A Comparison Between Two Oblique Test Protocols for Cycling Helmets

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Abstract

Based on accident reports, oblique head impacts associated with rotational acceleration occur frequently in cycling. Rotational acceleration stimulates brain tissue strain resulting in mild to severe brain injuries. Current bicycle helmet standards test for linear acceleration, but not for rotational acceleration. The proposed standard (EN13087-11) by the European Committee for Standardization (CEN) and the Angular Launched Impact (ALI) protocol are oblique test protocols which impart rotational acceleration to the head at three impact locations (Front_Y, Lateral_X and Lateral_Z). The CEN proposed standard drops the helmeted headform vertically onto a 45° steel anvil, while the ALI protocol launches the headform at an angle of 45° towards the steel surface. The CEN proposed standard may represent a cyclist falling vertically onto a curb, angled surface or motor vehicle. The ALI represents a cyclist skidding or falling over the handlebars and have been described as frequent-accident cases in the literature. Both protocols represent unique falling events in cycling which elicit distinct rotational head responses. The purpose of this study was to compare the dynamic head response and brain tissue deformation between the two oblique test protocols on two common types of cycling helmets (PVC shell-PU liner and ABS shell-EPS liner).

The study revealed that falling vertically onto a curb, angled surface or motor vehicle (CEN proposed standard), resulted in a greater rotational head response and brain tissue deformation, compared to frequent-accident events of skidding or falling over the handlebars (ALI protocol). Linear and rotational acceleration were significantly less on the PVC shell-PU liner compared to the ABS shell-EPS liner on both oblique test protocols. Distinct impact vectors associated with unique falling events in cycling create different rotational head responses and brain tissue deformation. Helmet standards should consider incorporating oblique testing methods, to manage mild and severe brain injuries associated with frequent falling events in cycling.

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$$G(t) = G_\infty + \ (G_0 - G_\infty)e^{-\beta t}$$

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CHAPTER 1: INTRODUCTION

1.1 Problem Statement

Cycling is a form of physical activity used by children and adults for transportation, sport, and recreation (Pucher et al., 2011; McIntosh et al., 2013). Despite cycling's popularity, cyclists have been described as the most unprotected road user groups, where the majority of severe injuries in cycling-related accidents occur to the head (Amoros et al., 2011; Rizzi, 2013). Head injury alone, is the cause of death in 69-93% of fatal bicycle accidents (Guichon & Myles, 1975; Wood & Milne, 1988; Ostrom et al., 1993; Elsen et al., 1997). According to the American Association of Neurological Surgeons, cycling was the most common sport/recreational activity that received head injury treatments in U.S emergency hospitals, followed by American football, and baseball/softball in 2009 (AANS, 2009). In Canada, cycling-related head injuries accounted for 24.7% of hospitalized injuries from 2006-2011 (Teschke et al., 2015). Between 2001-2004, within the Fraser Health Community of Canada, cycling had the highest proportion (34.8%) of concussion hospitalizations for sport and recreational activities (Ramsden et al., 2016).

Accident reports have documented oblique head impacts as the most common impact mechanism in cycling resulting in a head injury (Otte et al., 1999; Mills & Gilchrist, 2008; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013). An oblique impact in cycling involves the head impacting the ground at approximately 30°- 45°, eliciting rotational acceleration to the head (Ching et al., 1997; Aare & Halldin, 2003; Finan et al., 2008; Kleivan, 2013; Willinger et al., 2015). Rotational acceleration causes brain tissue strain, resulting in mild traumatic brain injuries (mTBI) such as a concussion, or more severe injuries such as diffuse axonal injury (DAI) (Holborn, 1943; Aare et al., 2004; Kleiven, 2013; Post and Hoshizaki, 2015). With oblique impacts occurring frequently, rotational mechanisms are considered for incorporation into current bicycle helmet standards, to test a helmet's ability to manage the rotational head acceleration a cyclist would experience in an accident (Halldin et al., 2001; Aare et al., 2003; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013). Out of 1024 computational accident simulations, Bourdet (2012) and colleagues reported skidding and falling over the handlebars as two realistic and classical accident scenarios of a cyclist fall.

In current helmet certification standards, the energy absorption of a cycling helmet is tested. The test involves a vertical impact onto a flat or kerbstone anvil to measure the resultant linear acceleration of the head (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012). However, a proposed standard (EN13087-11), by the European Committee for Standardization (CEN), requires a vertical head impact onto a 45°angled surface to elicit rotational acceleration (CEN, 2015). The design of this method reflects an oblique impact by emphasizing rotational acceleration to the head, to measure the resultant rotational velocity and rotational acceleration (Verschueren, 2009; Bourdet et al., 2012, 2014). The CEN proposed standard performs impacts at three locations; Front_Y, Lateral_X, and Lateral_Z, to induce rotation about the y-axis (within the sagittal plane), x-axis (within the frontal plane), and z-axis (within the transverse plane) of the headform, respectively (CEN, 2015) (Figure 8-Section 4.2.2).

This study introduced an angular launched impact (ALI) protocol that launches a cycling helmeted head at a 30° impact vector (with respect to the horizontal plane) onto a 15° steel anvil for an impact of 45°. The ALI protocol used in this study replicated real, frequent-accident events, such as skidding or falling over the handlebars after hitting a curb/object (Verschuren, 2009; Bourdet et al., 2012, 2014). These events represent a cyclist falling with a horizontal motion, leading to an oblique head impact (Verschuren, 2009; Bourdet et al., 2012, 2014). The CEN proposed standard is a vertical free fall of the head onto a 45° angled surface which may represent a cyclist falling onto a curb, angled road surface, or motor vehicle (Verschuren, 2009; Bourdet et al., 2012, 2014; CEN, 2015; Willinger et al., 2015). These unique falling events reflect differences in rotational head response between the two oblique test protocols. Current helmet standards require only linear acceleration as their pass-fail metric. The absence of a rotational acceleration measure for frequent falling events would make it difficult to predict the risk of brain injuries, such as a concussion (Holborn, 1943; Genarelli et al., 1982; COST 327, 2001; Hoshizaki et al., 2004; Kleiven, 2007; Post and Hoshizaki, 2015).

This study also measured brain tissue deformation as a result of the two oblique test protocols. A study by Kleiven (2013) reported greater strain levels in the brain tissue in oblique impacts due to rotational acceleration. Brain tissue deformation variables, such as maximal principle strain (MPS) and cumulative brain strain damage measure (CSDM), are reported to have the highest correlation in predicting the risk of a concussion and DAI respectively (Willinger and Baumgartner, 2003; Kleiven, 2007; Deck et al., 2008; Takhounts et al., 2013; Post et al., 2015).

An Acrylonitrile butadiene styrene (ABS) shell with an Expanded polystyrene (EPS) absorption liner (ABS shell-EPS liner), and a Polyvinyl Chloride (PVC) shell with a Polyurethane (PU) absorption liner (PVC shell-PU liner) were the two types of cycling helmets used in this study. The main purpose of two different types of helmets was to determine if the variation in dynamic response and brain tissue deformation between the two oblique test protocols were consistent for both types of helmets. This study also compared the rotational response and brain tissue strain between the helmets as they are both common types of helmets worn by cyclists (Depreitere, 2004a). The purpose of this study was to compare the dynamic head response and brain tissue deformation between two oblique test protocols on two types of cycling helmets.

1.2 Significance

In cycling, an oblique head impact is commonly observed in accident scenarios (Otte et al., 1999; Mills & Gilchrist, 2008; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013). The oblique forces at impact, create rotational brain motion, which is associated with mild and severe brain injuries such as a concussion and DAI (Holbourn, 1943; Genarelli et al., 1982; Kleiven, 2007; Kleiven, 2013; Post and Hoshizaki, 2015). Current standards test the absorbing capacity of a cycling helmet using a linear impact. Presently, there is no certification that tests a cycling helmets capacity to manage rotational acceleration. Therefore, helmet standards must incorporate oblique testing methods to measure the performance of cycling helmets under rotational acceleration (Halldin et al., 2001; Aare et al., 2003). This study introduced two oblique test protocols that reflect unique falling events. This is significant as it: 1) Informs helmet standards about the importance of oblique helmet testing, and 2) Informs helmet manufactures about the ability of cycling helmets to manage brain injuries associated with rotational acceleration, such as a concussion.

CHAPTER 2: REVIEW OF LITERATURE

Oblique head impacts are a common impact mechanism in cycling (Otte et al., 1999; Mills & Gilchrist, 2008; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013). Bicycle helmet standards test under linear conditions, which may not elicit the rotational head response associated with an oblique impact (Holborn, 1943; Genarelli et al., 1982; COST 327, 2001; Hoshizaki et al., 2004; Kleiven, 2007; Post and Hoshizaki, 2015). Therefore, oblique testing methods are necessary to determine how a helmet mitigates rotational acceleration associated with mild and severe brain injuries. (Halldin et al., 2001; Aare et al., 2003; CEN, 2015). The literature review will describe the characteristics of an oblique head impact. Head impact velocity, the angle of the head prior to impact, impact location, and oblique testing rigs will be discussed. Additionally, this review will describe the two types of helmets that were tested using the two oblique test protocols. Lastly, the review will discuss dynamic head response and brain tissue deformation which are common variables to predict the risk of head injuries in sport.

2.1 Head Impact Velocity

Computing the speed of a cyclist's head prior to impact requires advance technological devices/software. Using MADYMO (Mathematical Dynamics Model) software, Bourdet and colleagues (2012) simulated a total of 1024 cycling falls to predict the velocity and orientation of the cyclist's head upon impact. Falls due to either skidding or a fall to the ground after hitting a curb was specifically investigated and described as real frequent-accident cases. Bourdet and colleagues (2012) reported a head impact resultant velocity of approximately 6.9 m/s for skidding, and 6.4 m/s for hitting a curb when a cyclist is moving at a speed of 5.5 m/s. Normal and tangential velocities averaged 5.7m/s and 3.7m/s respectively for both impact events.

Previous studies also determined an average impact velocity of 6.9m/s to the ground in collisions involving a passenger vehicle (Otte et al., 1999; Richter et al., 2007; Verschueren, 2009; Bourdet et al., 2014). Injuries associated with these velocities ranged from skull fractures, contusions, and DAI's (Verschueren, 2009). Current standards test at 4.57 m/s to 6.2 m/s (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012). The CEN proposed standard tests at 6.5m/s to reflect the average head impact velocities in accident events (Verschueren, 2009; Bourdet et al., 2012, 2014; CEN., 2015). The ALI protocol also launched the head at an approximate velocity of 6.5m/s.

2.2 Head Impact Orientation

The orientation of a cyclist's head prior to impact is important in characterizing an oblique impact. Accident statistics from Germany, Canada, Belgium, and Finland determined an average cyclist's head impact angle of 30°- 45° to the ground (Harrison et al., 1996; Otte et al., 1999; Richter et al., 2001; Verschueren, 2009). However, current standards continue to test helmets under linear conditions (CEN, 2015). Pure linear impacts are uncommon and mainly result in injuries associated with skull fractures (Kleiven, 2013). Therefore, standards limit their helmet testing to select injuries such as a skull fracture, and may not test to cover all impacts involved in mTBIs and other severe brain injuries.

Bourdet et al (2012) reported head impact angles of 33.5° and 36° in skidding and curb hitting in their study, respectively. In a reconstruction of accident cases between a cyclist and a motor vehicle using MADYMO, reported angles ranged from 5° to 62° (Bourdet et al., 2014). Other studies also reported 30°-45° as average impact angles (Ching et al., 1997; Aare & Halldin, 2003; Finan et al., 2008; Kleivan, 2013). The CEN proposed standard does not angle the head prior to impact but uses a 45° impact surface to elicit rotational acceleration. The CEN proposed standard does not account for a cyclist falling with a horizontal motion observed in common accident events (Verschuren, 2009; Bourdet et al., 2012, 2014). However, the angled surface elicits rotational head acceleration to simulate an oblique impact (CEN, 2015). The ALI protocol will launch the head at 30° with respect to the horizontal plane onto a 15° (total impact angle of 45°) impact surface to cause rotational acceleration, and reflect the horizontal fall endured by a cyclist when skidding or falling over the handlebars (Verschuren, 2009; Bourdet et al., 2012, 2014).

2.3 Impact Location

Depreitere et al (2004b) examined the mechanical input on the head in 86 bicycle accidents (collisions with a motor vehicle and falls). The most common site of impact occurred to the parietal, temporal, and frontotemporal regions of the head. The injuries associated with these impact sites were skull fractures and brain lesions, such as extradural hematoma and DAIs (helmeted and non-helmeted cases). These findings reported that, along with frontal impacts, side impacts are a frequent site of impact (Williams, 1991; McIntosh et al., 1995; Ching et al., 1997; Zhang et al., 2001; Kleivan, 2003). Studies also observed the impact marks of a helmet after cycling accidents and found common impact locations at the front and side (Williams, 1991; Smith et al., 1994; Ching et al., 1997; McIntosh et al., 1998; Otte et al., 2014).

The CEN proposed standard tests at three impact locations; Front_y, Lateral_x and Lateral_z (CEN, 2015). These locations were also tested on the ALI protocol. Each location elicits rotational acceleration about the axis of the headform (x, y and z-axis). Impact locations for the CEN proposed standard were based off selected accident statistics (McIntosh et al., 1998; COST 327, 2001; Bourdet et al., 2012, 2014). McIntosh et al (1998) conducted a field study on 42 cases of pedal cycling accidents to determine helmet effectiveness during vehicle-related and single-cycle accidents. It was reported that a majority of non-fatal head injury cases received an impact to the temporal/parietal regions of the head. COST 327 (2001) investigated detailed motorcyclist head injuries for a two-year period (1996-1998) in Finland, Germany, and the United Kingdom. Brain behaviour as a result of impacts ranged from uninjured, concussions, and skull fractures (COST 327, 2001). Locations of helmet damage were: 26.9% lateral right, 26.3% lateral left, 23.6% frontal, and 21% rear. Bourdet et al (2012) also reported common impact locations at or below the helmet rim line and front-parietal area of the head. These locations are consistent with previous studies which reported that a majority of impact locations also occurred at or below the rim/borderline of the cycling helmet (Chang et al., 1997; Serre, 2007).

2.4 Helmet Type

The two helmets used in this study (ABS shell-EPS liner and PVC shell-PU liner) are classified as urban-style bicycle helmets, and are a modern and common style worn by recreational cyclists (Demarco et al., 2016; Helmets.org, 2017). The ABS and PVC shells are thermoplastic shells and make up the exterior portion (outer surface) of the helmet. The ABS is characterized by its hard and thin shell and is heavier in weight compared to the PVC shell (Chang et al., 2003). The PVC is characterized by its micro shell and lighter weight due its Zip Mold + liquid technology. The purpose of the shell is to disperse the impact energy over a large surface, as well as to protect and hold the inner absorption liner together (Depreitere, 2004a).

The inner absorption liner is characterized for its single high impact energy management and its capacity to absorb approximately 95% of impact energy (Hoshizaki, 2000; Depreitere, 2004a; Cernicchi et al., 2008; Hoshizaki et al., 2014). EPS is the most common absorption liner used in cycling helmets due to its low cost, low density, and sufficient absorption characteristics (Depreitere, 2004a). PU is a similar crushable foam to EPS, however, it is heavier and less commonly used than EPS (Depreitere, 2004a; Helmets.org, 2017).

A study by Hansen et al (2003) documented the types of helmets used by patients involved in bicycle-related head injuries. They reported the odds ratio (OR) of sustaining a head injury using an ABS shell helmet (OR=0.36) was less, compared to a foam shell helmet (OR=0.83). In a case study of cyclists who reported to the emergency room, there was an increase in head injury from users of a PVC/polycarbonate (PC) shell helmet over an ABS shell helmet (Thompson et al., 1996). Furthermore, in a study comparing BMX-style helmets to traditional helmets, the PVC shell-PU liner helmet experienced similar linear acceleration response curves compared to an ABS shell-EPS liner helmet when tested under the CPSC standard. (Demarco et al., 2016).

Micro shell helmets such as the PVC shell-PU liner are far more popular than the hard-shell helmets, such as the ABS shell-EPS liner due to their lighter weight (Depreitere, 2004a). Both helmet types, tested under the same standards, mitigate linear acceleration under 250-300g. Currently, there is no documentation on the dynamic response and brain tissue response between different helmet types on oblique impact methods. However, it would be informative to determine their rotational and brain tissue response under oblique conditions (Demarco et al., 2016). Therefore, this study compared the ABS shell-EPS liner and PVC shell-PU liner helmet using the two oblique test protocols, as well as to determine if the variation in response between the two oblique test protocols are consistent across each type of helmet.

2.5 A Summary of Current Helmet Standards and Oblique Laboratory Designs

Current testing standards follow a similar protocol to test the impact attenuation of a cycling helmet (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012). Modifications of the current standard have been done to test a helmet's response from an oblique impact. This section will review current standards and oblique testing rigs designed in previous studies.

2.5.1 Helmet Testing Standards

Current standards test the energy absorption of a helmet in a linear impact onto a flat steel anvil and/or kerbstone (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012). This is carried out using a twin wire or monorail drop rig tower for a guided free fall of the headform. Impact velocity ranges from 4.2m/s-6.2m/s, depending on the standard. Each impact must not exceed 250-300g of linear acceleration in order for a helmet to be approved and placed on the market. The design of current standards does not apply rotational acceleration to the head to fully represent the mechanism of frequent accident cases (Aare & Halldin, 2003; CEN, 2015).

2.5.2 Oblique Laboratory Testing Rigs

The first oblique testing rig was designed by Harrison and colleagues (1996). Their study investigated the effects of a tangential velocity component to a jockey cap's surface at impact. Video of jockey falls revealed a tangential velocity component of the jockey cap parallel to the ground. To replicate this mechanism, they vertically dropped a headform to impact a horizontally launched surface plate to produce rotational acceleration. Halldin and colleagues (2001) replicated a similar design for oblique testing in motorcycle helmets, however, the surface plate was covered with grinding paper to represent asphalt. This design was used to test head acceleration for bicycle crash simulations (Aare & Halldin, 2003; Mills & Gilchrist, 2008; McIntosh et al., 2012).

The CEN proposed standard drops the headform onto a 45° anvil to test the effects of rotational motion to the head (CEN, 2015). The impact surface is also covered in grinding paper to represent asphalt (coefficient of friction = 0.5). Lon's (2014) study compared the CEN proposed standard to the launched surface plate method. The study measured 5 trials of peak rotational acceleration, rotational velocity and linear acceleration of frontal area (Front_Y) impacts at 6 m/s for each method. The CEN reported greater mean rotational acceleration (9640 rad/s^2), rotational velocity (33.88 rad/s) and linear acceleration (149.24g), compared to the reported launched plate method's mean rotational acceleration (7500 rad/s^2), rotational velocity (25.78 rad/s), and linear acceleration (137.56g). The study further reported the CEN proposed method to be ergonomically usable for one investigator and fewer variables to control. The launched plate method effectively simulated falls to the ground of a single bicycle accident (no exterior factors), and a cyclist crashing into a moving vehicle, however, less ergonomically usable for one investigator. The variability or standard deviation for rotational acceleration, rotational velocity and linear acceleration was not reported.

No study has directly compared oblique testing designs to current helmet standards, despite the ongoing development of oblique testing designs for cycling helmets. A 2017 Folksam report tested eleven different bicycle helmets under a shock absorption test, using the CEN 1078 (current standard) and the CEN proposed standard (Stigson et al., 2017). The results reported a mean linear acceleration for the helmets of 175g for the shock absorption test. Rotational acceleration and rotational velocity were not measured for the shock absorption test. The CEN proposed standard reported a lower average linear acceleration of 119g for Front_y, 129g for Lateral_x, and 117g for Lateral_z locations. This study used one impact location (crown) at an impact velocity of 5.42m/s for the shock absorption test. Furthermore, the shock absorption test used a Magnesium K1A headform, which measured only linear acceleration about the x, y and z-axis, but not rotational acceleration or velocity. The CEN proposed standard uses a Hybrid III headform, which contains an array of nine accelerometers to measure both linear and rotational acceleration about each axis of the headform. Additionally, this study tested at an impact velocity of 6.0m/s for the CEN proposed standard (Stigson et al., 2017). Due to the methodological differences between the current standard (shock absorption test) and the CEN proposed standard, it would not be feasible to do a direct comparison unless location, headform type, and impact velocity across both methods were identical.

2.6 Measuring Helmet Performance

Dynamic response variables; rotational acceleration, rotational velocity, and linear acceleration characterize the inertial response of a head injury (King et al., 2003; Kleivan, 2013; Post & Hoshizaki, 2015). The dynamic response curves are used to determine brain tissue strain such as MPS and CSDM through finite element modelling (FEM) (Ueno et al., 1995; King et al., 2003; Kleivan, 2007; Deck et al., 2008). This section of the literature review will describe the variables used to measure the response of the cycling helmets on the oblique test protocols.

2.6.1 Rotational Acceleration & Rotational Velocity

The biomechanics associated with brain injury have focused on two injury types: injury resulting from linear acceleration, and injury resulting from rotational acceleration (Rowson and Duma, 2013). Rotational acceleration is a contributing variable to brain injury (Holbourn, 1943; Gennarelli et al., 1983; Hoshizaki and Brien, 2004). Holbourn et al (1943) used a rotating narrowed- necked flask, filled with water, to demonstrate the interaction between the brain and the skull. The rotational acceleration caused by a sudden rotation to the flask resulted in water staying behind and the flask rotating. However, the water particles attached to the side of the flask became dissociated, illustrating how rotational acceleration contributes to axonal damage in the brain due to the straining of brain tissue. Subsequently, linear acceleration is associated with skull fractures (Mertz et al., 1997; Kleiven, 2013; Post and Hoshizaki, 2015), and rotational acceleration is a significant variable to predict the risk of sustaining a concussion (Gennarelli et al., 1971, 1972, 1981, 1982, 1987; Zhang et al., 2004; Post and Hoshizaki, 2015; Kleiven, 2013).

Oblique impacts elicit rotational acceleration to the head causing strains at the brain tissue level (Kleiven, 2013; Post and Hoshizaki, 2015). Kleiven (2007) compared the response of direct and angled head impacts using FEM. Angular impacts resulting in rotational motion of the brain produced injuries such as DAI due to the incompressible properties of the brain tissue. With evidence reporting frequent oblique impacts (Otte et al., 1999; Mills & Gilchrist, 2008; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013), helmet standards should develop testing methods to elicit rotational acceleration (Halldin et al., 2001; Aare et al., 2003).

Rotational velocity is also a measure associated with mTBIs (Holburn, 1943). It measures the dynamic motion of the brain with respect to the skull leading to brain strain (Hardy, 2001; Takounts et al., 2013). When using the Simulated Injury Monitor (SIMon) Finite Element Head, rotational velocity has been found to correlate better with MPS and CSDM when compared to rotational acceleration (Takhounts et al., 2008; Takhounts et al., 2013; Ouckama & Pearsall, 2014). The test involved pendulum impacts on a hybrid III headform at unspecified impact locations and angles. Therefore, rotational velocity is an important measure, alongside rotational acceleration, when describing the biomechanics of brain injury. In the 2017 Folksam report, eleven different conventional cycling helmets were tested on the CEN proposed standard-the average rotational acceleration about the x (parietal level), y (upper part of the helmet), and z (rear part of the helmet) axis was 7.1 krad/s^2, 6.97 krad/s^2 and 6.21 krad/s^2, respectively. Mean rotational velocity was 26.0 rad/s, 32.6 rad/s and 26.7 rad/s about the x, y and z-axis, respectively (Stigson and Kullgren, 2017). Eighteen conventional helmets were tested on the CEN proposed standard in the 2015 Folksam report-the average rotational acceleration about the x, y, and z-axis were 6.4 krad/s^2, 6.7 krad/s^2 and 12.0 krad/s^2, respectively. Mean rotational velocity was 27.3 rad/s, 33.1 rad/s and 40.9 rad/s about the x, y and z-axis, respectively (Stigson and Kullgren, 2015). Both studies concluded that most helmets managed linear acceleration below 180g, however, all helmets needed improvement to reduce rotational energy.

2.6.2 Linear Acceleration

Linear acceleration was once the single metric used to analyze head impacts resulting in injury (Gurdjian et al., 1963; Holbourn, 1943). Linear acceleration correlates with the transient intracranial pressure within the brain after a direct impact to the skull (Gurdjian et al., 1968; King et al., 2003; Rowson and Duma, 2013). These pressure gradients are associated with focal head injuries such as a skull fracture (Holbourn, 1943; Gennarelli et al., 1972; King et al., 2003; Rowson and Duma, 2013). Gurdjian and colleagues (1963) investigated the mechanism and pathology of a concussion using mongrel dogs. Their research concluded that linear acceleration was a significant variable alongside rotational acceleration that contributed to brain injury.

Based on experiments involving air pressure applied to the dural sac of mongrel dogs, and cadaveric forehead impacts onto automotive instrument panels, it was demonstrated that intensity of intracranial pressure and duration of pressure imparted to the head were important to establish injury threshold (Gurdijian et al., 1955). These experiments led to the preliminary foundation of the Wayne State Tolerance Curve (WSTC). This curve describes the relationship between linear acceleration, duration of acceleration, and the onset of a skull fracture (Gurdijian et al., 1966; Versace et al., 1971; Eppinger et al., 1996). The onset of a skull fracture was thought to be associated with mild brain injuries, such as a concussion (Gurdijian et al., 1970). Additional tolerance curves such as the Gadd Severity Index (GSI) and Head Injury Criteria (HIC) were developed to determine the severity of an impact using linear acceleration (Versace et al., 1971; Gadd, 1996; Eppinger et al., 1996). These threshold curves became the foundation for standards to test helmets under linear conditions (Rowson and Duma, 2013). Research investigating the efficacy of cycling helmets determined that they effectively reduce the risk of sustaining skull fractures (Hodgson, 1990; Benz et al., 1993; Harrison et al., 1996; Thompson et al., 1996; Halldin, 2001; Scher, 2006; Mattei et al., 2012; Cripton et al., 2014; Stigson and Kullgren, 2015; Demarco et al., 2016). The 2015 & 2017 Folksam reports tested eleven and eighteen different models of cycling helmets using the CEN proposed standard at 6.0m/s, respectively (Stigson and Kullgren, 2015, 2017). All helmets were able to manage linear acceleration well below 250g at all impact locations (x, y, z).

2.6.3 Brain Tissue Deformation

Brain tissue deformation is a measure used to predict the risk of brain injury (King et al., 2003; Kimpara and Iwamoto, 2011; Post et al., 2013; Post et al., 2012a). Linear and rotational loading curves (dynamic head response) from physical impacts are used as inputs for the FEM to simulate and measure brain injury response (Kang et al., 1997; Zhang et al., 2001; Kleiven et al., 2002; Horgan and Gilchrist, 2003; King et al., 2003; Takhounts et al., 2003; Takhounts et al., 2008). The result of the FEM analysis is brain tissue deformation measured as MPS, CSDM, Von Mises strain and Maximum Axonal Strain (Willinger et al., 2015). MPS is the brain element experiencing maximum strain and has a greater correlation in predicting mTBI compared to dynamic head response (King et al., 2003; Willinger et al., 2003; Zhang et al., 2004; Kleiven, 2007). It is described as an elongation of the brain tissue relative to its original length through strain (Silva, 2006). Measured by the FEM, MPS is the highest strain value among the x, y and z axis of the brain model and is a common measure to characterize brain tissue deformation (Zhang et al., 2001; Kleiven et al., 2002; Willinger et al., 2003). Kleiven (2013) reported strain levels in the brain are greater in oblique impacts due to rotational acceleration using the Kungliga Tekniska Högskolan (KTH) brain model. Using the University College Dublin Brain Trauma Model (UCDBTM), Post and colleagues (2012b & 2013) reported rotational acceleration as highly correlated with MPS compared to linear acceleration in a variety of brain tissues and regions for impacts on ice hockey and American football helmets. A study on uniaxial stretching of squid axons, to evaluate the response of the nervous system to loading, suggested an MPS of approximately 10% as an axonal strain threshold for a concussion (Thibault, 1993). Although MPS is a common predictor of mTBIs such as a concussion, the measured strain value does not generalize to the global strain experienced by real brain tissue (Nahum et al., 1977; Hardy et al., 2001; Horgan and Gilchrist, 2003, 2004).

CSDM (%) is a brain tissue deformation variable as a result of the dynamic response loading curves into the FEM (Bandak, 1995; Willinger et al., 2015). It is described as the volume of brain tissue elements experiencing strain levels greater than a first principal strain threshold (Bandak, 1995; Takhounts et al., 2003). MPS is the prescribed threshold value, and the volume of all the brain elements above the MPS quantifies the CSDM (Bandak, 1995). A CSDM exceeding a 15% MPS threshold approximately predicts the risk of sustaining a DAI (Thibault, 1993). DAI is axonal strain and damage found in multiple foci throughout the brain, therefore, CSDM approximately correlates with predicting a DAI (Takhounts et al., 2003). Rotational acceleration and rotational velocity have been found to correlate well with CSDM compared to linear acceleration using the SIMon FEM (Takhounts et al., 2008). In the 2017 Folksam report, using the KTH model, average brain strain for the rotation at the Front_Y location was 19.5%, Lateral_x axis = 29.8% and Lateral_z axis = 26.6% (Stigson and Kullgren, 2017). The 2015 Folksam report, Front_Y= 35%, Lateral_x= 15% and Lateral_z axis = 44% were reported (Stigson and Kullgren, 2015).

CHAPTER 3: RESEARCH DESIGN

3.1 Research Question

Primary: Is there a main effect between the ALI protocol and the CEN proposed standard on dynamic head response and brain tissue deformation?

Secondary: Is there a main effect between the ABS-EPS helmet and PVC-PU helmet on dynamic head response and brain tissue deformation?

3.2 Objectives

1. Primary: To compare the dynamic head response and brain tissue deformation between the two oblique test protocols (ALI protocol and CEN-proposed standard) on each cycling helmet (ABS shell-EPS liner and PVC shell-PU liner).

2. Secondary: To compare the dynamic head response and brain tissue deformation between each cycling helmet (ABS shell-EPS liner and PVC shell-PU liner) on each oblique test protocol (ALI protocol and CEN-proposed standard).

3.3 Independent Variables (2x2x3):

a. Two Oblique Test Protocols:

1. ALI protocol
2. CEN proposed standard

b. Two Types of Cycling Helmets:

1. ABS shell-EPS liner
2. PVC shell-PU liner

c. Three Impact Locations:

1. Front_y axis
2. Lateral_x axis
3. Lateral_z axis

3.4 Dependent Variables:

a. Dynamic Head Response:

1. Peak Resultant Rotational Acceleration
2. Peak Resultant Rotational Velocity
3. Peak Resultant Linear Acceleration

b. Brain Tissue Deformation:

1. MPS (%)

2. CSDM (15% threshold)

3.5 Null Hypothesis

1. No significant three-way interaction effect between the oblique test protocols, helmet types and impact locations on dynamic head response (rotational acceleration, rotational velocity and linear acceleration).

2. No significant three-way interaction effect between the oblique test protocols, helmet types and impact locations on brain tissue deformation (MPS and CSDM-15%).

3. No significant two-way interaction effect between the oblique test protocols and impact locations on dynamic head response.

4. No significant two-way interaction effect between the oblique test protocols and impact locations on brain tissue deformation.

5. No significant two-way interaction effect between the oblique test protocols and helmet type on dynamic head response.

6. No significant two-way interaction effect between the oblique test protocols and helmet type on brain tissue deformation.

7. No significant main effect between the ALI protocol and CEN proposed standard on dynamic head response.

8. No significant main effect between the ALI protocol and CEN proposed standard on brain tissue deformation.

9. No significant main effect between the ABS shell-EPS liner and PVC shell-PU liner helmets on dynamic head response.

10. No significant main effect between the ABS shell-EPS liner and PVC shell-PU liner helmets on brain tissue deformation.

3.6 Experimental Hypothesis

1. It is hypothesized that the CEN-proposed standard will result in a greater dynamic head response than the ALI protocol on each cycling helmet (ABS shell-EPS liner and PVC shell-PU liner).

2. It is hypothesized that the CEN-proposed standard will result in a greater brain tissue deformation than the ALI protocol on each cycling helmet (ABS shell-EPS liner and PVC shell-PU liner).

3. It is hypothesized that the PVC shell-PU liner helmet will result in a greater dynamic head response than the ABS shell-EPS liner helmet on each oblique test protocol (ALI protocol and CEN-proposed standard).

4. It is hypothesized that the PVC shell-PU liner helmet will result in a greater brain tissue deformation than the ABS shell-EPS liner helmet on each oblique test protocol (ALI protocol and CEN-proposed standard).

CHAPTER 4: METHODOLOGY

4.1 Equipment

4.1.1 Angular Launched Impactor (ALI)

The ALI (Figure 1) was used to conduct the test protocol. The system comprises of a steel carriage to guide the headform at 30° (with respect to the horizontal plane). The steel carriage is passed through two wheels powered by an electric motor that generates the desired launching velocity. A 15° steel impact surface covered in 320 grit, 40 μm sandpaper (to represent asphalt) was located at the ending path of the system. Impact velocity was captured using a photoelectric time gate. The ALI was used in a previous study as a high-velocity impactor to reconstruct mixed martial art punching events (Kendall, 2016).

Figure 1. The angular launched impactor (ALI) angled at 30° with the helmeted (white) Hybrid III headform located at the top (left image) and bottom (right image) of the system. A 15° steel impact surface (anvil), covered with blue sandpaper (coefficient of friction =0.5), is located at the ending path of the system.

4.1.2 Monorail Drop Rig

A Cadex Monorail Drop Rig (Cadex Inc., St-Jean-Sur-Richelieu, QC) (Figure 2) was used to conduct the CEN proposed standard. The setup contains a halo support carriage which vertically guides the head freely for movement to occur in 6 degrees of freedom after impact onto a 45° steel anvil covered in 320 grit, 40 μm sandpaper. Impact velocity was captured using a photoelectric time gate located 0.02m above the steel anvil. The Cadex Monorail Drop Rig was used in recent studies to assess ice hockey goaltender masks, the performance of children and adult alpine ski helmets, reconstruct falling events, and to compare two anthropomorphic test devices (Koncan et al., 2015; Clark et al., 2017, Post et al., 2017, Koncan et al., 2018).

Figure 2. The Monorail Drop Rig with the helmeted (black) Hybrid III headform supported by the halo carriage. A 45° steel anvil is covered with blue sandpaper (coefficient of friction =0.5).

4.1.3 High-Speed Camera

A HighSpeed Imaging PCI-512 Fastcam camera (Photron USA Inc., San Diego, CA, USA) was used to ensure the correct location was impacted for both protocols, as well as to ensure an impact angle of 30° was achieved for the ALI protocol. All correct impacts were recorded.

4.1.4 Hybrid III HeadForm

A male 50th percentile Hybrid III headform (mass = 4.54 kg +/- 0.01kg; FTSS, Plymouth MI) containing an aluminum shell with a vinyl cover was used for both impact systems (Figure 3). A tri-axial accelerometer with three angular rate sensors (DTS 6DX0399 PRO MODEL) is mounted to the centre of gravity of the headform on a steel base plate (Diversified Technical Systems, Seal Beach, CA) (Figure 4). The sensors are connected to a DTS SLICE NANO modular data acquisition system (DAS) located superior to the tri-axial accelerometers and angular rate sensors. The DAS consists of one SLICE NANO battery stack, a SLICE NANO base +, and two

SLICE NANO bridges. The data acquired from the DAS were linear acceleration (g) and rotational velocity (rad/s) curves about the X, Y, and Z-axes of the headform. Acceleration signals were collected and sampled at 20, 000 Hz, this sampling rate was applied to gather sufficient data from each impact (SAE J211). A low pass filter of 300 Hz (CFC 180) was applied to the acceleration signals to obtain a clean rotational acceleration curve derived from rotational velocity. In a study that determined the effect of applying filters of different cut off frequencies in reconstructions of concussion events for American Football and Ice Hockey, Post et al (2016) reported a low pass filter of 300 Hz to have an insignificant effect on finite element strain despite 300 Hz having some effect on linear and rotational measures in some cases. The DTS SLICE DAS was recently used by Bonin and colleagues to determine the dynamic responses between cadaver heads and three headforms for motorcycle helmet impacts (2017).

Figure 3. The male 50^{th} percentile Hybrid III headform with the release attachment containing three aluminum pieces (neck).

Figure 4. The tri-axial accelerometers with the three angular rate sensors located at the centre of gravity of the headform.

The coordinate system of the headform is defined as the x-axis directed anterior-posterior with x positive directed anteriorly; the y-axis directed medial-lateral with y positive directed to the left of the headform; and the z-axis directed superior-inferior with z positive directed superiorly (Figure 5) (Hubbard and Mcleod, 1974). The approximate moment of inertia (I) of the Hybrid III headform with its release attachment (neck) are $I_X=3.34 \bullet 10^{-2} kg \bullet m^2$, $I_Y=4.09 \bullet 10^{-2} kg \bullet m^2$, and $I_Z=3.89 \bullet 10^{-2} kg \bullet m^2$ (Willinger and Baumgartner, 2001).

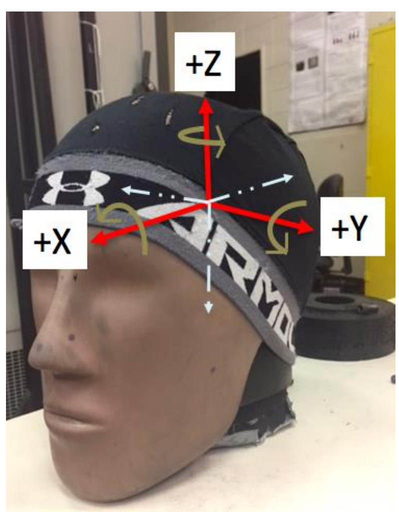


Figure 5. The coordinate system of the Hybrid III Headform (Hubbard and Mcleod, 1974).

4.1.5 Finite Element Model

The University College Dublin Brain Trauma Model (UCDBTM) was used to determine brain tissue deformation. The model was developed by performing computed tomography (CT) and magnetic resonance imaging (MRI) scans of a male cadaver head to create the geometry of the model, which enclosed 26,000 hexahedral elements. The dynamic head response time history curves were inputted to the UCDBTM to obtain MPS and CSDM-15%. The UCDBTM comprises of 10 components (the scalp, skull, pia, falx, tentorium, cerebrospinal fluid (CSF), grey and white matter, cerebellum and brainstem) with its material properties described in Tables 1 and 2 (Horgan and Gilchrist, 2003, 2004).

Table 1. Elastic material properties of the UCDBTM.

Material	Young's Modulus (MPa)	Poisson's Ratio	Density (kg/m^3)
Scalp	16.7	0.42	1000
Cortical bone	15000	0.22	2000
Trabecular bone	1000	0.24	1300
Dura	31.5	0.45	1130
Pia	11.5	0.45	1130
Falx and Tentorium	31.5	0.45	1130
CSF	15000	0.5	1000
Grey Matter	Viscoelastic	0.49	1040
White Matter	Viscoelastic	0.49	1040

Table 2. Viscoelastic material characteristics of the brain tissue for the UCDBTM.

	Shear modulus (kPa)		Decay Constant	Bulk modulus
	G_0	G_∞	(s^{-1})	GPa
White Matter	12.5	2.5	80	2.19
Grey Matter	10	2	80	2.19
Brain Stem	22.5	4.5	80	2.19
Cerebellum	10	2	80	2.19

The brain tissue was characterized as viscoelastic in shear with its brain behaviour represented by a linear viscoelastic model. The compressive nature of the brain tissue was defined as elastic. Shear characteristic of the viscoelastic brain was represented by the equation:

$$(1)\ \ G(t) = G_\infty + (G_0 - G_\infty)e^{-\beta t}$$

where G_∞, is defined as the long-term shear modulus, G_0, is the short-term shear modulus, and β is the decay factor (Horgan & Gilchrist, 2003).

4.1.6 The Bicycle Helmets

Six ABS shell-EPS liner and six PVC shell-PU liner bicycle helmets were used in this study. The ABS shell-EPS liner is an urban style helmet consisting of an ABS exterior shell and EPS interior absorption liner (Figure 6). The PVC shell-PU liner is also an urban style helmet consisting of a PVC exterior shell and PU interior absorption liner (Figure 6). Both helmet types had a dial adjustable comfort liner and retention straps. Helmet sizes were labelled as Large-ExtraLarge (57-60.5 cm) for both types. The ABS shell-EPS liner and PVC shell-PU liner weighed approximately 487g and 394g, respectively. The ABS shell and EPS liner thickness measured at 3.3 mm and 18.2 mm, respectively. The PVC shell and PU liner thickness measured at 0.9 mm and 21.3 mm, respectively.

Figure 6. Center image displays the exterior shells of the PVC shell-PU liner helmet (left) and the ABS shell-EPS liner (right). The far left and far right images display the interior design of the PU liner with the comfort foams, and EPS liner with the comfort foams, respectively.

4.2 Testing Protocols

The testing order for this study was as follows: Impacts were carried out with the ABS-EPS helmet first, and PVC-PU helmet second, at the Front_Y location on the ALI protocol. This was then repeated on the CEN proposed standard. Impacts were then carried out with the ABS-EPS helmet first, and PVC-PU helmet second, at the Lateral_X location on the ALI protocol. This was then repeated on the CEN proposed standard. Lastly, impacts were carried out with the ABS-EPS helmet first, and PVC-PU helmet second, at the Lateral_Z location on the ALI protocol. This was then repeated on the CEN proposed standard. Front_Y, Lateral_X and Lateral_Z was the impact order in this study and remained consistent for each helmet type. A randomize order was not used in this study in order to ensure consistent impact points on the helmets for each impact location. Each impact location has a specific neck orientation (release attachment) and in order to obtain a consistent point of impact per location the set up remained the same with the helmet only changed. A randomize order increases the variability in the location of the impact point by having to set the neck orientation for each impact. Similar test protocols describe a defined order of impact points when developing oblique testing protocols for cycling and motorcycle helmets such as crown, front, lateral/side and rear (Aare & Halldin et al., 2003; Meng et al., 2018).

Each helmet was impacted once at each location (Front_Y, Lateral_X, and Lateral_Z) for a total of 3 impacts per helmet. Three trials/impacts were collected per location (3), per helmet type (2) and per protocol (2), therefore, 6 helmets were used per protocol: three ABS-EPS and PVC-PU helmets for the ALI protocol and three ABS-EPS and PVC-PU helmets on the CEN proposed standard. A correct impact was defined as the helmeted headform impacting the desired impact location, impact velocity, and impact angle for each test protocol. Each impact was viewed and confirmed a correct impact using a High Speed Camera which recorded the impact (Section 4.1.3). Head impact velocity (6.5 m/s), head impact locations, and total head impact angle (45°) was consistent across both protocols. The inertia of the Hybrid III headform, and its release attachments (neck), was consistent between impact test protocols for each impact location.

4.2.1 Angular launched Impactor (ALI) Protocol

The helmeted headform was attached to a steel carriage, passed through two electrically motorized tires, guided on two steel rails through the centre of the ALI system, and impacted the surface. Dynamic response was collected at 20,000 Hz, filtered low pass at 300 Hz and stored for each impact on the DAS. A programmed speed of 6.5m/s was the impact velocity of the headform for the three impact locations: Front_Y, Lateral_X, Lateral_Z (Figure 7). The headform was launched at 30° (with respect to the horizontal plane) onto the 15° steel surface (total impact angle 45°) covered in sandpaper. Impacts began at the Front_Y location for all helmets, followed by Lateral_X and Lateral_Z. The torque applied to the headform at impact for the ALI protocol is displayed in Figure 9a.

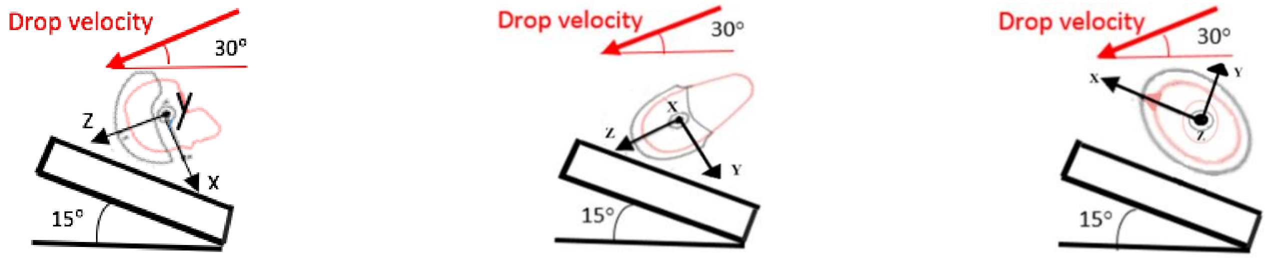


Figure 7. Impact locations Front_Y, Lateral_X and Lateral_Z (left to right) on the ALI (CEN, 2015).

4.2.2 CEN Proposed Standard

The helmeted headform was initially rested on a halo guided drop carriage on the monorail drop rig. The height of the drop carriage apparatus was elevated to a specific height to achieve a vertical impact velocity of 6.5m/s for all impacts at three locations: Front_Y axis, Lateral_X axis and Lateral_Z axis (Figure 8). Sandpaper was used to cover the 45° steel anvil impact surface. The torque applied to the headform at impact for the CEN proposed standard is displayed in Figure 9b.

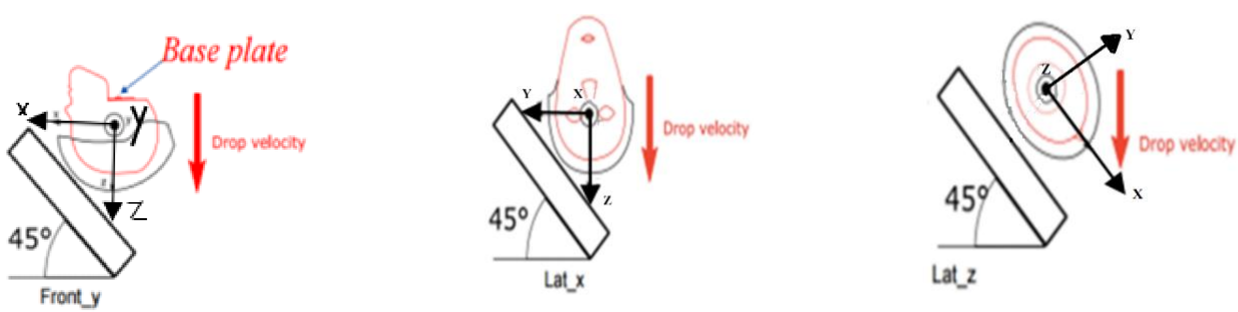
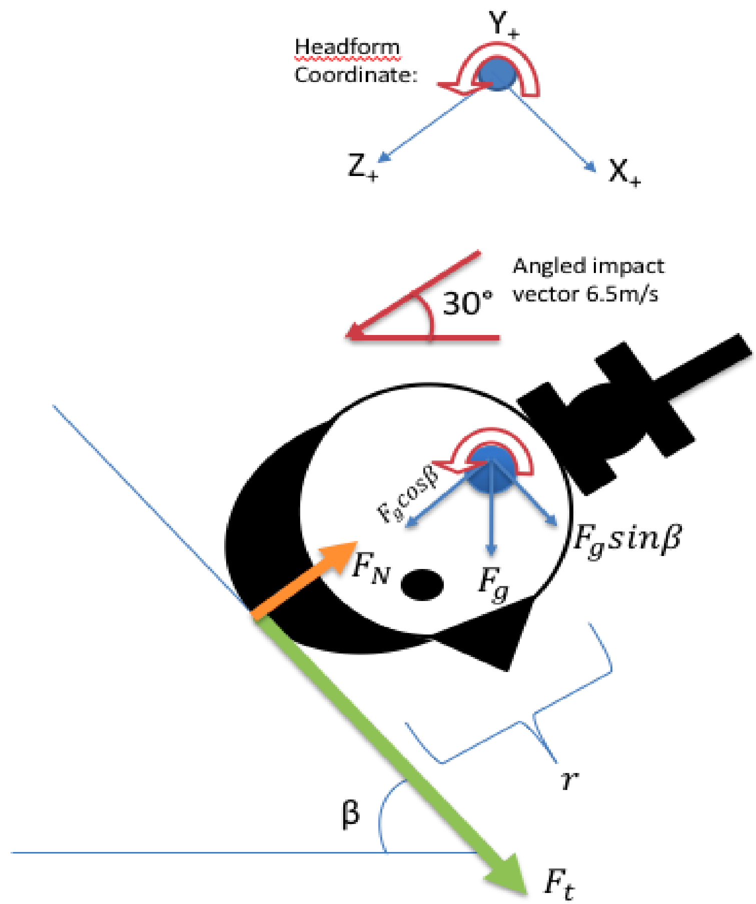


Figure 8. Impact locations Front_Y, Lateral_X and Lateral_Z (left to right) on the CEN proposed standard (CEN, 2015).



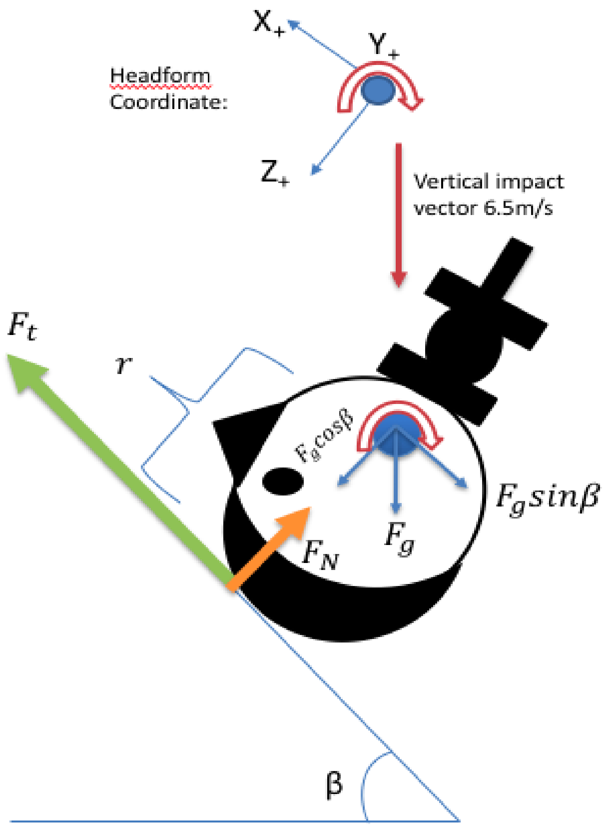
$$T_y = I_y \alpha_y$$

$$T_y = F_t \cdot r$$

$$T_y = (4.09 \cdot 10^{-2}\ kg \cdot m^2) \cdot (3495.1\ rad/s^2)$$
$$= 142.9\ \text{N} \cdot \text{m}$$

Where T_y = Torque applied by Anvil to Headform
I_y = Moment of Inertia of Headform + release attachement (neck)
α_y = Rotational acceleration of ALI-PVC2-Front_Y - T2 (Raw Data)
F_t = Tangential Force
r = distance from F_t to the axis of rotation Y+
β = 15°
F_N = Normal Force
F_g = Force of gravity at Center of Mass of Headform

Figure 9a. The Torque applied to the helmeted headform for the ALI protocol.



$$T_y = I_y \alpha_y$$

$$T_y = F_t \cdot r$$

$$T_y = (4.09 \cdot 10^{-2}\ kg \cdot m^2) \cdot (3789.4\ rad/s^2)$$
$$= 155.0\ \text{N} \cdot \text{m}$$

Where T_y = Torque applied by Anvil to Headform
I_y = Moment of Inertia of Headform + release attachement (neck)
α_y = Rotational acceleration of CEN-PVC1-Front_Y - T1 (Raw Data)
F_t = Tangential Force
r = distance from F_t to the axis of rotation Y+
β = 45°
F_N = Normal Force
F_g = Force of gravity at Center of Mass of Headform

Figure 9b. The Torque applied to the helmeted headform for the CEN proposed standard

4.3 Statistical/Data Analysis

The dependent variables of dynamic response and brain tissue deformation were reported as means and standard deviations. Each dependent variable (5) was tested on two oblique test protocols (2), for each helmet (2), at three impact locations (3). Therefore, a Three-Way ANOVA was the statistical test for each dependent variable. The test was used to determine if there was an interaction between the oblique test protocols, types of helmets and impact locations on the mean of each dependent variable. The alpha level was set at $p<0.05$ to establish significance. All data was analyzed using SPSS 21 for Windows (IBM Inc, Armonk, NY, USA). Along with a Three-Way ANOVA, a boxplot analysis was carried out to detect any outliers for each dependent variable. A Shapiro-Wilk test was used to assess for normality for each dependent variable. A Levene's test was carried out to test the variance of each dependent variable across all combinations of factors of the independent variables. These further assumption analyses were reported in the results section (Chapter 5). Although an ANOVA is fairly robust to violations of assumptions, a transformation was carried out and reported if these assumptions were violated in Appendix A.

A power analysis was performed using pilot data from equestrian helmeted impacts on the ALI protocol and CEN proposed standard. To achieve an 80% power ($\beta=0.2$), a computed number of impacts per location (N) for each oblique test protocol was N=1 for linear acceleration, rotational acceleration, rotational velocity, MPS, and CSDM-15% (DSS Researcher's Toolkit). Therefore, this study performed three impacts per impact location (3) for each helmet type (2) and for each oblique protocol (2).

4.4 Limitations

1. The male 50th percentile Hybrid III headform closely represents the geometry and mass of an adult head, however, it does not provide a true biofidelic response to impacts (Deng, 1989; Kendall et al., 2012). This is because the headform's composition of an aluminum skull cap and rubber skin material do not represent the true compliant nature of human skin tissue (Hodgson et al., 1971; Deng, 1989; Kendall et al., 2012)

2. The bicycle helmets in this study are limited to two different types of helmets. The results of this study cannot be generalized to other brands or types of helmets other than the ones used in this study.

3. The University College Dublin Brain Trauma Model (UCDBTM) will be used to determine the brain tissue strain from a head impact. This model was validated against cadaver head impact responses which may not result in an accurate prediction of the response of real human brain tissue (Nahum et al., 1977; Hardy et al., 2001; Horgan and Gilchrist, 2003, 2004).

4.5 Delimitations

1. The head impact velocity in this study was 6.5m/s. This velocity represents an average head impact velocity observed in real accident scenarios, which may not represent all the head impact velocities observed in real accident scenarios (Otte et al., 1999; Richter et al., 2007; Verschueren, 2009; Bourdet et al., 2012, 2014). However, it is the velocity used by the CEN proposed standard and was used in this study (CEN, 2015).

2. The head impact angle in this study was 45°. This angle represents an average head impact angle to the ground observed in real accident scenarios, which may not represent all the head impact angles observed in real accident scenarios (Ching et al., 1997; Aare & Halldin, 2003; Finan et al.,

2008; Kleivan, 2013). However, it is the head impact angle used by the CEN proposed standard and was used in this study (CEN, 2015).

3. A Hybrid III neck and a full body configuration was not attached to the Hybrid III headform, although studies suggest that the amplitude of the angular acceleration is affected by the neck and full body (COST 327, 2001; Beusenberg et al., 2001; Rueda, 2009; Verschueren, 2009; Ghajari et al., 2012). The CEN proposed standard does not use a neck or full body configuration, as studies show the effect of the neck is insignificant during the short time (5-10ms) the helmet comes into contact with the surface (Camacho et al., 2007; Ivancic, 2014; Willinger et al., 2015). Furthermore, the Hybrid III neck has been shown to be less human-like than numerical simulations without a neck (COST 327, 2015).

CHAPTER 5: RESULTS

The main objective of this study was to compare the dynamic head response and brain tissue deformation between the ALI protocol and CEN-proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets. The second objective was to compare the dynamic head response and brain tissue deformation between the ABS shell-EPS liner and PVC shell-PU liner cycling helmets on the ALI protocol and CEN proposed standard. A total of 35 impacts were conducted; 17 impacts on the ALI protocol, and 18 impacts on the CEN proposed standard. One data point collected on the ALI protocol with the ABS helmet at the Lateral_Z location was excluded, because, the impact was not considered a correct impact as described in Section 4.2. The results of peak resultant dynamic head response, consisting of rotational acceleration, rotational velocity and linear acceleration, and brain tissue deformation, consisting of MPS and CSDM-15% are presented in terms of means and standard deviations in this section.

5.1 Dynamic Response

5.1.1 Rotational Acceleration

Peak resultant rotational acceleration presented in terms of means and standard deviations in Table 3 and Figure 10.

Table 3. Mean peak resultant rotational acceleration and standard deviation (SD) for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets.

Helmet Type:	**ABS**	**PVC**	**ABS**	**PVC**	**ABS**	**PVC**
Location:	Front_Y		Lateral_X		Lateral_Z	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
ALI-Rotational Acceleration (rad/s^2)	5598.1 (1005.9)	3944.9 (554.7)	4424.6 (461.4)	3229.0 (111.1)	13487.7 (188.6)	13057.0 (1201.0)
Mean Difference (ABS-PVC)	1653.2		1195.6		430.7	
CEN-Rotational Acceleration (rad/s^2)	5786.9 (875.5)	4387.4 (264.5)	6112.0 (647.1)	5138.6 (1354.0)	16163.1 (1063.8)	13522.9 (592.9)
Mean Difference (ABS-PVC)	1399.5		973.4		2640.2	
Mean Difference (ALI-CEN)	-188.8	-442.5	-1687.4	-1909.6	-2675.5	-465.9

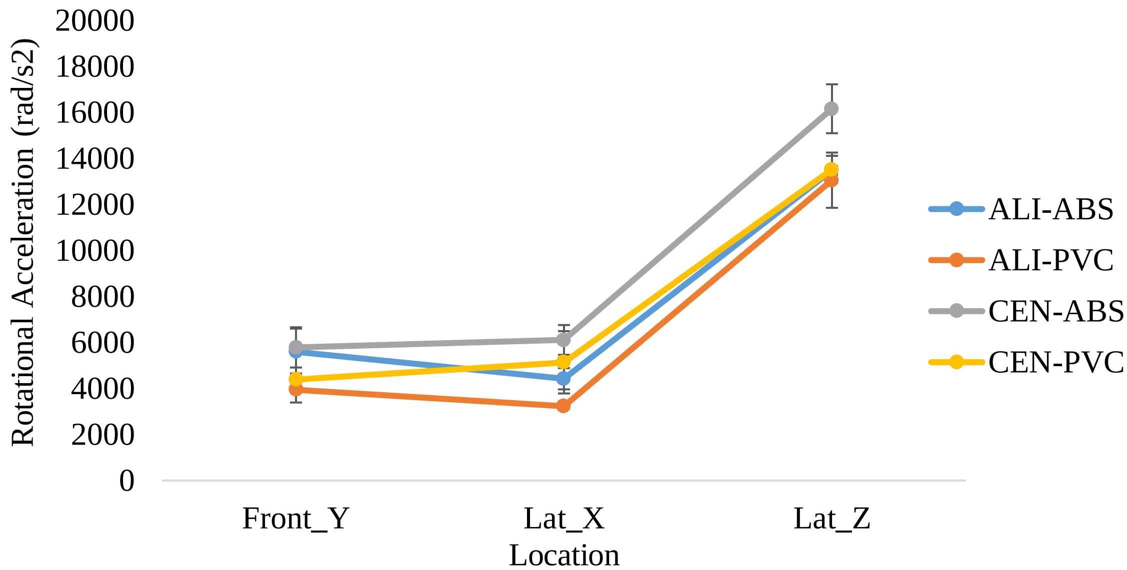


Figure 10. Mean peak resultant rotational acceleration and standard deviation for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets as a function of location.

Protocols and Helmet Types

A three-way ANOVA was conducted to determine the effects of protocol, helmet type, and location on rotational acceleration. There was no outlier as assessed through a box plot analysis. Rotational acceleration was normally distributed for all groups as assessed by Shapiro-Wilk's test of normality, $p>0.05$. There was homogeneity of variance as assessed by Levene's test for equality of variances, $p=0.226$. There was no statistically significant three-way interaction between protocol, helmet type, and location, $F(2,23)=2.1$, $p=0.145$. The two-way interaction effect between protocol and location on rotational acceleration was not statistically significant, $F(2,23)=2.9$, $p=0.077$. Therefore, an analysis of the main effect for protocol on rotational acceleration was performed, which indicated that the main effect was statistically significant, $F(1,23)=19.8$, $p<0.001$. All pairwise comparisons were run, reported 95% confidence intervals and p-values were Bonferroni-adjusted. The unweighted marginal means of rotational acceleration for the ALI protocol and CEN proposed standard were 7290.2 ± 199.1 rad/s^2 and 8518.5 ± 191.3 rad/s^2, respectively (Figure 11). The ALI was associated with a mean rotational acceleration of 1228.3 rad/s^2 (95% CI, 657.0 to 1799.5) lower than the CEN proposed method, a statistically significant difference, $p<0.001$. Therefore, the null hypothesis on the main effect of protocol for rotational acceleration was rejected.

The two-way interaction effect between helmet type and protocol was not statistically significant, $F(1,23)= 1.095$, $p=0.306$. The main effect for helmet type was statistically significant, $F=(1,23)= 19.8$, $p<0.001$. The unweighted marginal means of rotational acceleration for the ABS shell-EPS liner helmet and PVC shell-PU liner helmet was 8595.4 ± 199.1 rad/s^2 and 7213.3 ± 191.3 rad/s^2, respectively (Figure 12). The ABS shell-EPS liner helmet was associated with a mean rotational acceleration of 1382.1 rad/s^2 (95% CI, 810.9 to 1953.4) greater than the PVC shell-PU liner helmet, a statistically significant difference, $p<0.001$. Therefore, the null hypothesis on the main effect of helmet type for rotational acceleration was rejected.

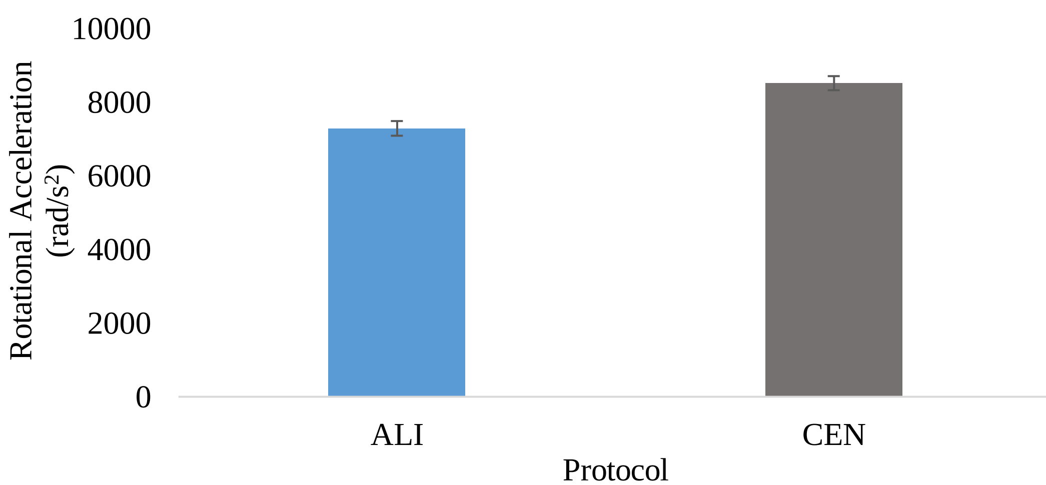


Figure 11. Unweighted marginal means of rotational acceleration between the protocols.

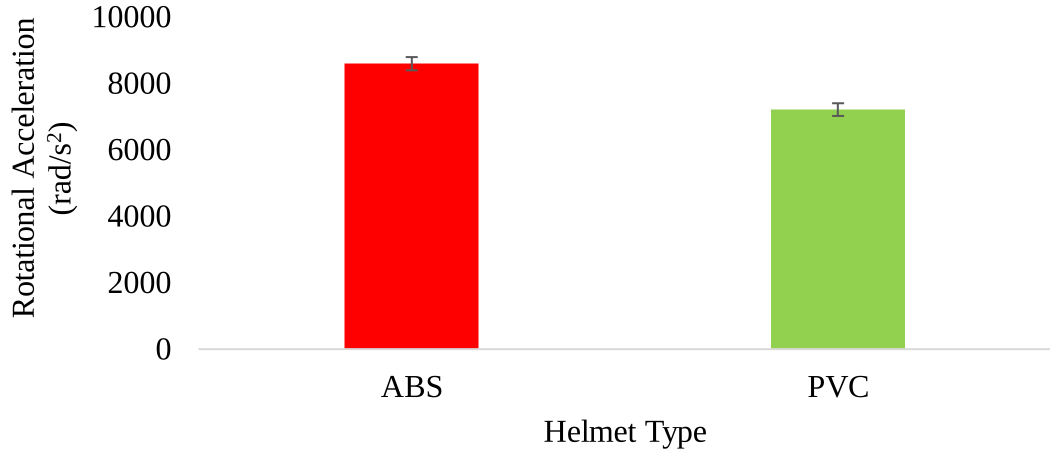


Figure 12. Unweighted marginal means of rotational acceleration between helmet types.

5.1.2 Rotational Velocity

Peak resultant rotational velocity are presented in terms of means and standard deviations in Table 4 and Figure 13.

Table 4. Mean peak resultant rotational velocity and standard deviation (SD) for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets.

Helmet Type:	**ABS**	**PVC**	**ABS**	**PVC**	**ABS**	**PVC**
Location:	Front_Y		Lateral_X		Lateral_Z	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
ALI-Rotational Velocity (rad/s)	21.9 (2.3)	20.3 (2.7)	15.8 (2.2)	14.7 (1.3)	40.0 (0.8)	42.6 (3.2)
Mean Difference (ABS-PVC)	1.6		1.1		-2.6	
CEN-Rotational Velocity (rad/s)	24.2 (6.6)	23.8 (2.3)	28.4 (1.2)	24.9 (2.8)	55.6 (0.9)	53.6 (2.3)
Mean Difference (ABS-PVC)	0.4		3.5		2.0	
Mean Difference (ALI-CEN)	-2.3	-3.9	-12.6	-10.2	-15.7	-10.9

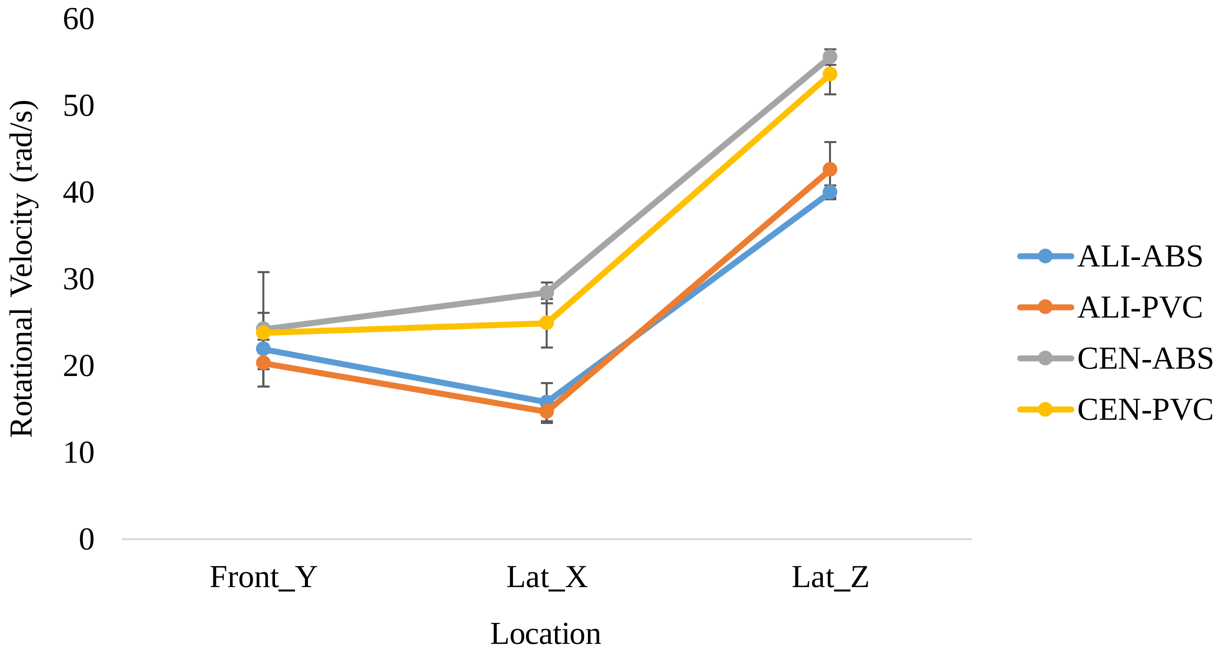


Figure 13. Mean peak resultant rotational velocity and standard deviation for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets as a function of location.

Protocols and Helmet Types

A three-way ANOVA was conducted to determine the effects of protocol, helmet type and location on rotational velocity. There was no outlier as assessed through a box plot analysis. Rotational velocity was normally distributed for all groups ($p>0.05$), except for the PVC-ALI-Front_Y condition ($p=0.035$) as assessed by Shapiro-Wilk's test. The test was run, as the ANOVA is fairly robust to deviations from normality (Laerd Statistics, 2015). Homogeneity of variance was violated as assessed by Levene's test for equality of variances, $p=0.046$. The ANOVA test was run because the number of impacts (N=3) between testing conditions were equal except for one condition (N=2). Therefore, the ANOVA is somewhat robust to heterogeneity of variance in these circumstances (Laerd Statistics, 2015). There was no statistically significant three-way interaction between protocol, helmet type, and location, $F(2,23)=0.754$, $p=0.482$. There was a statistically significant two-way interaction between protocol and location $F(2,23)=10.9$, $p<0.001$. Therefore, the null hypothesis on the two-way interaction between protocol and location was rejected for rotational velocity. Data are mean ± standard error, unless otherwise stated. The simple main effect of protocol on mean rotational velocity at the Lateral_X ($F(1,23)=47.6$, $p<0.001$) and Lateral_Z ($F(1,23=57.7$, $p<0.001$) location was statistically significant. All pairwise comparisons were made at the Lateral_X and Lateral_Z location with a Bonferroni adjustment. Rotational velocity at the Lateral_X location was 15.3 ± 1.2 rad/s (95% CI, 12.9 to 17.7) on the ALI protocol and 26.7 ± 1.2 rad/s (95% CI, 24.3 to 29.0) on the CEN proposed standard (Figure 14), a statistically significant difference of 11.4 rad/s (95% CI, 8.0 to 14.8), $p<0.001$. Rotational velocity at the Lateral_Z location was 41.3 ± 1.3 rad/s (95% CI, 38.6 to 44.0) on the ALI protocol and 54.6 ± 1.2 rad/s (95%CI, 52.2 to 57.0) on the CEN proposed standard (Figure 14), a statistically significant difference of 13.3 rad/s (95% CI, 9.7 to17.0), $p<0.001$.

The two-way interaction effect between helmet type and protocol was not statistically significant, $F(1, 23)= 1.072$, $p=0.311$. The main effect for helmet type was not statistically significant, $(F=1,23)= 1.0$, $p=0.319$. The unweighted marginal means of rotational velocity for the ABS shell-EPS liner helmet and PVC shell-PU liner helmet was 31.0 ± 0.7 rad/s and 30.0 ± 0.7 rad/s respectively (Figure 15). The ABS shell-EPS liner helmet was associated with a mean rotational velocity of 1.0 rad/s (95% CI, -1.0 to 3.0) greater than the PVC shell-PU liner helmet, not a statistically significant difference, $p=0.319$.

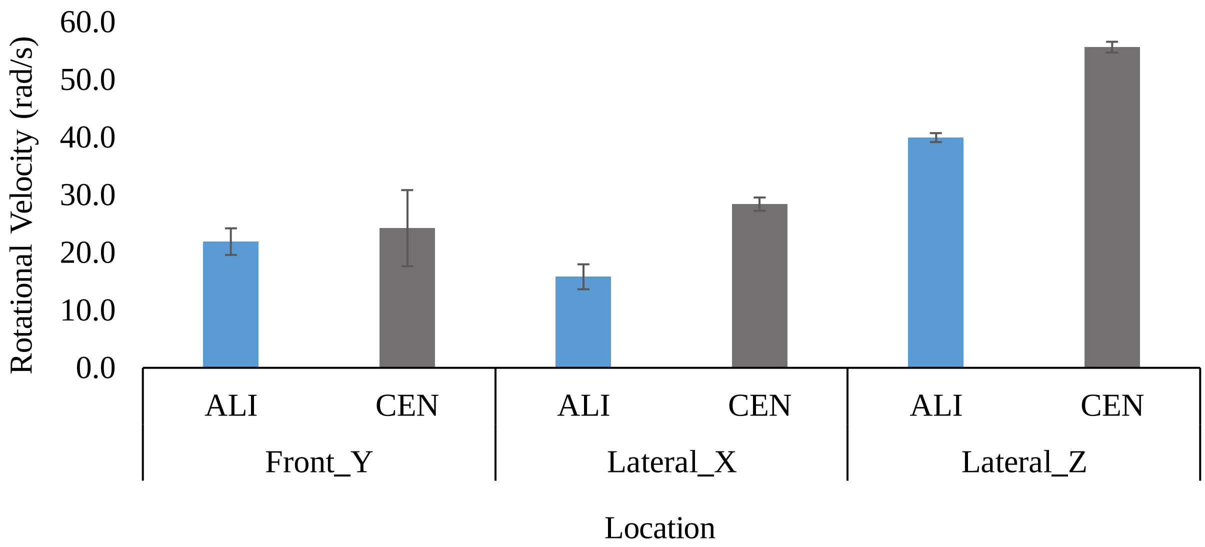


Figure 14. Means of rotational velocity between protocols at each location.

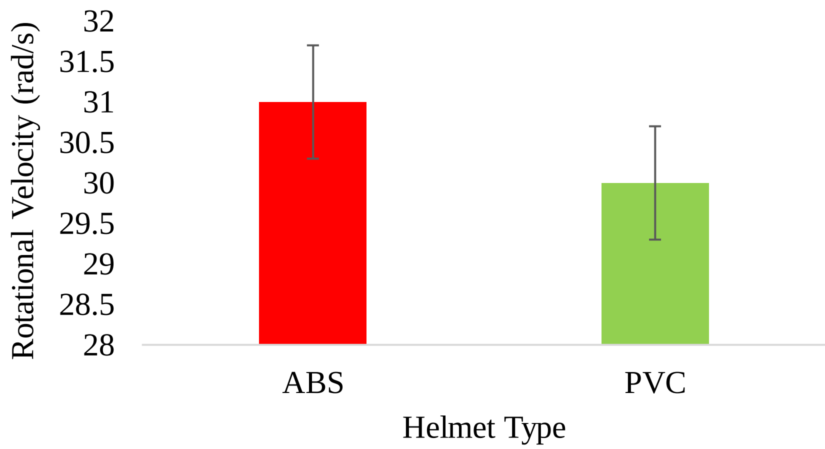


Figure 15. Unweighted marginal means of rotational velocity between protocols.

5.1.3 Linear Acceleration

Peak resultant linear acceleration are presented in terms of means and standard deviations in Table 5 and Figure 16.

Table 5. Mean peak resultant linear acceleration and standard deviation (SD) for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets.

Helmet Type:	**ABS**	**PVC**	**ABS**	**PVC**	**ABS**	**PVC**
Location:	Front_Y		Lateral_X		Lateral_Z	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
ALI-Linear Acceleration (g)	152.4 (20.2)	124.4 (6.0)	138.0 (2.2)	123.0 (4.3)	125.7 (6.2)	117.9 (5.2)
Mean Difference (ABS-PVC)	28.0		15		7.1	
CEN-Linear Acceleration (g)	131.4 (4.1)	109.6 (3.4)	124.0 (16.6)	116.6 (12.2)	110.6 (9.0)	103.1 (5.1)
Mean Difference (ABS-PVC)	21.8		7.4		7.5	
Mean Difference (ALI-CEN)	21.0	14.8	14.0	6.4	15.1	14.9

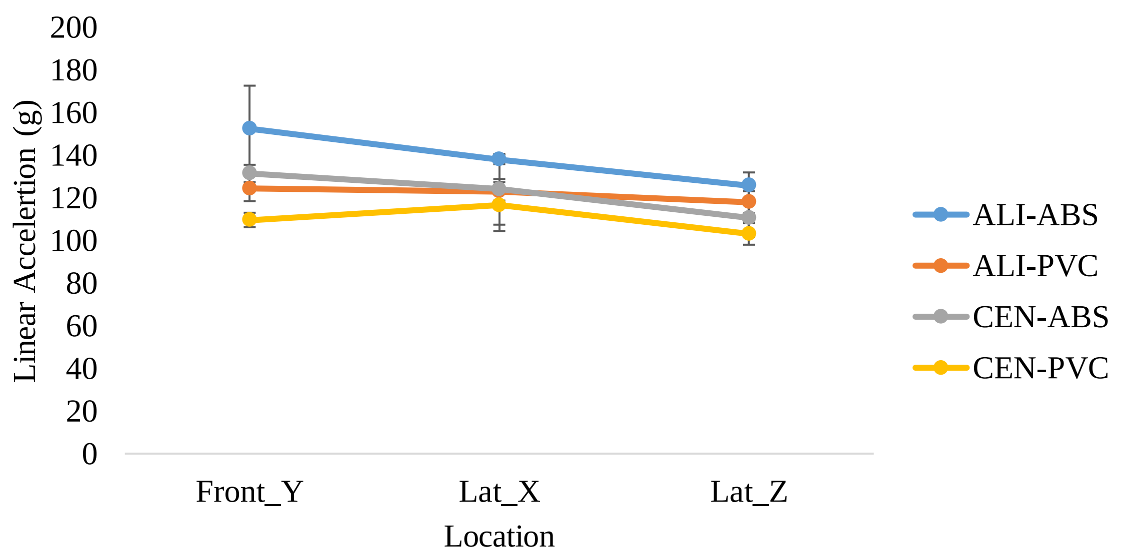


Figure 16. Mean peak linear acceleration and standard deviation for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets as a function of location.

Protocols and Helmet Types

A three-way ANOVA was conducted to determine the effects of protocol, helmet type, and location on linear acceleration. There was no outlier as assessed through a box plot analysis. Linear acceleration was normally distributed for all groups as assessed by Shapiro-Wilk's test of normality, $p>0.05$. Homogeneity of variance was violated as assessed by Levene's test for equality of variances, $p=0.025$. The ANOVA was run because the number of impacts (N=3) between testing conditions were equal except for one condition (N=2). Therefore, the ANOVA is somewhat robust to heterogeneity of variance in these circumstances (Laerd Statistics, 2015). There was no statistically significant three-way interaction between protocol, helmet type, and location, $F(2,23)=0.116$, $p=0.891$. The two-way interaction effect between protocol and location on linear acceleration was not statistically significant, $F(2,23)=0.485$, $p=0.622$. Therefore, an analysis of the main effect for protocol on linear acceleration was performed, which indicated that the main effect was statistically significant, $F(1,23)=19.1$, $p<0.001$. All pairwise comparisons were run, where reported 95% confidence intervals and p-values were Bonferroni-adjusted. The unweighted marginal means of linear acceleration for the ALI protocol and CEN proposed method were 130.2 ± 2.4g and 115.8 ± 2.3g respectively (Figure 17). The ALI protocol was associated with a mean linear acceleration of 14.4g (95% CI, 7.6 to 21.2) greater than the CEN proposed standard, a statistically significant difference, $p<0.001$. Therefore, the null hypothesis on the main effect of protocol for linear acceleration was rejected.

The two-way interaction effect between helmet type and protocol was not statistically significant, $F(1,23)= 1.095$, $p=0.306$. The main effect for helmet type on linear acceleration was statistically significant, $(F=1,23)= 19.8$, $p<0.001$. The unweighted marginal means of linear acceleration for the ABS shell-EPS liner helmet and PVC shell-PU liner helmet was 130.4 ± 2.4g and 115.8 ± 2.3g respectively (Figure 18). The ABS shell-EPS liner helmet was associated with a mean linear acceleration of 14.6g (95% CI, 7.8 to 21.4), greater than the PVC shell-PU liner helmet, a statistically significant difference, p<0.001. Therefore, the null hypothesis on the main effect of helmet type for linear acceleration was rejected.

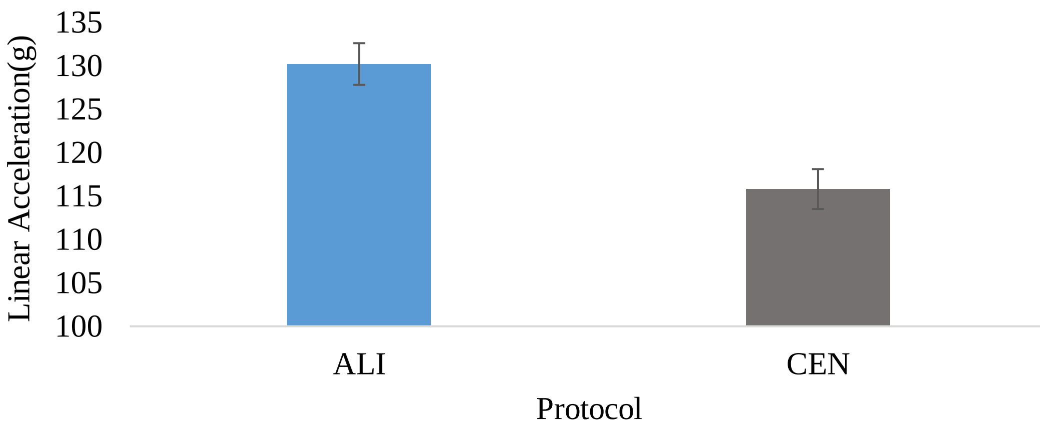


Figure 17. Unweighted marginal means of linear acceleration between protocols.

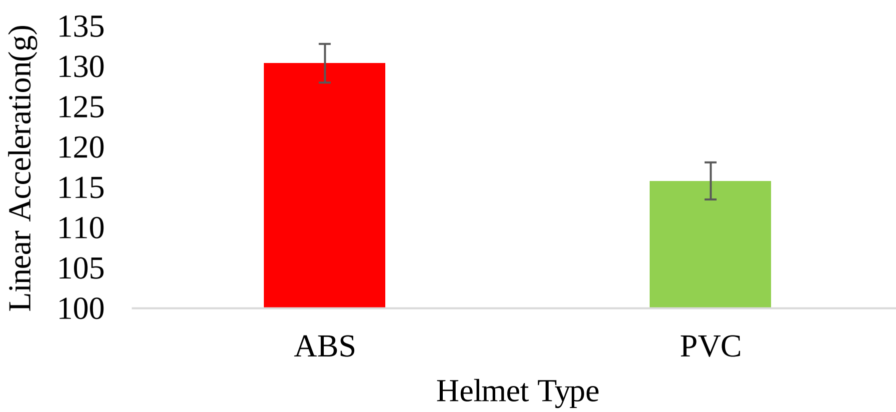


Figure 18. Unweighted marginal means of linear acceleration between helmets.

5.2 Brain Tissue Deformation

5.2.1 MPS

Peak resultant MPS are presented in terms of means and standard deviations in Table 6 and Figure 19.

Table 6. Mean peak resultant MPS % and standard deviation (SD) for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets.

Helmet Type:	**ABS**	**PVC**	**ABS**	**PVC**	**ABS**	**PVC**
Location:	Front_Y		Lateral_X		Lateral_Z	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
ALI-MPS (%)	30.3 (4.5)	27.2 (4.5)	26.2 (2.2)	24.4 (1.4)	41.3 (1.4)	46.3 (5.2)
Mean Difference (ABS-PVC)	3.1		1.8		-5.0	
CEN-MPS (%)	31.0 (7.6)	30.0 (3.0)	38.0 (3.4)	32.0 (5.6)	88.0 (2.8)	80.0 (4.6)
Mean Difference (ABS-PVC)	1.0		6.0		8.0	
Mean Difference (ALI-CEN)	-0.7	-3.7	-12.0	-7.7	-46.4	-33.1

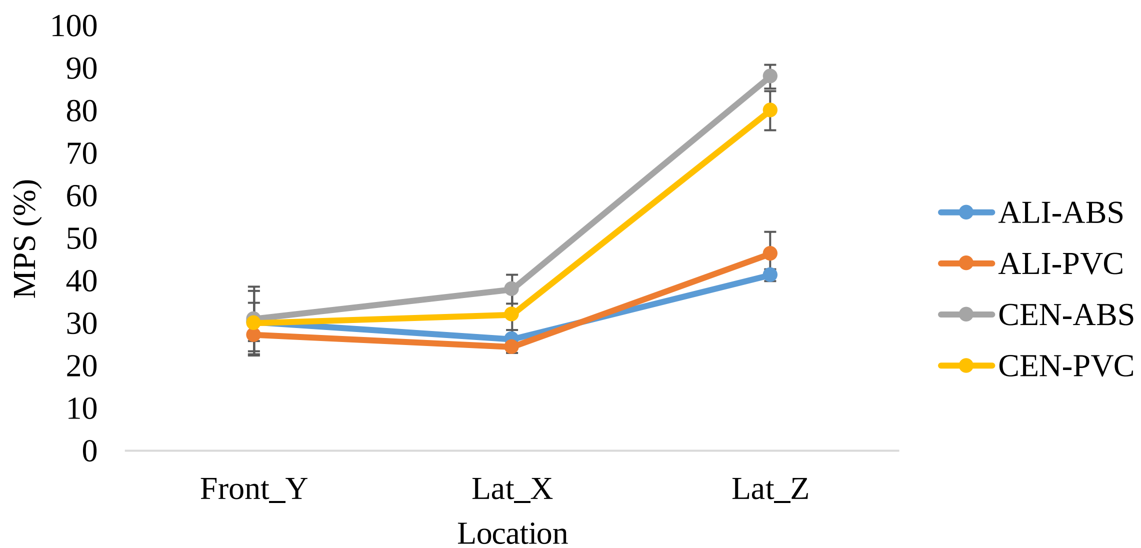


Figure 19. Mean peak MPS and standard deviation for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets as a function of location.

Protocols and Helmet Types

A three-way ANOVA was conducted to determine the effects of protocol, helmet type and location on MPS. There was no outlier as assessed through a box plot analysis. MPS was normally distributed for all groups as assessed by Shapiro-Wilk's test, $p>0.05$. Homogeneity of variance was assumed as assessed by Levene's test for equality of variances, $p=0.213$. There was no statistically significant three-way interaction between protocol, helmet type, and location, $F(2,23)=2.3$, $p=0.125$. There was a statistically significant two-way interaction between protocol and location $F(2,23)=58.7$, $p<0.001$. Therefore, the null hypothesis on the two-interaction between protocol and location was rejected for MPS. Data are mean ± standard error, unless otherwise stated. The simple main effect of protocol on mean MPS at the Lateral_X ($F(1,23)=15.8$, $p=0.001$) and Lateral_Z ($F(1,23)=224.8$, $p<0.001$) location was statistically significant. All pairwise comparisons were made at the Lateral_X and Lateral_Z location with a Bonferroni adjustment. MPS at the Lateral_X location was 25.3 ± 1.8 % (95% CI, 21.6 to 29.0) on the ALI protocol and 35.3 ± 1.8 % (95% CI, 31.6 to 39.0) on the CEN proposed standard (Figure 20), a statistically significant difference of 10.0 % (95% CI, 4.8 to 15.1), $p=0.001$. MPS at the Lateral_Z location was 43.8 ± 2.0 % (95% CI, 39.7 to 47.9) on the ALI protocol and 83.6 ± 1.8 % (95%CI, 79.9 to 87.2) on the CEN proposed standard (Figure 20), a statistically significant difference of 39.8 % (95% CI, 34.3 to 45.3), $p<0.001$.

The two-way interaction effect between helmet type and protocol was not statistically significant, $F(1, 23)= 2.9$, $p=0.105$. The main effect for helmet type on MPS was not statistically significant, $(F=1,23)= 2.8$, $p=0.108$. The unweighted marginal means of MPS for the ABS shell-EPS liner helmet and PVC shell-PU liner helmet was 42.5 ± 1.1 % and 40 ± 1.0 % respectively (Figure 21). The ABS shell-EPS liner helmet was associated with a mean MPS of 2.5 % (95% CI,

-0.6 to 5.5) greater than the PVC shell-PU liner helmet, not a statistically significant difference, p=0.108.

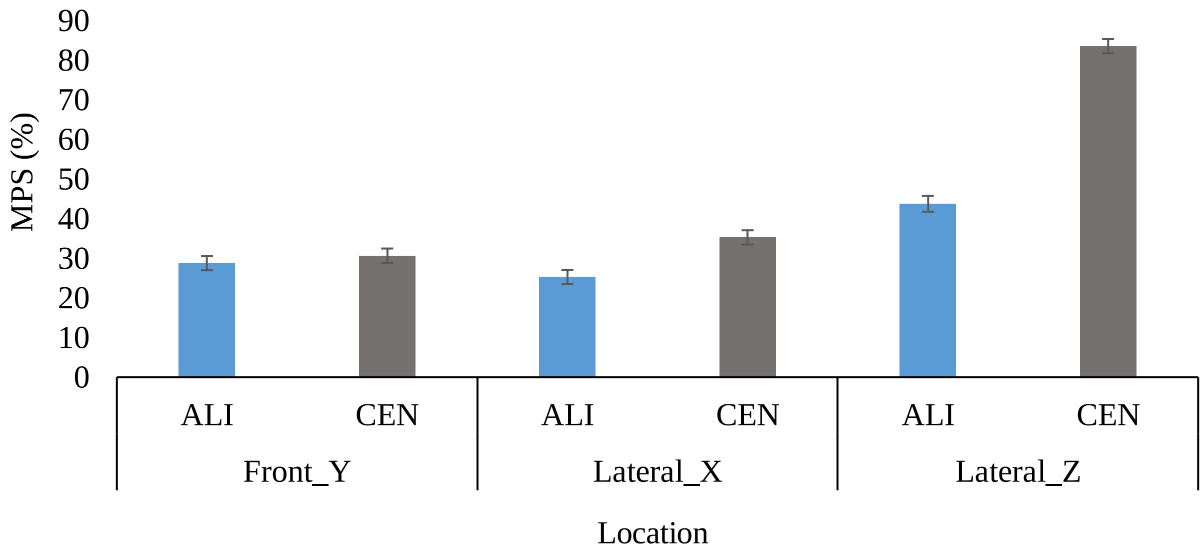


Figure 20. Means of MPS between protocols at each location.

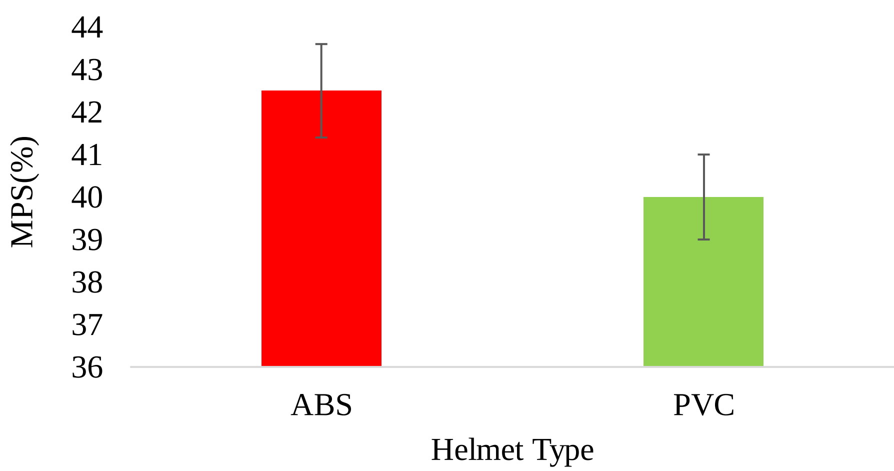


Figure 21. Unweighted marginal means of MPS between helmets.

5.2.2 CSDM-15%

CSDM are presented in terms of means and standard deviations in Tables 7 and Figure 22.

Table 7. Mean peak resultant CSDM-15% and standard deviation (SD) for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets.

Helmet Type:	**ABS**	**PVC**	**ABS**	**PVC**	**ABS**	**PVC**
Location:	Front_Y		Lateral_X		Lateral_Z	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)
ALI-CSDM-15% (%)	11.7 (6.4)	6.3 (4.0)	1.3 (0.6)	1.0 (0.0)	80.5 (3.5)	83.7 (5.0)
Mean Difference (ABS-PVC)	5.4		0.3		-3.2	
CEN-CSDM-15% (%)	18.7 (15.0)	11.0 (4.0)	19.7 (6.1)	13.3 (9.0)	93.7 (0.6)	90.3 (0.6)
Mean Difference (ABS-PVC)	7.7		6.4		3.4	
Mean Difference (ALI-CEN)	-7.0	-4.7	-18.3	-12.3	-13.2	-6.7

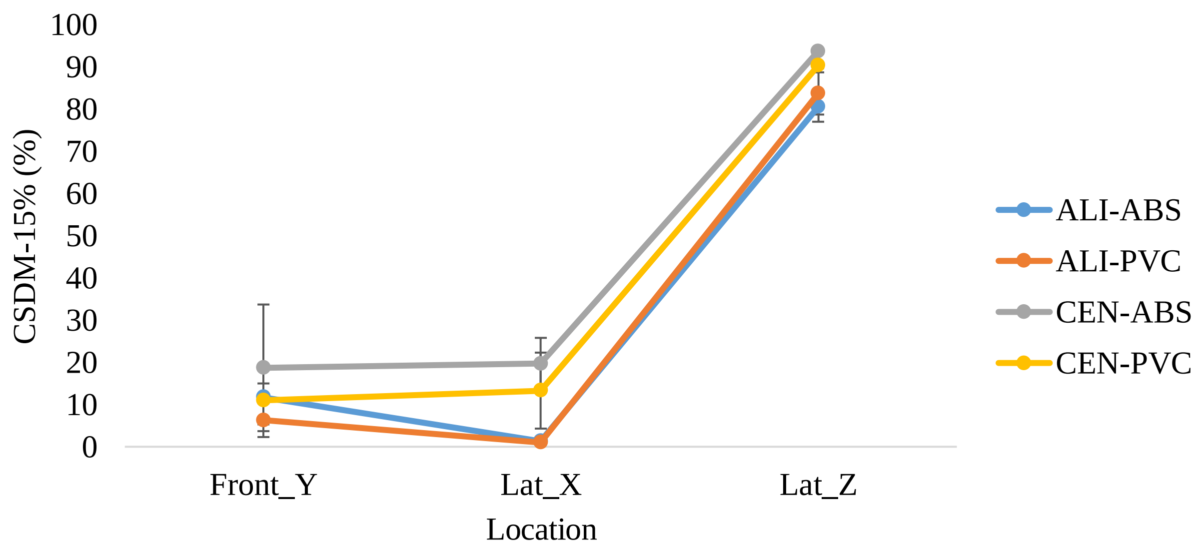


Figure 22. Mean peak CSDM-15% and standard deviation for the ALI protocol and CEN proposed standard on the ABS shell-EPS liner and PVC shell-PU liner cycling helmets as a function of location.

Protocols and Helmet Types

A three-way ANOVA was conducted to determine the effects of protocol, helmet type and location on CSDM-15%. There was no outlier as assessed through a box plot analysis. CSDM-15% was normally distributed for all groups ($p>0.05$), except for the ALI-ABS-Lateral_X, ALI-PVC-Front_Y, CEN-ABS-Lateral_Z and CEN-PVC-Lateral_Z conditions ($p<0.001$) as assessed by Shapiro-Wilk's test of normality. The test was run as the ANOVA is fairly robust to deviations from normality (Laerd Statistics, 2015). Homogeneity of variance was violated as assessed by Levene's test for equality of variances, $p=0.012$. The test was run because the number of impacts (N=3) between testing conditions were equal except for one condition (N=2). Therefore, the ANOVA is somewhat robust to heterogeneity of variance in these circumstances (Laerd Statistics, 2015). There was no statistically significant three-way interaction between protocol, helmet type, and location, $F(2,23)=0.097$, $p=0.908$. The two-way interaction effect between protocol and location on CSDM-15% was not statistically significant, $F(2,23)=1.7$, $p=0.197$. Therefore, an analysis of the main effect for protocol on CSDM-15% was performed, which indicated that the main effect was statistically significant, $F(1,23)=23.7$, $p<0.001$. All pairwise comparisons were run, where reported 95% confidence intervals and p-values were Bonferroni-adjusted. The unweighted marginal means of CSDM-15% for the ALI protocol and CEN proposed standard were $30.8 \pm 1.5\%$ and $41.1 \pm 1.5\%$ respectively (Figure 23). The ALI was associated with a mean CSDM-15% of 10.4 % (95% CI, 6.0 to 14.8) lower than the CEN proposed standard, a statistically significant difference, $p<0.001$. Therefore, the null hypothesis on the main effect of protocol for CSDM-15% was rejected.

The two-way interaction effect between helmet type and protocol was not statistically significant, $F(1, 23)= 1.4$, $p=0.257$. The main effect for helmet type on CSDM-15% was not statistically significant, $(F=1,23)= 2.4$, $p=0.134$. The unweighted marginal means of CSDM-15%

for the ABS shell-EPS liner helmet and PVC shell-PU liner helmet was 37.6 ± 1.5 % and 34.3 ± 1.5 % respectively (Figure 24). The ABS shell-EPS liner helmet was associated with a mean CSDM-15% of 3.3 % (95% CI, -1.2 to 7.7) greater than the PVC shell-PU liner helmet, not a statistically significant difference, p=0.134.

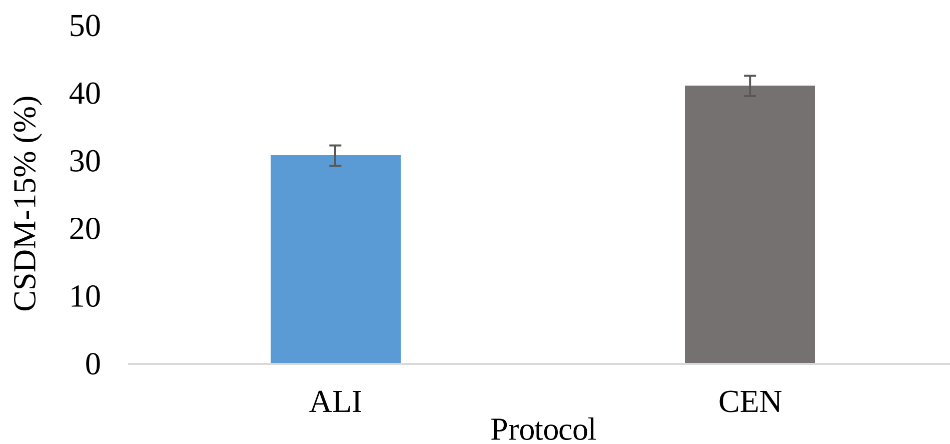


Figure 23. Unweighted marginal means of CSDM-15% between protocols.

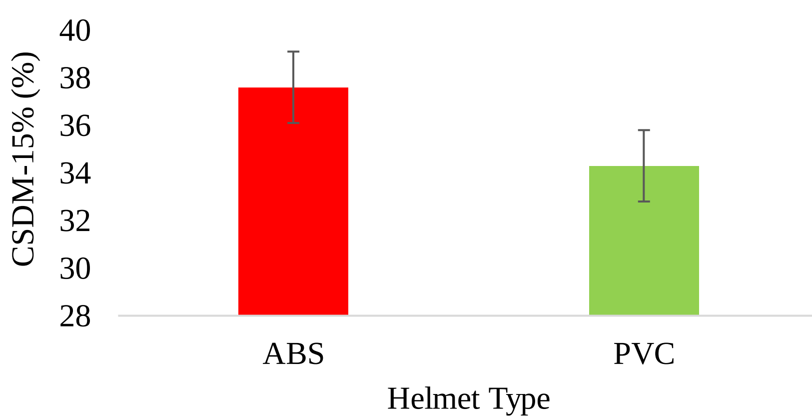


Figure 24. Unweighted marginal means of CSDM-15% between helmets.

CHAPTER 6: DISCUSSION

Oblique head impacts associated with rotational acceleration frequently occur in cycling-related head injuries (Otte et al., 1999; Mills & Gilchrist, 2008; Verschueren, 2009; Bourdet et al., 2012, 2014; McIntosh et al., 2013). Rotational acceleration causes strain to the brain tissue resulting in mild and severe brain injuries (Holborn, 1943; Aare et al., 2004; Kleiven, 2013; Post and Hoshizaki, 2015). Current bicycle helmet standards test the energy absorption of helmets under linear conditions, but not against rotational acceleration (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012). The purpose of this study was to compare the dynamic head response and brain tissue deformation between two oblique test protocols on two types of cycling helmets. As well as to compare the dynamic head response and brain tissue deformation between the two types of cycling helmets on each oblique test protocol.

6.1 Dynamic Response

6.1.1 Between Protocols

Rotational acceleration imparted by the CEN proposed standard was significantly greater than the ALI protocol. This effect was consistent across each helmet type and location. Differences in rotational response were expected due to the protocols unique impact vectors. The vertical impact vector illustrated on the CEN proposed standard may explain the greater rotational acceleration of the head, compared to the angled impact vector illustrated on the ALI protocol. Rotational acceleration (α) about an axis is directly proportional to torque (τ) about the same axis $(\tau = I \cdot \alpha)$. Since the moment of inertia (I) of the headform plus release attachment remained constant across protocols, the CEN proposed standard applied a greater torque to the head at impact compared to the ALI protocol (Figure 9a and 9b). Rotational velocity imparted on the CEN proposed standard was significantly greater than the ALI protocol at the Lateral_X and Lateral_Z locations. This difference was expected, because rotational acceleration was derived from rotational velocity in this study. The Front_Y impact may have been less sensitive to differences in rotational velocity between protocols, because of the impact occurring more dominantly through the center of gravity (McLean et al., 1997; Aare et al., 2004).

Linear acceleration imparted by the ALI protocol was significantly greater than the CEN proposed standard. This effect was consistent for each helmet type and location. The difference in impact vectors between protocols may also explain the differences in linear acceleration. Linear acceleration (a) through an axis is directly proportional to the net force (F) through the same axis ($F = m \cdot a$). Therefore, a greater force (F) was applied to the headform at impact from the ALI, as headform mass (m) remained constant between protocols. All impacts in this study managed linear acceleration below 180g since the protocols were not pure linear impacts. This is well below the 250-300g passing threshold for most cycling helmet standards (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012).

Despite the differences in dynamic response between the protocols, each protocol may represent the responses of unique falling events of cyclists in oblique head impacts (Verschuren, 2009; Bourdet et al., 2012, 2014). The ALI's angled impact vector may represent a cyclist falling in a horizontal motion, such as skidding or falling over the handlebars after hitting a curb or object (Verschuren, 2009; Bourdet et al., 2012, 2014). These types of falling events have been reported as real frequent-accident accident cases by Bourdet and colleagues (2012). The CEN proposed standard's vertical impact vector may represent a cyclist falling vertically onto a curb, motor vehicle, or angled surface to elicit rotational acceleration (Verschuren, 2009; Bourdet et al., 2012, 2014; CEN, 2015). Falling events associated with vertically impacting a curb, motor vehicle or angled surface resulted in a greater head rotation compared to the frequent-accident events of skidding or falling over the handlebars. These distinct head impact events resulted in oblique impacts that created different rotational head motion associated with a concussion and DAI (Klevian, 2013). Therefore, it is important for helmet standards to introduce oblique testing methods to measure the performance of cycling helmets under rotational acceleration (Halldin et al., 2001; Aare et al., 2003; Willinger et al., 2015).

6.1.2 Between Helmet Types

Linear and rotational acceleration was significantly less for the PVC shell-PU liner compared to the ABS shell-EPS liner. Differences in linear and rotational acceleration between helmet types can be the result of their different absorption liners and external shell materials. Thickness of shell and liner material between helmets may also be responsible for this difference. The ABS shell-EPS liner consisted of a thicker shell, however, the thicker absorption liner of the PVC-PU helmet may have mitigated linear and rotational acceleration at a greater extent compared to the ABS shell-EPS liner. This was consistent with a study comparing BMX-style helmets to traditional helmets, where an increased liner thickness resulted in lower peak headform accelerations (Demarco et al., 2016). Mills and Gilchrist (2006) also tested various bicycle helmet designs and reported an increased thickness with a lower density absorption foam reduces dynamic response. Both helmet types reported linear acceleration below 180g for all impacts, which is below the threshold standard and may reduce the risk of sustaining a skull fracture in oblique impact conditions (EN 1078, 1997; SNELL B95, 1998; CPSC, 1998; ASTM F1447-12, 2012, Mertz et al., 1997). This study observed cracking of both shell and liner on the PVC shell-PU liner helmet on both protocols (2 helmets on ALI protocol and 3 helmets on CEN proposed standard) at the Front_Y location (Appendix B). However, only one ABS shell-EPS liner helmet cracked on both the shell and liner on the ALI protocol. Demarco and colleagues (2016) also observed frequent cracking from the PVC-PU helmet compared to other helmets when tested under the CPSC standard.

Limited research has reported on the dynamic response and brain tissue deformation of cycling helmets of various shells and liners on oblique impacts (Hui and Yu, 2002; McIntosh et al., 2013; Demarco et al., 2016; Stigson et al., 2017). A study by Demarco and colleagues (2016) reported a linear acceleration of 345g for the PVC shell-PU liner helmet under the CPSC standard. This result exceeds the 300g threshold for the CPSC standard, however, their testing location occurred below the prescribed test line. While their study reported a poor performance of the PVC shell (PU liner) under linear impact conditions, this study suggests the helmet performed significantly better (in terms of linear acceleration and rotational acceleration) under oblique impact conditions compared to the ABS shell-EPS liner. The PVC helmet may protect poorly under linear impact conditions, but better under oblique impacts.

6.2 Brain Tissue Deformation

6.2.1 Between Protocols

Differences in brain tissue response may be the result of the unique head impact vectors between the oblique test protocols, therefore, different falling events are also sensitive to differences in strain and cumulative brain strain levels. MPS was significantly greater on the CEN proposed standard compared to the ALI protocol at the Lateral_X and Lateral_Z locations. CSDM-15% was also significantly greater on the CEN proposed standard compared to the ALI protocol for all locations-this effect was consistent for each helmet type. Falling events associated with vertically impacting a curb, motor vehicle or angled surface resulted in a greater brain tissue deformation compared to the frequent-accident events of skidding or falling over the handlebars (Bourdet et al., 2012). Dynamic response-time history curves were inputted into the FEM which may explain the parallel differences in MPS and dynamic head response between the protocols. The greater rotational motion of the brain elicited by the CEN proposed standard caused greater strain to the brain tissue which is characterized by its incompressible properties (Holbourn, 1943; Kleiven, 2013; Post and Hoshizaki, 2015). Due to the sensitivity of brain tissue to rotational loading (Kleiven, 2006), rotational acceleration has been found to be moderately correlated with MPS compared to linear acceleration (Rueda et al., 2011; Post et al., 2013).

Rotational velocity has also been found to be an appropriate measure to predict MPS, which may explain for statistical significance in MPS and rotational velocity at the Lateral_X and Lateral_Z locations (Aare et al., 2004; Takhounts et al., 2008; Takhounts et al., 2013; Ouckama & Pearsall, 2014). CSDM-15%, a variable to predict DAI, is the volume of brain tissue experiencing strain above 15% (Takhounts et al., 2013). Dynamic response-time history curves were also inputted to the FEM to determine CSDM-15%, therefore, differences in CSDM-15% between the protocols were expected (Takhounts et al., 2008). The results demonstrated both oblique protocols, each of which represent different falling events, are capable of producing different levels of CSDM-15% associated with DAI.

6.2.2 Between Helmet Types

There were no significant differences in brain tissue deformation between the two helmet types across protocol and locations. The results suggest that the helmet types are sensitive to changes in terms of linear and rotational head dynamics, however, the helmets performed equally in terms of brain tissue response. No differences at the tissue level, may indicate that regardless of the type of absorption liner and shell of the two helmets, the brain experiences similar tissue strain for the two oblique protocols. Rotational velocity has been found to correlate better with MPS and CSDM compared to rotational acceleration under certain impact conditions (Takhounts et al., 2013; Ouckama & Pearsall, 2014). Since there were no significant differences in rotational velocity between helmet types, this may explain the absence of differences between helmet types for MPS and CSDM-15%. This study is the first of its kind to report brain tissue response between two different types of helmets on two separate oblique test protocols. Further research is required to establish the brain tissue response of several types of helmets under oblique conditions.

6.3 Conclusion

The CEN proposed standard elicited greater rotational responses and brain tissue deformation than the ALI protocol for each helmet type. Therefore, the null hypothesis on the main effect of protocol for dynamic response and brain tissue deformation was rejected. Falling events associated with vertically impacting a curb, motor vehicle or angled surface resulted in a greater rotational response and brain tissue deformation compared to the frequent-accident events of skidding or falling over the handlebars. The distinct impact vectors associated with the unique falling events for each protocol resulted in significant differences in rotational head motion and brain tissue deformation. It was found that the PVC shell-PU liner helmet managed lower rotational and linear responses compared to the ABS shell-EPS liner on each oblique protocol. However, both helmets managed similar brain tissue responses. Therefore, the null hypothesis on the main effect of helmet type was rejected for linear and rotational acceleration but accepted for brain tissue deformation. The results suggest that helmet material, such as the type of shell and inner absorption liner, influences dynamic response and brain tissue deformation. Overall, this study informs helmet standards of the importance of incorporating oblique helmet testing to account for mild and severe brain injuries. As well as to consider impact vectors related to distinctive falling events to create unique rotational head responses. This study informs helmet users and manufactures on how two common types of helmets perform under frequent falling events associated with rotational conditions. The next step is to integrate rotational conditions into current helmet standards and designing helmets towards protection against rotational acceleration.

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APPENDIX A

Rotational Velocity

A transformation was applied, however, the ALI-PVC-Front_Y condition still violated normality, $p=0.039$. Homogeneity of variance remained violated, $p=0.034$. Three-way statistic, $p=0.626$, two-way statistic between protocol and location, $p<0.001$, main effect of protocol, $p<0.001$ and main effect of helmet type, $p<0.309$.

Linear Acceleration

A transformation was applied, however, homogeneity of variance remained violated, $p=0.036$. Three-way statistic, $p=0.886$, two-way statistic between protocol and location, $p=0.672$, main effect of protocol, $p<0.001$ and main effect of helmet type, $p<0.001$.

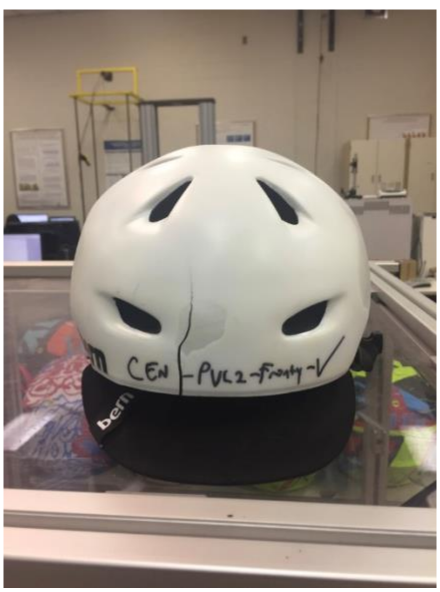
CSDM-15%

A transformation was applied, however, the ALI-ABS-Lateral_X, ALI-PVC-Front_Y, CEN-ABS-Lateral_Z and CEN-PVC-Lateral_Z conditions still violated normality, $p<0.001$ and homogeneity of variance, $p<0.001$. Three-way statistic, $p=0.672$ and two-way interaction between protocol and location, $p<0.001$. The CEN was significantly greater than the ALI at the Lateral_X location ($p<0.001$).

APPENDIX B

Helmet damage to the PVC shell-PU liner helmet on the ALI protocol at the Front_Y location:

Helmet damage to the PVC shell-PU liner helmet on the CEN proposed standard at the Front_Y location:



Helmet damage to the ABS shell-EPS liner helmet on the ALI protocol at the Front_Y location: