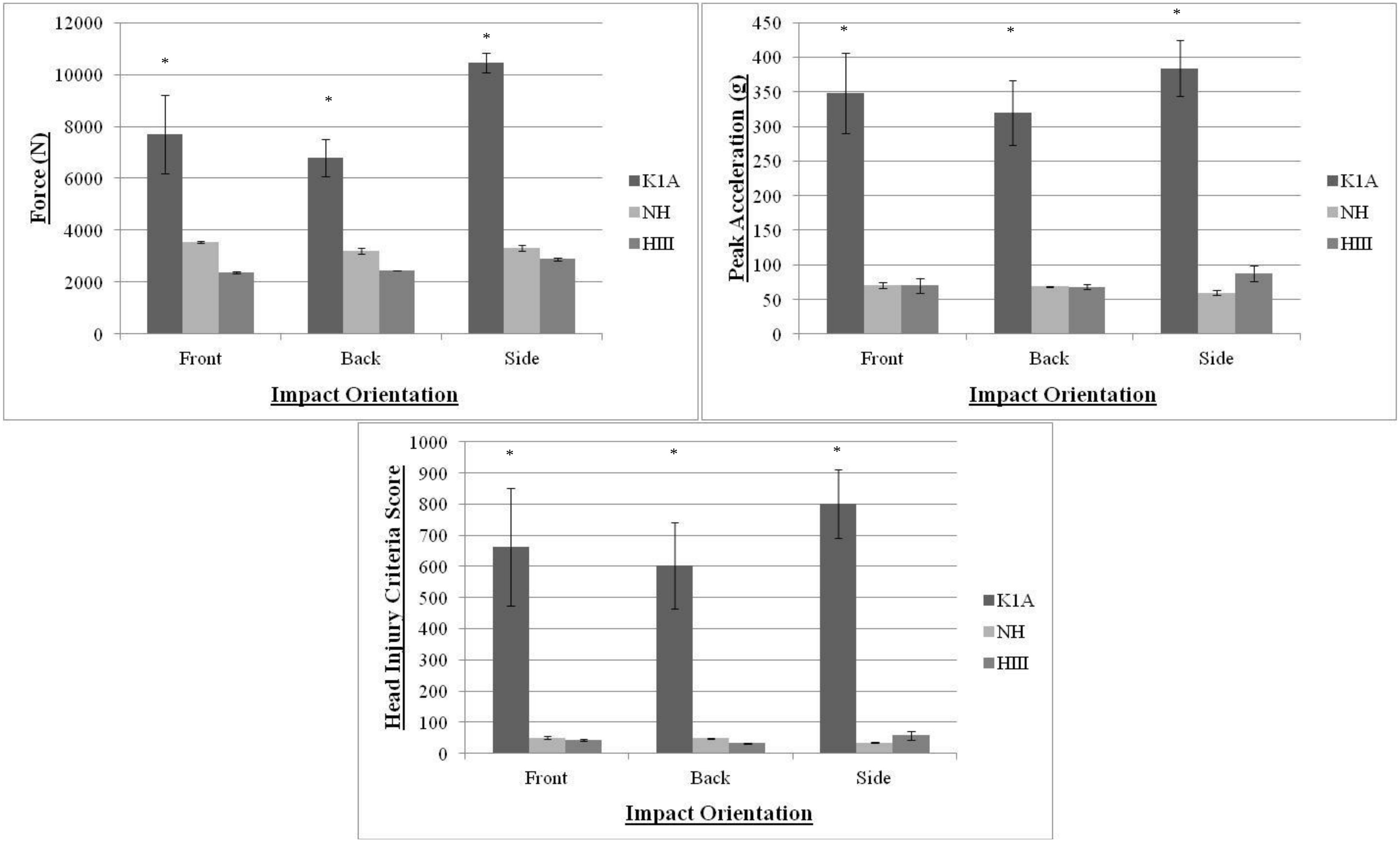


Figure 4-5: Average (SD) outcome variables for all unhelmeted trials separated by impact orientation (* indicates significant main effect of headform, *p<.05*)

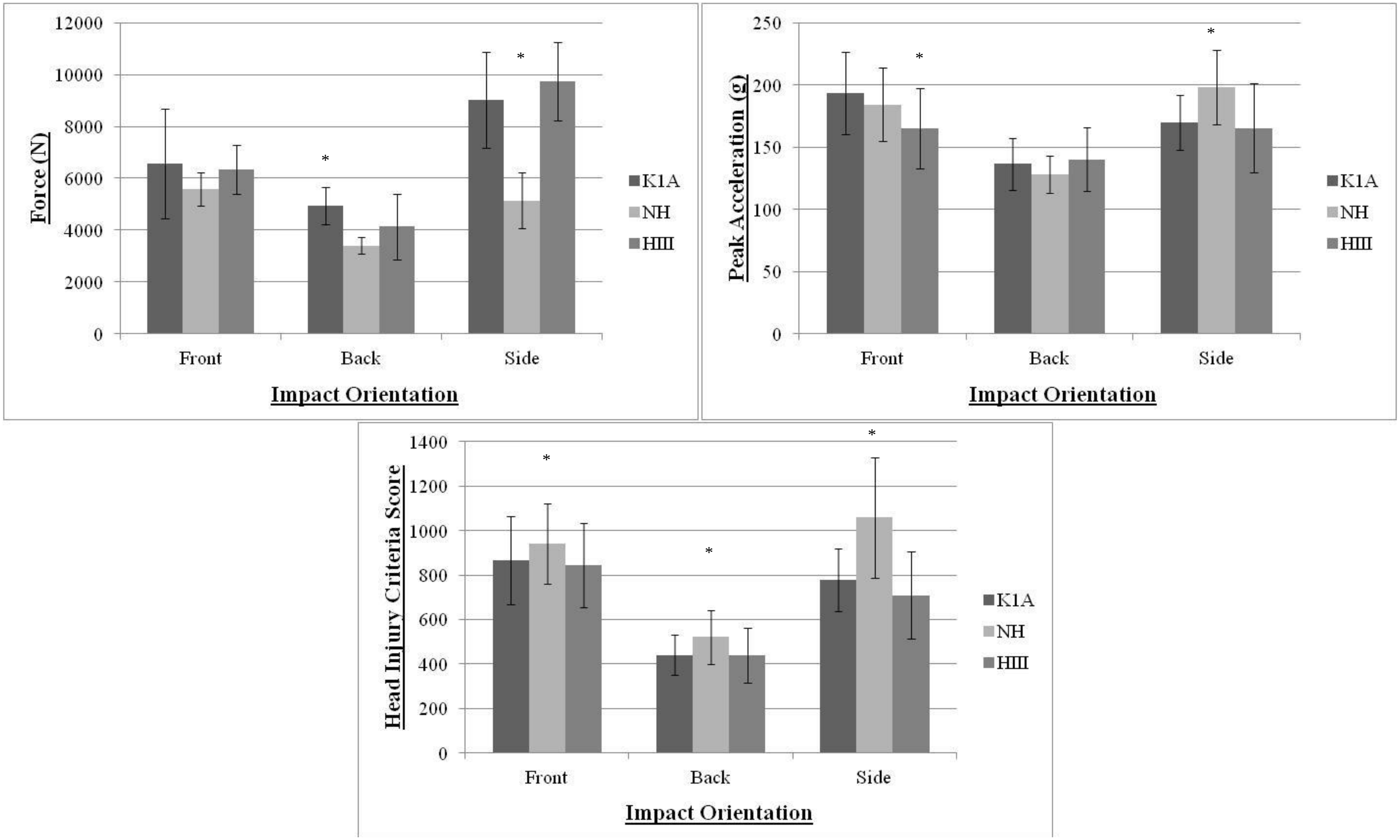


Figure 4-6: Average (SD) outcome variables for helmeted trials separated by headform (* indicates significant main effect of headform, *p<.05*)

4.4.2 Headform differences with a helmet

A three-way ANOVA indicated an insignificant three-way interaction between headform, impact orientation, and helmet type. However, two-way interactions between all variables were significant except *Fmax* for the interaction headform by helmet type (Table 4-4). To determine the influence of each main factor on the outcome variables, further one and two-way ANOVAs were completed to assess adjusted R^2 results. Headform only explained 1.5-12.2% of the outcome variables whereas impact orientation explained 30.7-46.8% and helmet type explained 19.8-39.4% (Table 4-5).

Due to the large influence of impact orientation on the outcomes, two-way ANOVAs were completed for headform by helmet type at each impact orientation. No significant interaction effects were found Table 4-6, $p>.05$). However, the tests indicated a significant main effect of helmet type for all variables (*Fmax*, *Gmax* and *HICmax*) at all impact orientations (front, back, and side of the headform) except for *Fmax* at the front of the headform ($p<.001$-$.021$, Table 4-4, Figure 4-6). A significant main effect was also seen for headform at all impact orientations, although effects were different across the orientations. Main effects were observed for two of three variables at the front and back of the headform (*Gmax* and *HICmax* for the front and *Fmax* and *HICmax* at the back) whereas all three variables were significantly different at the side of the headform Table 4-6).

Post-hoc tests revealed that significance changed between headforms depending on the outcome variable (Figure 4-6). The K1A headform only showed significantly higher values for one outcome variable (*Fmax*) during impacts to the back of the headform (by 19.1-44.8%). Force (*Fmax*) was only significant during side of the head impacts on the NH headform when compared to the other two headforms (differences ranged from 41.5-45.8%, Figure 4-6). Peak acceleration (*Gmax*) values was only significant for two instances, the HIII headform had significantly lower values during front of the head impacts (10.3-14.6%) and the NH headform had significantly higher values during side of the head impacts (18.3-21.3%, Figure 4-6). Finally, *HICmax* values were significantly higher for the NH headform when compared to the other two headforms for all impact orientations (8.7-54.3%, Figure 4-6).

Table 4-4: Three-way ANOVA results of headform, impact orientation, and helmet use for Fmax, Gmax, and HICmax

	Headform		Impact Orientation		Helmet Type	
	F(df)	**p**	**F(df)**	**p**	**F(df)**	**p**
Fmax	27.7(2)	<.001*	79.7(2)	<.001*	13.5(2)	<.001*
Gmax	5.6(2)	.006*	45.1(2)	<.001*	43.7(2)	<.001*
HICmax	55.1(2)	<.001*	199.1(2)	<.001*	142.1(2)	<.001*

	Headform*Impact Orientation		Headform*Helmet Type		Helmet Type*Impact Orientation		Headform*Impact Orientation*Helmet Type	
	F(df)	**p**	**F(df)**	**p**	**F(df)**	**p**	**F(df)**	**p**
Fmax	9.0(4)	<.001*	1.1(4)	.389	4.1(4)	.006*	0.4(7)	.903
Gmax	5.3(4)	.001*	2.8(4)	.037*	3.5(4)	.013*	0.6(7)	.776
HICmax	10.1(4)	<.001*	3.2(4)	.020*	4.0(4)	.006*	1.7(7)	.137

***** indicates significance at an alpha level of 0.05

Table 4-5: Adjusted R^2 results for each independent variable

Factor	Fmax	Gmax	HICmax
Headform	0.122	0.015	0.088
Impact Orientation	0.468	0.307	0.465
Helmet Type	0.198	0.354	0.394

Table 4-6: Two-way ANOVA results of headform and helmet type across each impact orientation

Impact Orientation - Front						
	Headform		Helmet Type		Headform*Helmet Type	
	F(df)	p	F(df)	p	F(df)	p
Fmax	1.4(2)	.284	3.5(2)	.051	0.3(4)	.877
Gmax	5.9(2)	.011*	26.4(2)	<.001*	1.2(4)	.343
HICmax	14.7(2)	<.001*	260.3(2)	<.001*	1.6(4)	.220
Impact Orientation - Back						
	Headform		Helmet Type		Headform*Helmet Type	
	F(df)	p	F(df)	p	F(df)	p
Fmax	5.7(2)	.013*	4.9(2)	.021*	2.3(3)	.111
Gmax	0 .1(2)	.899	6.8(2)	.007*	1.5(3)	.263
HICmax	10.3(2)	.001*	22.5(2)	<.001*	2.1(3)	.139
Impact Orientation - Side						
	Headform		Helmet Type		Headform*Helmet Type	
	F(df)	p	F(df)	p	F(df)	p
Fmax	48.9(2)	<.001*	17.8(2)	<.001*	0.5(4)	.749
Gmax	10.1(2)	.001*	22.0(2)	<.001*	1.5(4)	.240
HICmax	31.8(2)	<.001*	33.9(2)	<.001*	2.2(4)	.115

***** indicates significance at an alpha level of 0.05

4.4.3 Headform repeatability

For unhelmeted trials, the K1A headform was only repeatable within ±5% for one of nine instances whereas the NH and HIII headforms were both repeatable for five of nine instances (Table 4-7). During helmeted trials, the K1A and HII headforms were repeatable within ±5% for 9 of 27 instances and the NH headform was repeatable for 12 of 24 (Table 4-8).

Intra-class correlation coefficients (ICC) (two-way mixed for absolute agreement, average measures) ranged from 0.543-0.991 for the K1A headform, 0.892-0.968 for the NH headform, and 0.701-0.994 for the HIII headform (Table 4-9).

Example raw data trials (force and acceleration data) for all helmets and headforms can be found in Appendix D while Appendix E has examples of all three trials for each helmet.

Table 4-7: Results from unhelmeted repeatability trials at 1 m/s, maximum % range from respective average values (impact velocity ranged from ±4%)

	Front			Back			Side		
	Fmax	Gmax	HICmax	Fmax	Gmax	HICmax	Fmax	Gmax	HICmax
K1A	21.8	16.7	30.8	12.0	16.8	26.7	4.2	11.8	15.7
NH	1.1	7.3	8.3	4.3	0.6	1.7	3.7	6.8	8.8
HIII	1.7	18.3	5.6	0.6	5.2	6.9	1.9	14.6	26.6

*shaded boxes indicate values over the ±5% range

Table 4-8: Results from helmeted repeatability trials, maximum % range from respective average values

	Front			Back			Side		
	Xilo	Cross	Kask	Xilo	Cross	Kask	Xilo	Cross	Kask
K1A	**Impact velocity ranged from ±2.0%**								
Fmax	36.4	42.3	31.0	14.2	7.5	8.1	15.5	18.3	6.4
Gmax	13.4	14.6	5.1	7.1	7.6	16.5	3.8	4.3	5.5
HICmax	1.6	1.4	4.3	17.6	7.9	9.5	3.2	7.0	5.5
NH	**Impact velocity ranged from ±3.9%**								
Fmax	3.4	7.9	10.1	11.5	7.3	NA	2.4	8.1	2.9
Gmax	2.8	4.4	19.4	7.2	3.5	NA	4.9	9.3	2.4
HICmax	4.6	1.9	7.4	20.6	1.3	NA	6.8	18.4	2.0
HIII	**Impact velocity ranged from ±3.0%**								
Fmax	2.2	4.3	1.6	37.6	11.3	20.4	2.9	13.5	17.8
Gmax	4.4	6.9	11.0	14.2	8.3	23.2	5.1	12.9	22.0
HICmax	7.0	5.7	5.1	6.1	6.2	25.2	5.6	17.5	25.3

*shaded boxes indicate values over the ±5% range

Table 4-9: Intra-class correlation coefficients for all headforms

	Unhelmeted			Helmeted		
	Fmax	Gmax	HICmax	Fmax	Gmax	HICmax
K1A	0.915	0.543	0.561	0.835	0.938	0.991
NH	0.895	0.892	0.967	0.968	0.938	0.967
HIII	0.994	0.701	0.866	0.958	0.824	0.951

4.4.4 Helmet testing at low and higher energy impacts

Peak acceleration and HIC scores were compared with their respective IARVs developed for the Hybrid III anthropometric testing device (ATD). At the ‘MV velocity’ of 7.7 m/s, *Gmax* exceeded the IARV of 180g (5% chance of an average male sustaining a skull fracture) for seven out of nine possible scenarios. Furthermore, the two that did not exceed the IARV had values within 2% of the reference value (Figure 4-8). Only two out of nine possible combinations for *Gmax* during the ‘standard velocity’ condition exceeded the IARV, both of which were for the Kask helmet. These trends continued for the HIC scores with all nine combinations at the ‘MV velocity’ exceeding the IARV of 700 (5% chance of an average male sustaining an AIS 4+ brain injury) and only three of nine possible combinations at the ‘standard velocity’ exceeding the reference value (Figure 4-9).

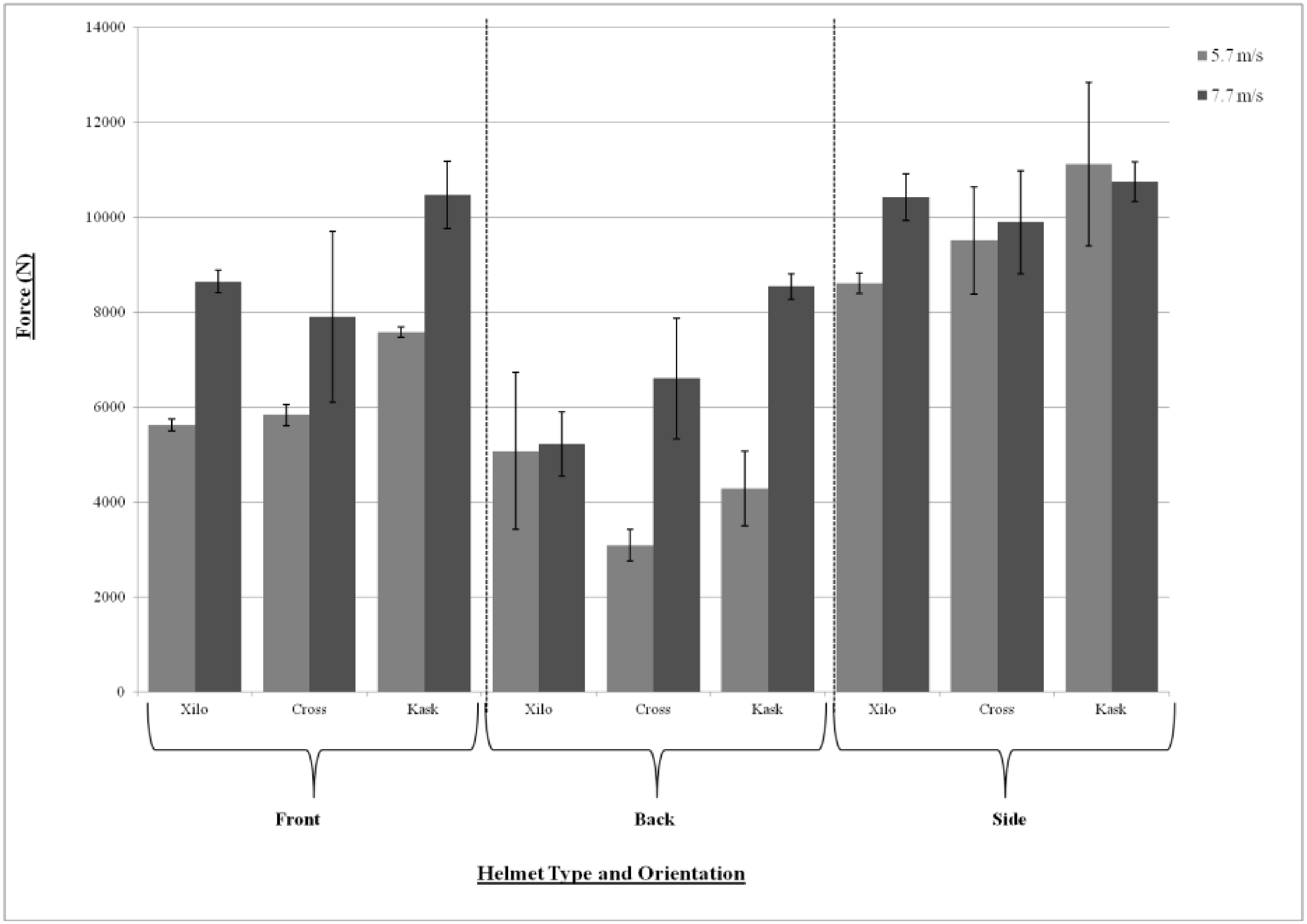


Figure 4-7: Mean (SD) peak force for all drops with the HIII headform at 5.7 and 7.7 m/s for all impact orientations (IARVs and reference values from the CSA are also highlighted on the graphs)

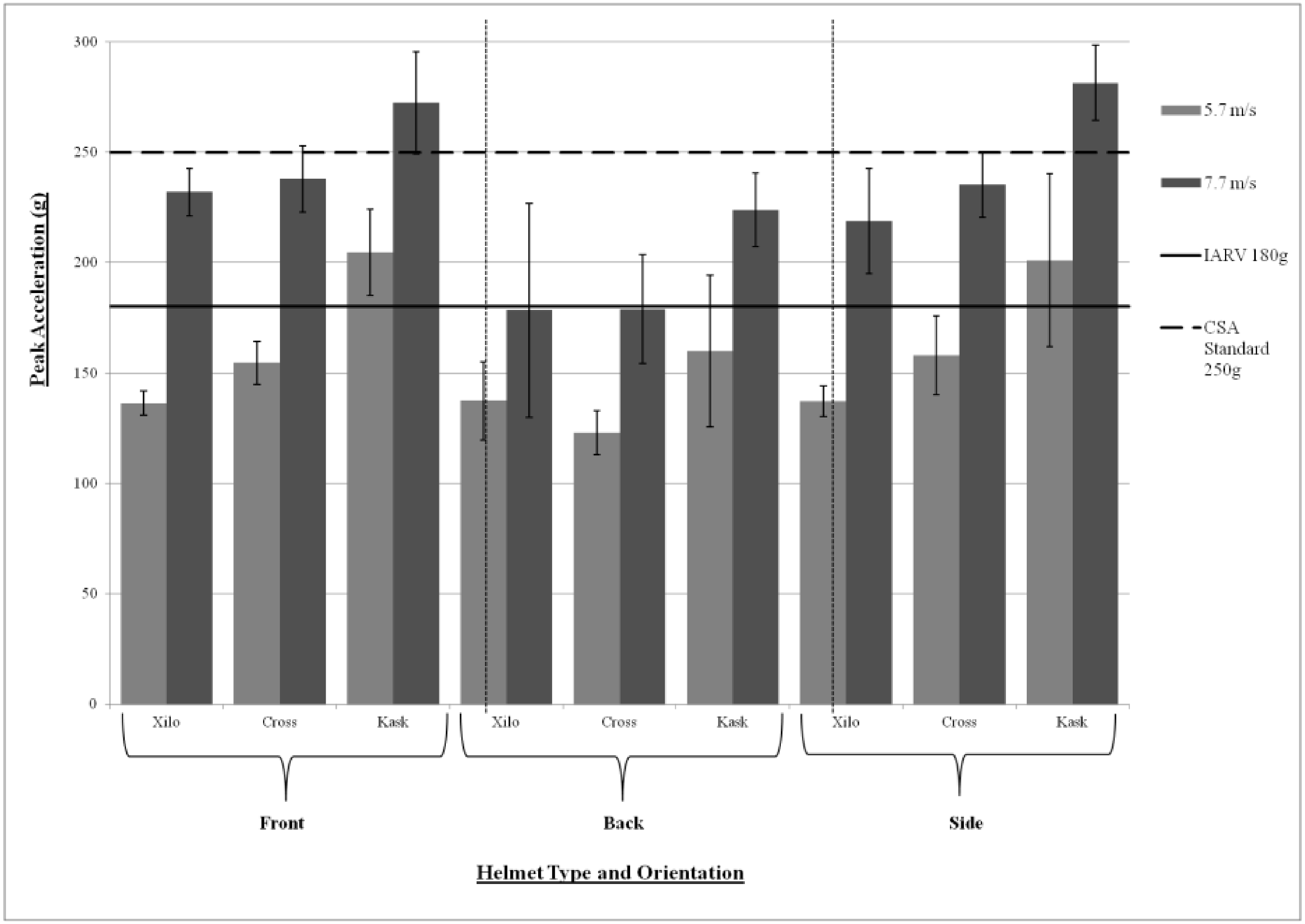


Figure 4-8: Mean (SD) peak acceleration for all drops with the HIII headform at 5.7 and 7.7 m/s for all impact orientations (IARVs and reference values from the CSA are also highlighted on the graphs)

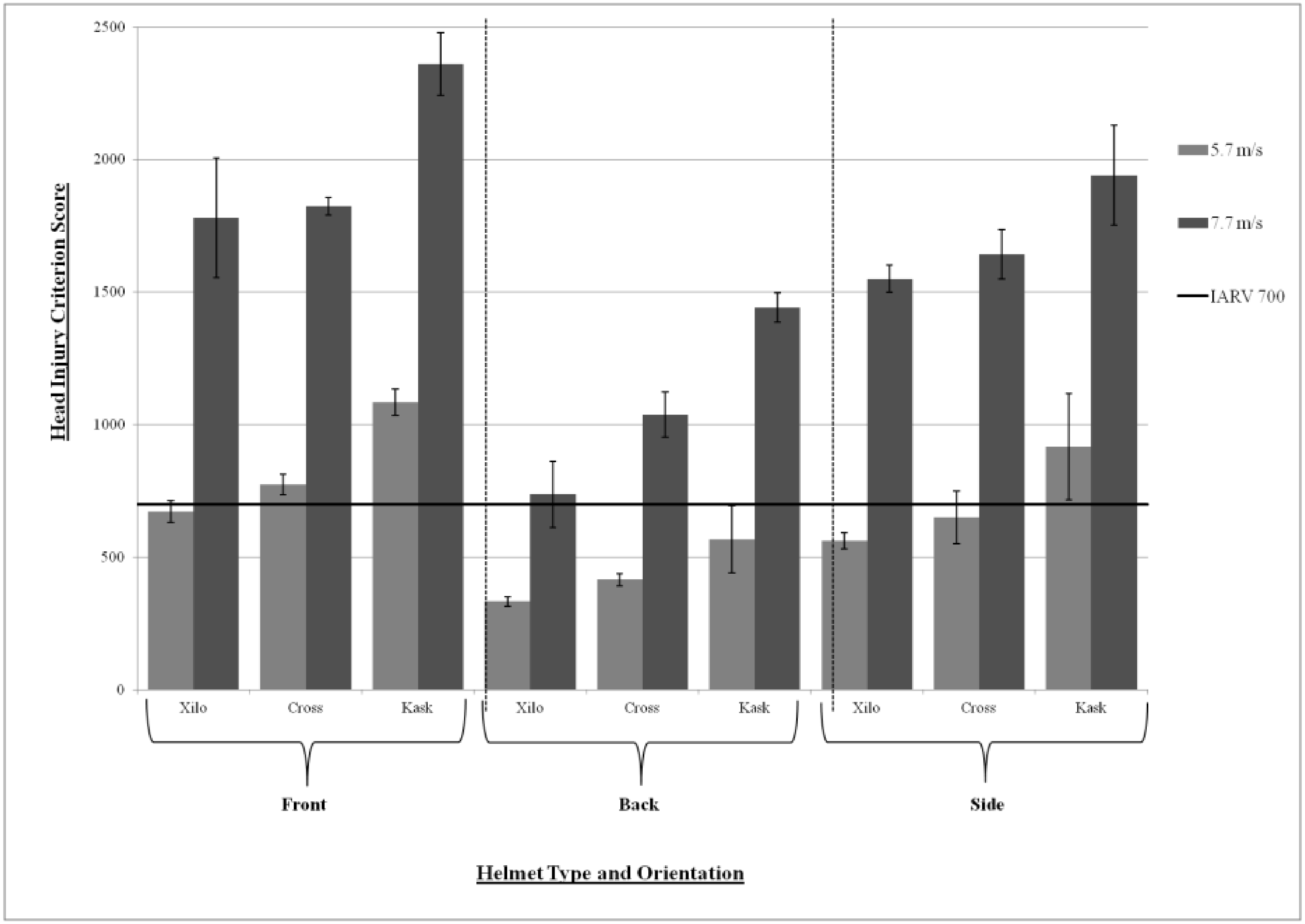


Figure 4-9: Mean (SD) HIC score for all drops with the HIII headform at 5.7 and 7.7 m/s for all impact orientations (IARVs and reference values from the CSA are also highlighted on the graphs)

4.5 Discussion

The goals of the current study were twofold. The first goal was to compare peak dynamic headform responses between three surrogate headforms currently used in various testing standards and test the repeatability of the headform responses. The hypothesis that the K1A headform (the surrogate currently used in cycling helmet standards) would show higher peak accelerations, peak forces, and HIC scores than the HIII and NH headforms was supported for nine out of nine possible comparisons during unhelmeted trials (3 impact orientations * 3 outcome variables (*Fmax, Gmax, HICmax*)). No interaction effects were found, thus, the hypothesis that these comparisons would not change for impact orientations was also supported. During helmeted impacts, peak dynamic headform responses were different when compared to the unhelmeted impacts. A main effect of headform was seen for seven of nine possible comparisons (3 impact orientations * 3 outcome variables). However, post-hoc tests revealed that the trends in outcomes were different compared to unhelmeted trials. The K1A headform only showed significantly larger peak forces during back of the head impacts. Rather, the NH headform had significantly different outcomes compared to the K1A and HIII headforms for five of the seven significant comparisons. Therefore, helmeted impacts influenced the peak responses of the headforms (relating to hypothesis 1c).

The hypothesis that the headforms would be repeatable within ±5% of their average outcome variables (hypothesis 1d) was not supported with a range of one to five out of nine possible instances where the headforms were repeatable within the chosen range. The NH headform appeared to be the most repeatable with all instances for unhelmeted trials falling within ±9% of their average outcome variables. When intra-class correlation coefficients were calculated (two-way mixed, absolute agreement), the K1A headform showed two moderate agreement measurements for peak acceleration and HIC scores during unhelmeted trials, but excellent agreement during helmeted trials. However, the HIII and NH headforms had strong to excellent reliability agreements between outcomes for both unhelmeted and helmeted trials (ICC > 0.7). Impact velocity had some variance across the trials (up to ±4%, Table 4-7, Table 4-8) which may explain some of the differences in repeatability. Other variance may have been explained by the test system. While the headform/ball arm assembly is designed to be rigid, the headforms are attached to the ball arm using four screws and the ball arm is attached to the drop tower with two screws; thus, it is possible that some deformation or movement throughout the trials within these attachments contributed to variability in headform repeatability.

Material properties of the headforms seem to be the most likely contributing factor to the differences between headforms during unhelmeted comparisons. The K1A headform is constructed of a magnesium alloy with no covering (Table 4-1). Due to the rigid properties of the K1A surrogate, upon impacting a stiff target surface such as the metal plate in this experiment, the accelerations are expected to be higher because there is little deformation experienced by both the headform and the target surface. This phenomenon can be explained using the work energy principle and has previously been observed in a study comparing custom-made compliant and rigid headforms dropped onto target surfaces with varying stiffness (Selvan, Halls, Zheng, & Chandra, 2013). The force (*F*) acting on the body of mass (*m*) depends on the deformation (*d*) experienced by the body (Equation 4.3, (Selvan et al., 2013). V_f and v_i are the final and initial velocities of the body.

$$0.5mv_f{}^2 - 0.5mv_i{}^2 = F * d \qquad \textbf{(4.3)}$$

For the more compliant headforms, the urethane covering (NH) and the vinyl covering (HIII) deform upon impact to the stiff target surface, therefore decreasing the force and acceleration acting on the bodies. Furthermore, the stiff headform induces resonance in the test system when impacting the metal plate (as illustrated in Figure 4-4). The noise induced during these unhelmeted impacts may have also contributed to the variance seen in the K1A outcome variables (Table 4-7).

When helmets are involved, the compliant gel in the interior of the NH headform may have contributed to its higher outcome variables (Table 4-1). Selvan et al. (2013) found that their compliant surrogate had higher peak accelerations and HIC values when compared with their rigid surrogate for impacts to a softer target surface. The influence of a helmet would act in a similar way as the softer target surface seen in Selvan et al.'s (2013) work. It provides a less stiff impact area between the surrogate and the target surface. Upon impacts with a helmet, the compliant gel continues to travel forward colliding with the urethane shell which has a higher stiffness than the helmet causing higher peak accelerations than the more rigid headforms (Selvan et al., 2013). This effect would likely simulate more closely the brain experiencing a higher load than the skull when a head impacts a stiff surface while wearing a helmet.

The NH headform also has a slightly higher mass than the other headforms (Table 4-1) which may contribute to the observed differences from the K1A and HIII headforms (Kendall et al., 2012). Stuart et al. (2013) compared dynamic responses between the K1A and HIII headforms during frontal helmeted impacts. They found that the headforms performed similarly with the K1A headform producing slightly higher accelerations (Stuart et al., 2013). The results from the current study agree with these findings for not only frontal impacts but impacts to the side and back of the headform as well (Figure 4-6).

It seems apparent that the HIII headform may be used as an equivalent tool to the K1A headform in helmet testing. Furthermore, the HIII headform has been developed (for frontal impacts) for data collection and calculation of the HIC score which may currently be the most clinically relevant head injury assessment tool (Kendall et al., 2012). The NH headform produced higher outcome variables during helmeted impacts. Therefore, using this headform in testing standards would likely allow for a more conservative testing scenario. Selvan et al. (2013) found that their compliant headform seemed to more closely replicate the dynamic responses of a cadaver head compared to the rigid surrogate which may also be the case for the NH headform in the current study. However, the lack of available (specific) information about the construction of the brain cavity (i.e. location, material properties, pressure etc.) and a quantified comparison to cadaver heads represents concern for use of this headform in head impact research or helmet testing. Finally, the HIII headform was designed to be able to withstand high energy impacts. Therefore, out of the three surrogate headforms investigated, it appears the HIII headform is currently the most useful for head impact research related to cycling helmets.

While the current study was able to investigate peak dynamic responses between three surrogate headforms, without the inclusion of cadavers, no comparisons of the headforms to human data are possible. Therefore, despite the similarities in outcome variables between the more rigid headforms, concerns regarding how closely the headforms simulate the realistic behavior of a skull undergoing the same impacts are still relevant. No three dimensional dynamic information (e.g. rotational acceleration) was collected during this study. This was chosen because the authors chose to use a monorail set-up with a rigid neck to compare the peak headform responses during impacts similar to those used in the standards (Canadian Standards Association, 2009). Completing a similar study using methods to compare three dimensional impact measurements across headforms may provide further insight into injury mechanisms and other tools linked with injury outcomes (e.g. GAMBIT criteria). Finally, no detailed insight into the construction of the headforms was recorded. Specifically, it's likely that the location and pressure of the compliant gel in the interior of the NH headform would play a role in the accelerations experienced by this headform. Without details about the construction, it's hard to draw conclusions about the true mechanisms behind the outcome variables.

Instrumentation to cadaver heads during previous experiments has typically only allowed for measurements of linear acceleration (i.e. three-dimensional dynamic responses were not captured) despite recent suggestions that rotational acceleration may be a large contributing factor in brain injury (Selvan et al., 2013). Thus, future work should not only focus on improving direct comparisons of the surrogate headforms to cadaver heads but also developing technology to advance our measurement of three-dimensional dynamic responses in cadavers with the anticipation of improving injury thresholds and criteria that can then be linked to surrogate headforms and helmet testing.

The second goal of the current study was to compare helmeted responses to impacts at the velocity used in the current standards (5.7 m/s) and a velocity more realistic of cyclist/motor vehicle collisions (7.7 m/s, determined in study one (Chapter 3). As hypothesized, most of the helmets provided adequate injury mitigation at the 'standard velocity' (five of 18 helmet averages exceeded the IARVs) but not at the 'MV velocity' (17 of 18 helmet averages exceeded the IARVs; Figure 4-8, Figure 4-9).

Recent research has investigated the effectiveness of one cycling helmet's ability to reduce the severity of impact during frontal impacts to the helmet (Cripton, Dressler, Stuart, Dennison, & Richards, 2014). The current study characterized the effectiveness of three separate helmets, at three price points (34, 59, 209 CAD), to reduce head injuries at both the velocity currently used in testing standards and a head impact velocity that may be more indicative of realistic collisions for cyclists involving a motor vehicle. It is important to note that the current testing standards are not designed to reduce head accelerations below the values for severe skull and brain injuries and that all three helmet brands produced linear acceleration values well below the testing standard cut-offs (<250g) when tested at the 'standard velocity'. Recent findings in head injury research suggest that concussive impacts can occur at even lower accelerations than the injury assessment reference values (80-100 g) (Rowson & Duma, 2013; Zhang, Yang, & King, 2004). Using this acceleration range, the helmets tested in this study mitigated injury risk, but not below concussive ranges in any condition (Figure 4-8). Helmets were not initially designed to protect against injuries like concussions which is why current testing standard thresholds are set at ~250-300 g (Swart, 2003). However, comparing accelerations to concussive impact ranges provides important information to consumers for understanding the role of their helmet. It may also encourage helmet manufacturers to approach helmet design from a different perspective and set a high benchmark for helmet safety.

All helmets, when impacted at a velocity of 7.7 m/s, exceeded the IARV guidelines putting the risks for brain injuries coded as an AIS 4+ in the range of 5% to >95% and risks for skull fracture in the range of 10-80% (Figure 4-9, (Mertz et al., 2003)). Type of AIS 4+ brain injuries may include hematomas (subdural or arachnoid), large contusions, or diffuse axonal injuries causing a loss of consciousness for greater than six hours (Schmitt et al., 2014). Helmeted impacts at the 'standard velocity' seemed to decrease the range of AIS 4+ brain injury risk from <1% to 20% and skull fracture risk from <1-10% (Figure 4-9).

Similar to previous arguments in support of bicycle helmet effectiveness (Attewell et al., 2001; Cripton et al., 2014; McDermott et al., 1993; Thompson et al., 1996), the current study generally agrees with these findings for impacts up to 5.7 m/s during frontal, back, and side impacts. However, for impacts at a higher energy level (such as 7.7 m/s), the injury mitigation offered by the helmets significantly decreases so that the probability of severe injury may reach up to 95%. Therefore, while the current study still promotes the use of helmets to mitigate injury, the mitigating capacity of helmets decreases following collisions causing head velocities 7-8 m/s, potentially changing head injury outcomes.

Furthermore, injury mitigation appears to depend on the helmet type (i.e. brand) as injury assessment values changed depending on helmet type in the current study. Interestingly, the lower ends of the injury threshold ranges were attributed to the least expensive helmet while the higher ends corresponded to the most expensive helmet (Figure 4-8, Figure 4-9). Correlating price-point with the mitigating capacity of helmets is a relatively new contribution to the helmet literature. It has previously been shown that the price-point of full-face mountain biking helmets also does not necessarily indicate improved injury mitigation as defined by peak acceleration, peak force, and HIC scores (Warnica et al., 2014). Helmet material properties and design of the helmets (specifically foam thickness and density) likely play a large role in the mitigating capacity. Upon visual observation of the current helmets (Xilo, Cross, and Kask), the materials of the more expensive helmet (Kask) appeared to be stiffer and lighter than the less expensive helmets (Xilo and Cross) which appeared to have thicker foam liners. For full-face helmets, the stiffness of the chin-bars seemed to also play a large role in determining the injury mitigation by the helmets with stiffer chin-bars not necessarily improving outcome variables designed to describe the mitigating capacity (Chang, Chang, Chang, Huang, & Wang, 2000; Warnica et al., 2014). Helmet properties of the individual helmet brands in this study also likely influenced the outcome variables and thus, the mitigating capacity. As such, helmet manufacturers should be encouraged to be more transparent about the mitigation the individual helmets offer, either by developing standards that test helmets at different velocities or with a rating system for the helmets mitigating capacity. Future work should continue to assess differences in helmet design, potentially investigating the contribution of foam thickness and volume to energy absorption during impact.

This part of the study had several limitations. The outcome variables chosen were peak measures of the dynamic linear responses. Thus, the underlying mechanisms of the changes in outcomes variables were not captured. Secondly, the methods employed in this study were taken from current testing standards in order to create a controlled experiment. As previously criticized (Hoshizaki & Brien, 2004), the types of impacts seen in testing standards likely do not represent a realistic cyclist fall. To improve this biomechanical representation of a real-world event, other features could be added to the model such as a surrogate neck and torso mass (Nelson & Cripton, 2010). This would also help to provide a system for investigating three dimensional injury mechanics of the impact events (i.e. rotational acceleration, injury mechanisms at the neck).

In summary, the results suggest that the three most common surrogate headforms used in various helmet testing standards do not always produce the same responses to impacts. Helmeted impacts significantly changed the trends of dynamic headform responses to impacts so that the headform produced by NOCSAE with a compliant gel interior produced higher outcome variables than the other headforms which may represent a more conservative estimate of head injury outcomes. However, the specific construction of this headform and how it compares to responses of the human head is unknown. Thus, the HIII headform currently appears to represent the most useful for head impact research as it has the ability to withstand high energy impacts, is linked to clinically relevant head injury assessment tools, and also produces repeatable and reliable results. Researchers should continue to investigate dynamic responses of cadavers, in both one and three dimensions, to be able to design a surrogate headform capable of reproducing human responses to head impacts.

When relating injury severity to impact velocity, it was found that head AIS scores (head injury severity) are only weakly, positively correlated with predicted impact velocity. Therefore, there are likely contributions from other elements in the impact (e.g. helmet materials, impact surface). In order to further understand this correlation and be able to design models to simulate these experiments, future work should focus on investigating relationships between head injury severity and contributions of other impact characteristics.

Finally, in the current study, helmets generally mitigated head injury beneath thresholds for impacts at 5.7 m/s as expected. However, for higher energy impacts (7.7 m/s), injury thresholds were exceeded, and risk for an average male of sustaining a severe injury (skull fracture or AIS 4+ brain injury) increased to a range of 5-95%. Furthermore, head injury risk reduction appeared to be affected by helmet brand with more expensive helmets not necessarily improving injury mitigation. These results indicate the need for future work into helmet design, material properties and helmet mitigating capacity. Helmet manufacturers should be encouraged to be transparent about their products and provide more information to consumers regarding the relative performance of their helmets.

Chapter 5: Thesis Summary

My thesis was designed to address gaps in literature concerning cycling incidents in the region of Southern Ontario and to investigate the relative influence of headform type and impact velocity on outcome variables associated with biomechanical tests used to assess the mitigating capacity of cycling helmets. Study one is novel in several regards. It is the first study to characterize impact scenarios for cyclist collisions and falls in Southern Ontario that resulted in injury claims. The dataset used to characterize these incidents was novel as it was provided by a local forensic engineering company, Giffin Koerth Smart Forensics. These companies have the unique opportunity to access a substantial amount of detail from police and hospital reports as well as any information provided by clients. Furthermore, their experts have often created accident reconstruction reports which contain approximate vehicle and cyclist speeds and orientations. This information is valuable to our collection of scientific research for future understanding and experimental design in the area of cycling collisions. The collected information may also be beneficial to the region of Southern Ontario for designing road infrastructure and promoting cycling safety. Future epidemiological investigators should be encouraged to utilize databases such as those from forensic engineering companies, or insurance companies, who have access to a wide range of characteristics of cycling collisions and falls rather than the limited police or hospital reported data which is typically used.

A second goal for study one was to determine an impact velocity that was more realistic of cyclist/motor vehicle collisions in Southern Ontario. A multi-body modeling software (PC Crash 9.0, Datentechnik Group, Linz, Austria) was used to model a subset of the collisions investigated in study one. The average head impact velocity obtained from modeling 30 collisions was 7.7 m/s., which was on the lower end of ranges seen in previous studies using mathematical models to simulate similar collisions (Carter & Neal-Sturgess, 2009; Fanta et al., 2013; Ito et al., 2014). The multi-body model in PC Crash has only been validated for pedestrians using post-mortem human subject data and not yet for cyclists. However, the purpose of the current reconstructions was to approximate some head impact velocities in the dataset of the first study to provide an estimate of an impact velocity seen in higher energy scenarios that may be used in the third part of study two. The head impact velocities from current data subset were correlated with head injury severity score (AIS score) to provide some validity to the results.

Study two of my thesis was designed to investigate the relative influence of headform type and the realism of test impact energies on outcome variables associated with biomechanical tests used to assess the mitigating capacity of cycling helmets. Peak dynamic headform responses were compared between three surrogate headforms. During unhelmeted impacts, the magnesium headform (K1A, currently used in testing standards) produced significantly higher responses. However, during helmeted impacts, the responses changed between the headforms with the biofidelic NOCSAE (NH) headform, constructed with an interior gel, consistently producing higher responses. Higher peak responses may provide more conservative estimates if this headform were used in helmet testing. However, it is unclear if the NH headform is providing an accurate replication of a human head impact response because there is no published literature available comparing impact dynamics of this headform to cadaver heads. Without more details into the material properties of this headform (specifically the interior gel), accurate comparisons to human head impacts cannot be done. The headform created by General Motors as part of the crash test dummy (Hybrid III) produced reliable and repeatable impacts. It is also able to withstand higher energy impacts without safety devices such as helmets. Finally, the HIII headform has been linked to clinically relevant head injury assessment tools (HIC). Therefore, the current study agreed with previous findings in literature (Stuart et al., 2013) that the Hybrid III headform may be used as an equivalent tool to the currently used magnesium (K1A) headform in helmet testing and determined that, out of the three surrogate headforms investigated, the HIII headform is currently the most useful tool in head impact research related to cycling helmets.

The second goal of study two was to compare helmeted responses to impacts at the velocity used in the current Canadian helmet testing standard (5.7 m/s) and to impacts at a velocity that may be more realistic of cyclist/motor vehicle collisions (7.7 m/s, determined by the second experiment of study one). When tested at the ‘standard velocity’, three brands of cycling helmets (at three different price-points) mitigated injury so that, generally, an adult sized male had less than a 5% chance of suffering a skull fracture or obtaining a severe brain injury. However, when the same brands of helmets were tested at the higher, ‘MV velocity’, the helmets exceeded injury thresholds putting an average size male over the threshold of injury at a 5% chance of obtaining a skull fracture or severe brain injury. At this velocity, risks increased up to 95% in some conditions.

Helmets should still be promoted to mitigate injury in the event of a collision or fall from a bicycle. The current results demonstrated that helmeted tests produced outcome variables below injury thresholds at an impact velocity of 5.7 m/s. However, helmeted impacts at a higher impact velocity representing cyclist/MV collisions (of 7.7 m/s) did not decrease outcome variables below injury thresholds suggesting that, at these higher impact velocities, the mitigating capacity of helmets decreases. Injury mitigation offered by the helmets appears to depend on the helmet type (i.e. brand). Interestingly, the lower ends of the injury threshold ranges were attributed to the least expensive helmet while the higher ends corresponded to the most expensive helmet. Material properties and geometry (volume) of the helmets likely contributed to the differences seen in helmet responses across brands. These are important considerations for future research in helmet design and testing as well as helmet innovation. Finally, the outcomes from this study should encourage helmet manufacturers to be more transparent about the abilities of each helmet brand to mitigate injury.

References

4N6XPRT Systems.Expert AutoStats, vehicle specs database, accident investigation. Retrieved Nov/11, 2014, from http://www.4n6xprt.com/4n6as.htm

American Society for Testing and Materials. (2004). ASTM F1292-04, standard specification for impact attenuation of surfacing materials within the use zone of playground equipment.

American Society for Testing and Materials. (2012). F1446 - standard test methods for equipment and procedures used in evaluating the performance characteristics of protective headgear. *American Society for Testing and Materials,* , F1446.

Amoros, E., Chiron, M., Thélot, B., & Laumon, B. (2011). The injury epidemiology of cyclists based on a road trauma registry. *BMC Public Health, 11*

Attewell, R. G., Glase, K., & McFadden, M. (2001). Bicycle helmet efficacy: A meta-analysis. *Accident Analysis and Prevention, 33*(3), 345-352.

Benson, B. W., Hamilton, G. M., Meeuwisse, W. H., McCrory, P., & Dvorak, J. (2009). Is protective equipment useful in preventing concussion? A systematic review of the literature. *British Journal of Sports Medicine, 43 Suppl 1*, i56-67. doi:10.1136/bjsm.2009.058271; 10.1136/bjsm.2009.058271

Billette, J., & Janz, T. (2011). *Injuries in canada: Insights from the canadian community healthy survey.* Statistics Canada.

Bourdet, N., Deck, C., Carreira, R. P., & Willinger, R. (2012). Head impact conditions in the case of cyclist falls. *Proceedings of the Institution of Mechanical Engineers, Part P: Journal of Sports Engineering and Technology, 226*(3-4), 282-289.

Canadian Standards Association. (2009). *Cycling helmets. D113.2-1989_4211439.* Canadian Standards Association.

Carmont, M. R. (2008). Mountain biking injuries: A review. *British Medical Bulletin, 85*, 101-112. doi:10.1093/bmb/ldn009; 10.1093/bmb/ldn009

Carter, E., & Neal-Sturgess, C. E. (2009). MADYMO reconstruction of a real-world collision between a vehicle and cyclist. *International Journal of Crashworthiness, 14*(4), 379-390.

Chang, C., Chang, L., Chang, G., Huang, S., & Wang, C. (2000). Head injury in facial impact—a finite element analysis of helmet chin bar performance. *Journal of Biomechanical Engineering, 122*(6), 640-646.

Chinn, B., Doyle, D., Otte, D., & Schuller, E. (1999). Motorcyclists head injuries: Mechanisms identified from accident reconstruction and helmet damage replication. *IRCOBI Conference Proceedings and SAE International, 1999-13-003*, 53.

CHIRPP. (2008). *Canadian hospitals injury reporting prevention program - injury brief - injuries associated with BICYCLES.* CHIRPP.

Chow, T. K., Corbett, S. W., & Farstad, D. J. (1995). Do conventional bicycle helmets provide adequate protection in mountain biking? *Wilderness & Environmental Medicine, 6*(4), 385-390.

Chow, T. K., Bracker, M. D., & Patrick, K. (1993). Acute injuries from mountain biking. *The Western Journal of Medicine, 159*(2), 145-148.

CIHI. (2011). *Cycling injury hospitalizations in canada, 2009-2010 from the national trauma registry minimum data set.* CIHI.

CIHI. (2013). *National trauma registry (NTR) - CIHI - retreived 13-07-15.*

City of Toronto. (2011). *Cyclist collision summary.* (No. City of Toronto: Transportation Services Management Centre Traffic Safety Unit).

Consumer Product Safety Commission. (1998). Safety standard for bicycle helmets: Final rule. *CFR Part, 1203.*

Cripton, P. A., Dressler, D. M., Stuart, C. A., Dennison, C. R., & Richards, D. (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 & Prevention, 70*, 1-7.

Curnow, W. J. (2005). The cochrane collaboration and bicycle helmets. *Accident Analysis & Prevention, 37*(3), 569-573.

Depreitere, B., Van Lierde, C., Maene, S., Plets, C., Vander Sloten, J., Van Audekercke, R., Van der Perre, G., & Goffin, J. (2004). Bicycle-related head injury: A study of 86 cases. *Accident Analysis & Prevention, 36*(4), 561-567.

Dixon, J. L., & Brodie, I. K. (1993). The new ISO standard for ice hockey helmets and face protectors: Moving toward international standards harmonization and conformity assessment. *ASTM Special Technical Publication, 1212*, 192-192.

Eilert-Petersson, E., & Schelp, L. (1997). An epidemiological study of bicycle-related injuries. *Accident Analysis & Prevention*, *29*(3), 363-372.

Elvik, R., & Mysen, A. B. (1999). Incomplete accident reporting: Meta-analysis of studies made in 13 countries. *Transportation Research Record: Journal of the Transportation Research Board*, *1665*(1), 133-140.

EN, B. (1997). 1078. helmets for pedal cyclists and for users of skateboards and roller skates. *British Standards Institution, London,*

Fanta, O., Boucek, J., Hadraba, D., & Jelen, K. (2013). Influence of the front part of the vehicle and cyclist's sitting position on the severity of head injury in side collision. *Acta of Bioengineering and Biomechanics, 15*(1)

Gennarelli, T. A., & Wodzin, E. (2006). AIS 2005: A contemporary injury scale. *Injury*, *37*(12), 1083-1091.

Giffin Koerth Forensic Engineering. (2013). Giffin koerth forensic engineering. Retrieved May, 2013, from http://www.giffinkoerth.com/

Highway traffic act - 104 - cycling helmets, Government of Ontario, Canada (2012).

Hagel, B. E., & Barry Pless, I. (2006). A critical examination of arguments against bicycle helmet use and legislation. *Accident Analysis & Prevention, 38*(2), 277-278.

Haileyesus, T., Annest, J. L., & Dellinger, A. M. (2007). Cyclists injured while sharing the road with motor vehicles. *Injury Prevention : Journal of the International Society for Child and Adolescent Injury Prevention, 13*(3), 202-206. doi:10.1136/ip.2006.014019

Hansen, K. S. d., Engesæter, L. B., & Viste, A. (2003). Protective effect of different types of bicycle helmets. *Traffic Injury Prevention, 4*(4), 285-290.

Hillman, M. (1993). *Cycle helmets: The case for and against* Policy Studies Institute.

Hoshizaki, T. B., & Brien, S. E. (2004). The science and design of head protection in sport. *Neurosurgery, 55*(4), 956-967.

Isaksson-Hellman, I. (2012). A study of bicycle and passenger car collisions based on insurance claims data. *Annals of Advances in Automotive Medicine / Annual Scientific Conference ...Association for the Advancement of Automotive Medicine.Association for the Advancement of Automotive Medicine.Scientific Conference, 56*, 3-12.

Ito, D., Yamada, H., Oida, K., & Mizuno, K. (2014). Finite element analysis of kinematic behavior of cyclist and performance of cyclist helmet for human head injury in vehicle- to- cyclist collision. *IRCOBI Conference Proceedings,*

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

Kim, J., Kim, S., Ulfarsson, G. F., & Porrello, L. A. (2007). Bicyclist injury severities in bicycle–motor vehicle accidents. *Accident Analysis & Prevention, 39*(2), 238-251.

Kloss, F. R., Tuli, T., Haechl, O., & Gassner, R. (2006). Trauma injuries sustained by cyclists. *Trauma, 8*(2), 77-84.

Linn, S., Smith, D., & Sheps, S. (1998). Epidemiology of bicycle injury, head injury, and helmet use among children in british columbia: A five year descriptive study. canadian hospitals injury, reporting and prevention program (CHIRPP). *Injury Prevention : Journal of the International Society for Child and Adolescent Injury Prevention, 4*(2), 122-125.

Lustenberger, T., Inaba, K., Talving, P., Barmparas, G., Schnuriger, B., Green, D., . . . Demetriades, D. (2010). Bicyclists injured by automobiles: Relationship of age to injury type and severity--a national trauma databank analysis. *The Journal of Trauma, 69*(5), 1120-1125. doi:10.1097/TA.0b013e3181d0f68b; 10.1097/TA.0b013e3181d0f68b

Mackay, M. (2007). The increasing importance of the biomechanics of impact trauma. *Sadhana, 32*(4), 397-408.

Maimaris, C., Summers, C. L., Browning, C., & Palmer, C. R. (1994). Injury patterns in cyclists attending an accident and emergency department: A comparison of helmet wearers and non-wearers. *BMJ (Clinical Research Ed.), 308*(6943), 1537-1540.

Maki, T., & Kajzer, J. (2001). The behavior of bicyclists in frontal and rear crash accidents with cars. *JSAE Review, 22*(3), 357-363.

Malczyk, A., Bauer, K., Juhra, C., Schick, S., & Münster, C. J. (2014). Head injuries in bicyclists and associated crash characteristics. *IRCOBI Conference Proceedings,*

Marjoux, D., Baumgartner, D., Deck, C., & Willinger, R. (2008). Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria. *Accident Analysis & Prevention, 40*(3), 1135-1148.

McDermott, F. T., Lane, J. C., Brazenor, G., & Debney, E. A. (1993). The effectiveness of bicyclist helmets: A study of 1710 casualties. *Journal of Trauma and Acute Care Surgery, 34*(6), 834-845.

McDowell, M. A., & National Center for Health Statistics (US). (2008). *Anthropometric reference data for children and adults: United states, 2003-2006* US Department of Health and Human Services, Centers for Disease Control and Prevention, National Center for Health Statistics.

Mertz, H. J., Irwin, A. L., & Prasad, P. (2003). *Biomechanical and Scaling Bases for Frontal and Side Impact Injury Assessment Reference Values, SAE Technical Paper*(No. 2003-22-0009)

Mertz, H. J., Prasad, P., & Nusholtz, G. (1996). *Head Injury Risk Assessment for Forehead Impacts, SAE Technical Paper*(No. 960099)

Mills, N., & Gilchrist, A. (2006). Bicycle helmet design. *Proceedings of the Institution of Mechanical Engineers, Part L: Journal of Materials Design and Applications, 220*(4), 167-180.

Mills, N. J. (1990). Protective capability of bicycle helmets. *British Journal of Sports Medicine, 24*(1), 55-60.

Millward, D. (2009,). Cars could have external airbags. *The Telegraph*

Moser, A., Steffan, H., & Kasanický, G. (1999). *The Pedestrian Model in PC-Crash-the Introduction of a Multi Body System and its Validation.*

Mountain equipment coop - MEC. (2014). Retrieved Feb, 2014, from http://www.mec.ca/Main/home.jsp

Nahum, A. M., & Melvin, J. (2002). *Accidental injury: Biomechanics and prevention* Springer.

National Operating Committee on Standards for Athletic Equipment. (2009). *Standard test method and equipment used in evaluating the performance characteristics of protective headgear/equipment.* (No. Document 001 - 08m08b).NOCSAE.

Nelson, T. S., & Cripton, P. A. (2010). A new biofidelic sagittal plane surrogate neck for head-first impacts. *Traffic Injury Prevention, 11*(3), 309-319.

Newman, J. (1986). A generalized acceleration model for brain injury threshold (GAMBIT). *Proc. IRCOBI Conf,* 121-131.

Newman, J. A., Shewchenko, N., & Welbourne, E. (2000). A proposed new biomechanical head injury assessment function - the maximum power index. *SAE International, 2000-01-SC16.*

Ommaya, A., & Hirsch, A. (1971). Tolerances for cerebral concussion from head impact and whiplash in primates. *Journal of Biomechanics, 4*(1), 13-21.

Persaud, N., Coleman, E., Zwolakowski, D., Lauwers, B., & Cass, D. (2012). Nonuse of bicycle helmets and risk of fatal head injury: A proportional mortality, case-control study. *CMAJ : Canadian Medical Association Journal = Journal De L'Association Medicale Canadienne, 184*(17), E921-3. doi:10.1503/cmaj.120988; 10.1503/cmaj.120988

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,*

Pucher, J. R., Buehler, R., & Edward, J. (2011). *Analysis of bicycling trends and policies in large north american cities: Lessons for new york* University Transportation Research Center Region 2.

Ribbers, G. M. (2013). Brain injury: Long term outcome after traumatic brain injury. *International Encyclopedia of Rehabilitation,*

Rivara, F. P., Thompson, D. C., & Thompson, R. S. (1997). Epidemiology of bicycle injuries and risk factors for serious injury. *Injury Prevention : Journal of the International Society for Child and Adolescent Injury Prevention, 3*(2), 110-114.

Robinson, D. L. (2006). No clear evidence from countries that have enforced the wearing of helmets. *BMJ (Clinical Research Ed.), 332*(7543), 722-725. doi:10.1136/bmj.332.7543.722-a

Rodarius, C., Mordaka, J., & Versmissen, T. (2008). *Bicycle safety in bicycle to car accidents* TNO.

Rosman, D. L., & Knuiman, M. W. (1994). A comparison of hospital and police road injury data. *Accident Analysis & Prevention, 26*(2), 215-222.

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

SAE, J. (2003). 211/1, instrumentation for impact Test–Part 1–Electronic instrumentation. *Society of Automotive Engineers, Warrendale PA,*

Schmitt, K., Niederer, P. F., Cronin, D. S., Muser, M. H., & Walz, F. (2014). Methods in trauma biomechanics. *Trauma biomechanics* (pp. 15-53) Springer.

Selvan, V., Halls, V., Zheng, J., & Chandra, N. (2013). Comparing the behavior of homogeneous vs. fluid filled solid headforms under blunt impact loading conditions. *ASME 2013 International Mechanical Engineering Congress and Exposition,* V03AT03A015-V03AT03A015.

Shorten, M. R., & Himmelsbach, J. A. (2003). Sports surfaces and the risk of traumatic brain injury. *Sports Surfaces.University of Calgary, Calgary,* , 49-69.

Simms, C., & Wood, D. (2009). *Pedestrian and cyclist impact: A biomechanical perspective* Springer.

SMARTRISK. (2009). *The economic burden of injury in canada.* (). Toronto, ON: SMARTRISK.

SNELL. (1995). *Standard for protective headgear*. (No. B95).

Stats Canada. (2013). *Canadian community health survey - annual component (CCHS).* ().Stats Canada.

Stuart, C., Cripton, P., Dressler, D., Dennison, C., & Richards, D. (2013). Bicycle helmet efficacy using hybrid III and magnesium headforms. *Ohio State Injury Biomechanics Symposium,* Ohio, USA.

Swart, R. (2003). The history of bicycle helmets - bicycle helmet safety institute. Retrieved August 18, 2013, from www.helmets.org/history.htm

Takhounts, E. G., Eppinger, R. H., Campbell, J. Q., Tannous, R. E., Power, E. D., & Shook, L. S. (2003). *On the Development of the SIMon Finite Element Head Model,*

Thompson, D. C., Nunn, M. E., Thompson, R. S., & Rivara, F. P. (1996). Effectiveness of bicycle safety helmets in preventing serious injury. *Journal of American Medical Association, 276*(24), 1974-1975.

Thompson, D. C., Rivara, F., & Thompson, R. (1999). Helmets for preventing head and facial injuries in bicyclists. *Cochrane Database of Systematic Reviews, 4*

Thompson, M. J., & Rivara, F. P. (2001). Bicycle-related injuries. *American Family Physician, 63*(10), 2007-2014.

Thorn, D. R., Hurt Jr, H. H., & Smith, T. A. (1998). Motorcycle helmet test headform and test apparatus comparison. *16th ESV Conference, Windsor, Canada, NHTSA,*

Transport Canada. (2012). *Canadian motor vehicle traffic collision statistics: 2010 - transport canada*. ().Transport Canada.

Walker, B. (2005). Heads up. *Cycle, June/July,* , 42-45.

Warnica, M. J., Park, J., Cook, G., Parkinson, R., Callaghan, J., & Laing, A. C. (2014). *The influence of helmet type and repeated impacts on the biomechanical effectiveness of full-face mountain biking helmets.* Unpublished manuscript.

Zhang, L., Yang, K. H., & King, A. I. (2004). A proposed injury threshold for mild traumatic brain injury. *Journal of Biomechanical Engineering, 126*(2), 226-236.

Appendix A: Variable Classifications and References for Rationale

Variable	Categories	References
Day	Sun, Mon, Tue, Wed, Thurs, Fri, Sat	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997; Haileyesus, et al., 2007; Toronto, 2011)
Month	Jan, Feb, Mar, Apr, May, Jun, Jul, Aug, Sept, Oct, Nov, Dec	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997; Haileyesus, et al., 2007; Toronto, 2011)
Driver Age	<16, 16-34, 35-54, 55+	(Kim, et al., 2007) – chose to keep driver age groups the same as cyclist age groups
Driver Gender	M, F	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997; Haileyesus, et al., 2007; Lustenberger, et al., 2010; Rivara, et al., 1997; Toronto, 2011)
Cyclist Age	<16, 16-34, 35-54, 55+	(Kim, et al., 2007; Lustenberger, et al., 2010) - used similar groups, the groups were chosen to make sure there were enough cases in the categories
Cyclist Gender	M, F	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997; Haileyesus, et al., 2007; Lustenberger, et al., 2010; Rivara, et al., 1997; Toronto, 2011)
Time of Day	6:00-9:59, 10:00-14:59, 15:00-17:59, 18:00-21:59, 22:00-5:59	(Kim, et al., 2007; Toronto, 2011) and based on the spread of information
Lighting Condition	Daylight, Dusk/dawn, Dark	(Kim, et al., 2007)
Road Surface	Dry, wet, icy, muddy	(Kim, et al., 2007)
Type of Bicycle	Mountain, BMX, Hybrid, Road, Child, Electric	Not previously done in other studies – groups chosen based on spread of information
Colour of Bicycle	Black, white, blue, green, yellow, grey, red, other	Not previously done in other studies – groups chosen based on spread of information
Reflective Lights	Y, N	Not previously done in other studies
Type of Vehicle	Car, pickup, minivan, SUV, van, bus, heavy truck, motorcycle, other	(Kim, et al., 2007)
Cyclist Injury (MAIS)	1, 2, 3, 4, 5, 6(fatal)	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997)
Specific Cyclist Injury (AIS)	1, 2, 3, 4, 5, 6(fatal)	(Amoros, et al., 2011; Eilert-Petersson & Schelp, 1997)
Cyclist Injury	Upper extremities, lower	(Amoros, et al., 2011; CHIRPP, 2008;

Location	extremities, face, head, trunk, neck and spine, internal, fatal, other	Rivara, et al., 1997)
Head Impact Occurrence	Y, N	(CIHI, 2011)
Orientation of Head Impact	Front, side, back, front boss, rear boss, face	(Depreitere, et al., 2004)
Helmet Used	Y, N	(CHIRPP, 2008; Depreitere, et al., 2004; Kim, et al., 2007; Rivara, et al., 1997; Toronto, 2011)
Estimated Cyclist Speed at Impact	0-10, 11-20, 21-30, >30	Not previously done in other studies – groups chosen based on spread of information
Estimated Vehicle Speed at Impact	0-20, 21-40, 41-60, 61-80, >80	Articles (Isaksson-Hellman, 2012) typically used increments of 10 km/h but chose to collapse categories based on spread of info
Location of Incident	Highway, urban, municipal, industrial, service, residential, rural, private	(Amoros, et al., 2011; Kim, et al., 2007; Toronto, 2011)
Direction of Vehicle at Impact	Perpendicular, from behind, head on, sideswipe, door	(Kim, et al., 2007)
Area of Impact	Pedestrian crosswalk, driveway, curb lane (middle), curb lane (curb side), passing lane, intersection, gravel shoulder, on-ramp, bicycle lane	Not previously done in other studies
Source of Injury	A pillar, windshield, window, bumper, fender, hood, ground, crush injuries	(Maki, Kajzer, Mizuno, & Sekine, 2003)

Appendix B: PC Crash model inputs and outputs

Project	Speed of vehicle at impact (km/h)	Speed of cyclist at impact (km/h)	Orientation of vehicle and cyclist at impact	Vehicle type	Vehicle shape	Cyclist's head AIS score	Head orientation at impact	Helmet use (Y/N)	Surface of head impact	Head Impact Velocity (*$Head_{vel}$*) (m/s)
1	46	10	Cyclist at 20°	2008 Chevrolet Suburban	Hatchback	5	Back	N	Hood	13.2
2	40	15	Perpendicular	1996 Pontiac Grand Prix	Sedan	1	Right side	N	Windshield	5.1
3	40	20	Parallel – from behind	1999 Ford Armored Van	Van	5	Back	N	Ground	11.4
4	32.5	22.5	Perpendicular	1998 Mazda Protégé	Sedan	3	Left side	Y	Ground	6.9
5	35	21	Cyclist at 70°	2003 Pontiac Aztek	Sedan	4	Right side	N	Ground	8.2
6	50	10	Perpendicular	2001 Kia Sportage	Sedan	3	Front	N	Passenger Window	4.4
7	50	15	Head on	1992 Dodge Shadow ES 2DR Coupe	Sedan	6	Front Left Boss	Y	Windshield	4.7
8	20	3	Perpendicular	1994 Chrysler Intrepid	Sedan	3	Right side	N	Ground	3.6
9	25	30	Cyclist at 70°	2000 Oldsmobile	Sedan	2	Front	Y	Hood	8.6
10	33	27.5	Perpendicular	2002 Ford Explorer pulling dual axle utility trailer	Van	1	Back	Y	Ground	6.3
11	15	20	Perpendicular	2005 Ford ZX4	Sedan	2	Front	N	Ground	6.7
12	20	20	Perpendicular	2004 Dodge Caravan	Van	2	Right side	Y	Windshield	7.2
13	80	20	Perpendicular	2007 Kia Sportage	Hatchback	6	Left side	N	A pillar	12.0
14	41	11	Perpendicular	1990 Mercedes 300E	Sedan	2	Likely right side	N	Ground	5.2
15	35	27.5	Perpendicular	2009 Dodge Grand Caravan	Van	3	Front Left Boss	N	Windshield	6.5
16	50	8	Perpendicular	1997 Chevrolet Lumina	Sedan	2	Right side	N	Windshield	7.8
17	15	16	Perpendicular	2000 Chevrolet Malibu	Sedan	2	Left side	N	Hood	11.8
18	67.5	30	Parallel – from behind	1988 Oldsmobile Cutlass	Sedan	5	Back	Y	A pillar	16.7
19	27	20	Head on	2006 Pontiac Grand Prix	Sedan	3	Front	N	A pillar	8.0
20	43	10	Perpendicular	1999 Honda Civic	Sedan	2	Left side	Y	Windshield	8.4
21	15	10	Cyclist at 70°	2007 Ford Focus	Hatchback	1	Right side	Y	Ground	5.4
22	31.5	20.5	Perpendicular	1988 GMC Pickup	Van	5	Right side	N	Ground	6.5
23	35	20.5	Perpendicular	1994 Oldsmobile Achieva	Sedan	5	Right side	N	Windshield	6.5
24	65	10	Cyclist at 70°	2006 Toyota Corolla	Sedan	2	Front	N	Roof	6.0
25	20	30	Perpendicular	2004 Toyota Sienna	Van	2	Face	Y	Window	8.6
26	40	27.5	Perpendicular	2005 Ford Taurus	Sedan	5	Left side	N	Ground	7.9
27	41	14	Perpendicular	1991 Plymouth Sundance	Hatchback	5	Front	N	Windshield	6.2
28	26	15	Perpendicular	2008 Ford Escape	Van	3	Back	N	Ground	7.6
29	37.5	20	Cyclist at 120°	2007 Ford Crown Victoria	Sedan	3	Face	N	Windshield	7.6
30	51.5	21.5	Cyclist at 70°	2011 Ford F250	Van	4	Right side	Y	Ground	6.2
									AVERAGE	**7.7**

Appendix C: Sample head impact force and head velocity traces for one model

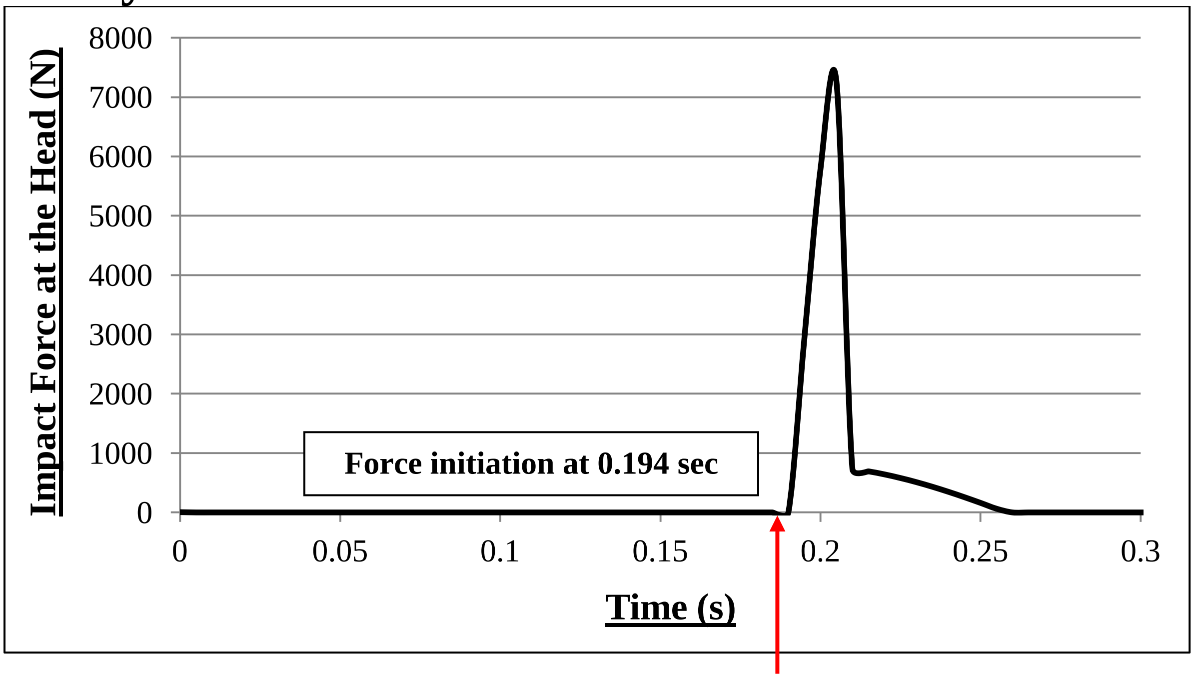


Figure C-1: Force vs. time data for one modeled cyclist/MV collision (#18 in Appendix B, red arrow shows initiation of force at the head)

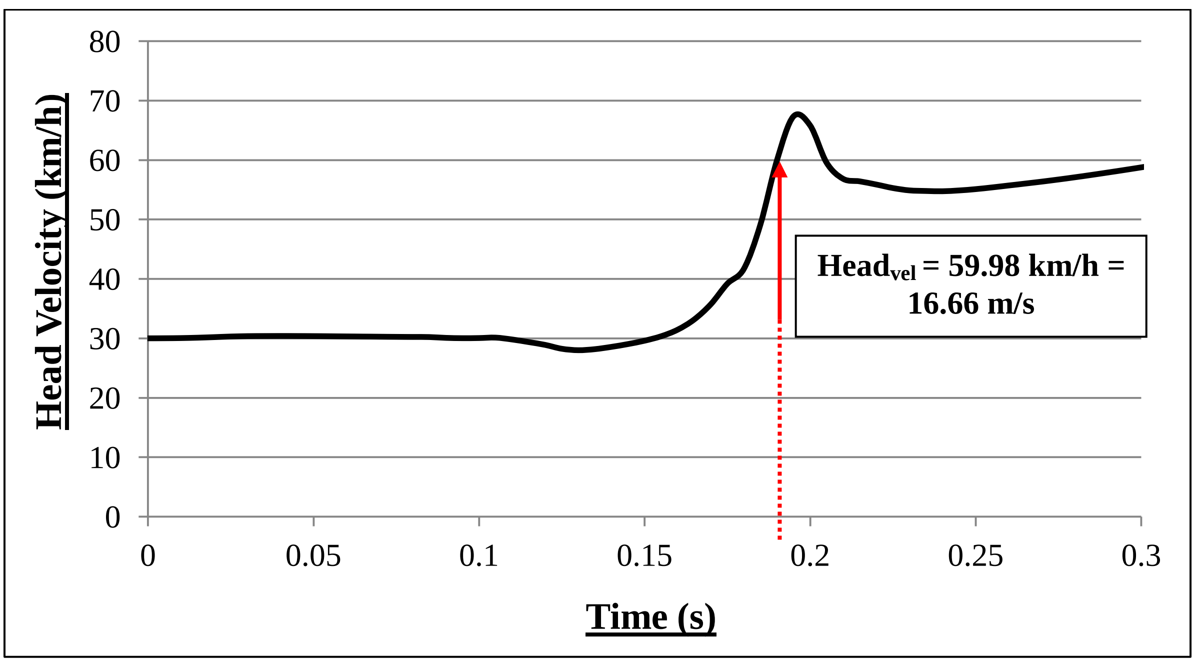


Figure C-2: Head velocity vs. time data for same modeled cyclist/MV collision (#18 in Appendix B, red arrow shows $Head_{vel}$ used for data analysis in Chapter 3, experiment two)

Appendix D: Force and acceleration traces for a sample trial of each condition

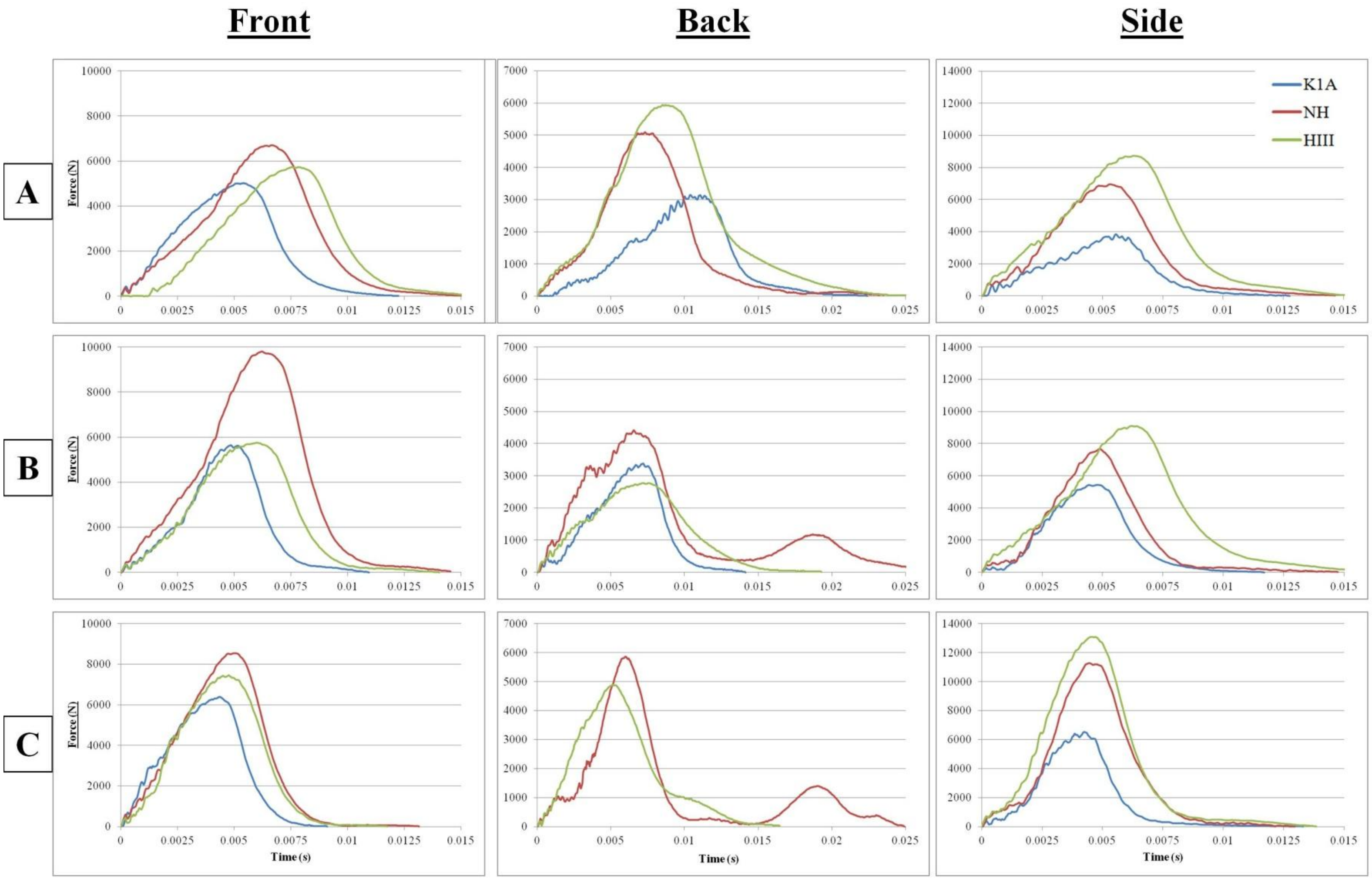


Figure D-1: Force data from sample trials at an impact velocity of 5.7 m/s for three impact orientations of all three headforms and helmets (A - Xilo, B - Crossover, C - Kask)

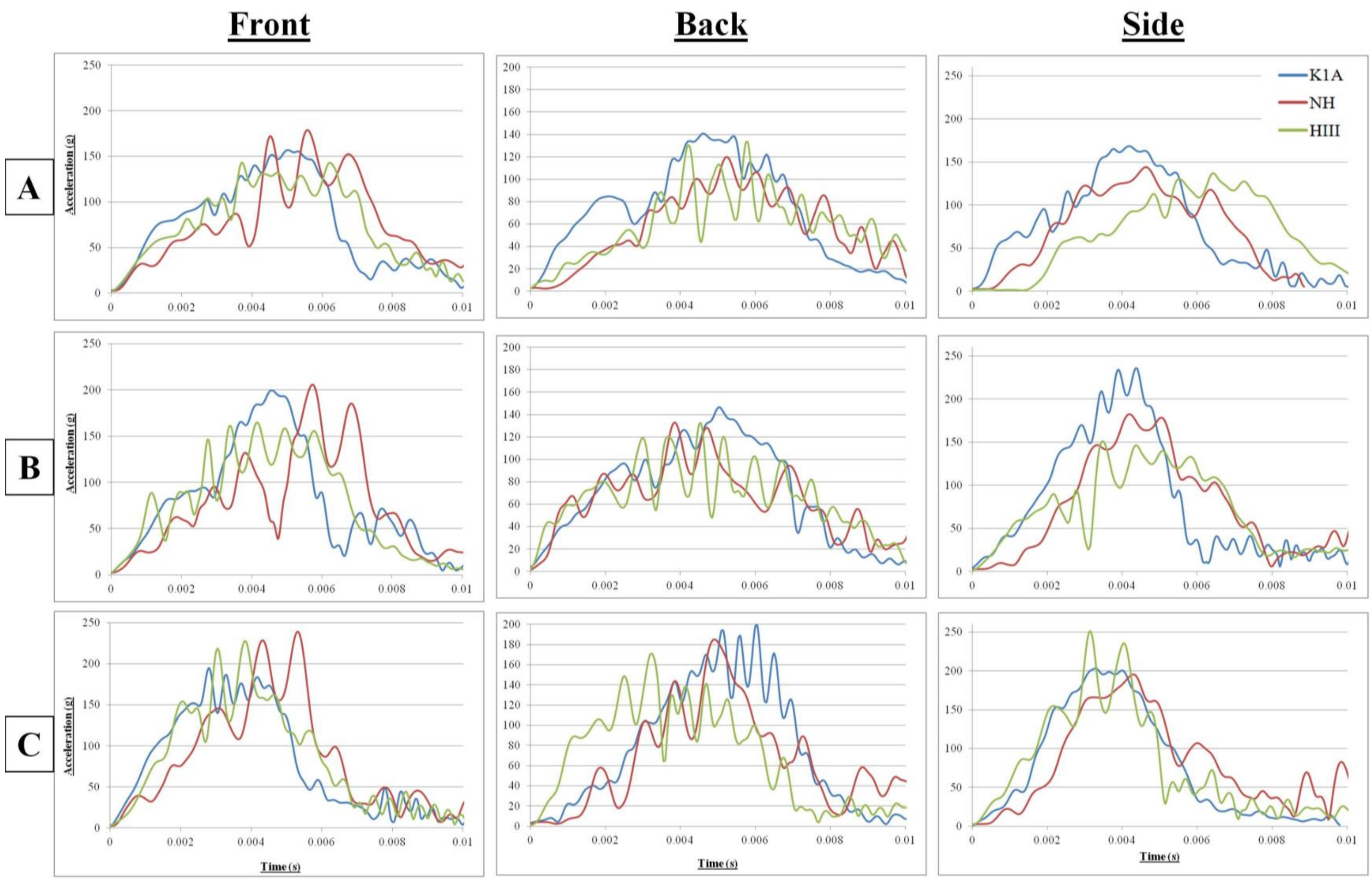
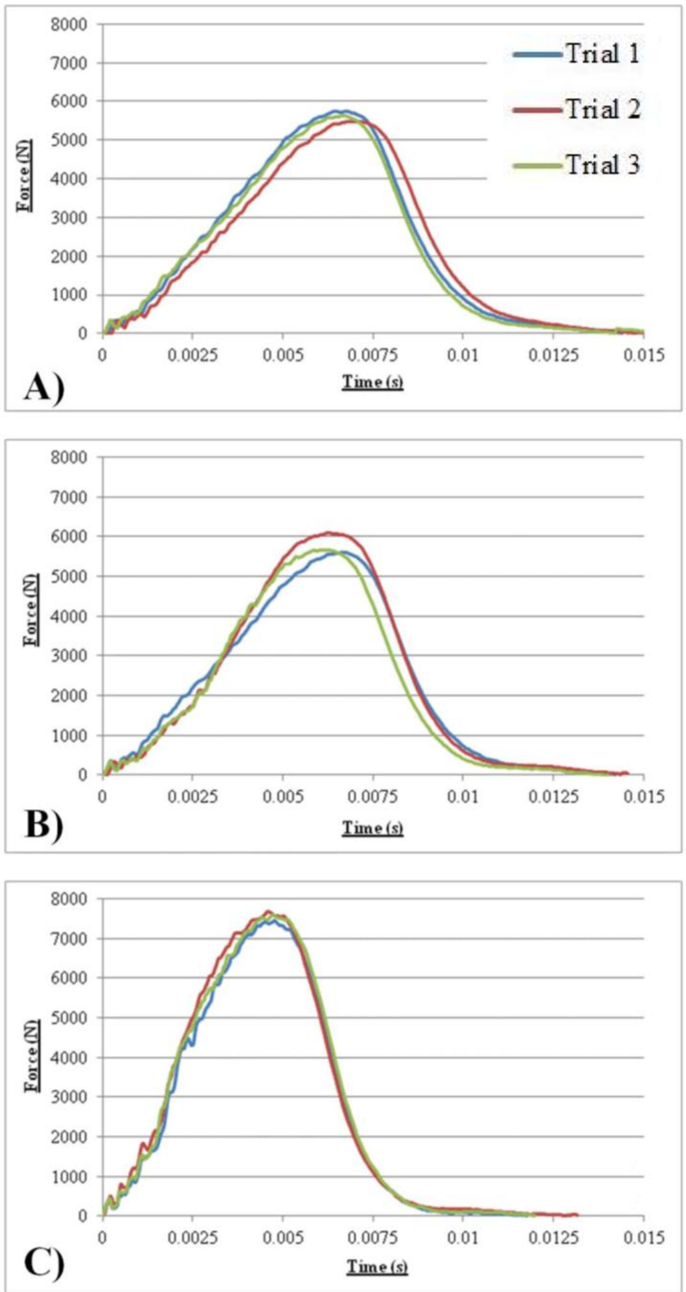


Figure D-2: Resultant acceleration data from sample trials at an impact velocity of 5.7 m/s for three impact orientations of all three headforms and helmets (A - Xilo, B - Crossover, C – Kask)

Appendix E: Sample trials for all helmets during frontal impacts on the HIII headform



c

Figure E-1: : Example force traces for three trials for each helmet brand (A - Xilo, B - Cross, C - Kask) at impact velocity of 5.7 m/s during frontal impacts on the HIII headform.

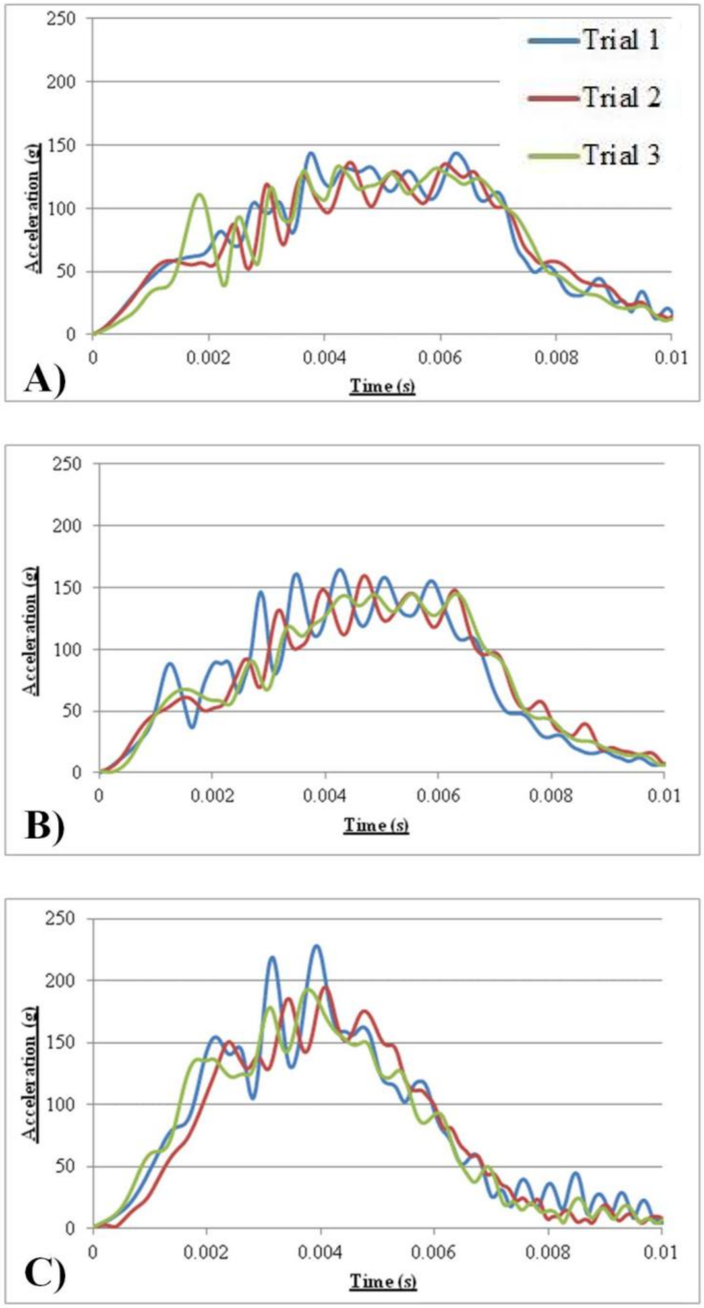


Figure E-2: Example acceleration traces for three trials for each helmet brand (A - Xilo, B - Cross, C - Kask) at impact velocity of 5.7 m/s during frontal impacts on the HIII headform