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**The Influence of Impact Mass, Inbound Velocity and System Compliance on the Dynamic Response
of a Hybrid III Head Form**

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**The Influence of Impact Mass, Inbound Velocity and System Compliance
on the Dynamic Response of a Hybrid III Head Form**

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Thesis submitted to the
Faculty of Graduate and Postdoctoral Studies
In partial fulfillment of the requirements
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ABSTRACT

Sport helmet certification standards solely prescribe the use of vertical drop towers in their protocols; a mechanism which mimics the head impacting the ground. Athletes are more commonly injured via collisions with other players; however, and the current certification standards do not incorporate this mechanism of injury in their test methodologies. At the most basic level, these two injury mechanisms can be differentiated by impact mass. The purpose of this study was to determine the effect of impact mass on the dynamic response of a Hybrid III head form. This was carried out by impacting the front location of the un-helmeted head form over a range of inbound velocities (2.0 m/s, 3.0 m/s and 4.0 m/s) and system compliances (rigid neck, compliant neck and unrestrained head and neck) on two different test systems represented by impact mass (monorail vertical drop tower and horizontal linear impactor) to more completely characterize the relationship. Significant main effects and interactions were observed for impact mass, inbound velocity and system compliance on peak resultant translational acceleration and peak resultant rotational acceleration ($p < 0.05$). Impacts on the monorail vertical drop tower (greater impact mass), resulted in higher acceleration values than the same impacts on the horizontal linear impactor, implying that the monorail generates more severe impacts. The greatest accelerations were also seen at the higher velocities, indicating that athletes impacted at higher velocities are at greater risk of suffering mild traumatic brain injury (mTBI). The relationship was non-linear for system compliance: the least compliant system generated the lowest translational accelerations, followed by the most compliant system, and ending with the intermediate level of system compliance. Rotational acceleration increased with increasing system compliance.

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CHAPTER 1. INTRODUCTION

Sports and recreational activities are common platforms for head and brain injury, affecting a reported 17,000 Canadians and as many as 3.8 million Americans per year (Canadian Institute for Health Information, 2006; Langlois, Rutland-Brown & Wald, 2006). These injuries range from mild to severe, and often have long-lasting and devastating consequences (Bayly, Cohen, Leister, Ajo, Leuthardt, & Genin, 2005; Levine, 1999). In 2005, researchers identified three effective ways of reducing the risk and incidence of head and brain injuries in sports: i) enforcing rules and regulations which control game play; ii) educating and training athletes in injury avoidance; and iii) developing personal protective equipment which lessens the amount of energy transferred to the head and brain (McIntosh & McCrory, 2005). The focus of this study is on the latter; specifically, the development of helmets.

Sport helmets were developed to mitigate the risk of suffering head and brain injury by dispersing impact forces away from the skull, and attenuating impact energies (Hoshizaki & Brien, 2004). Although they were initially designed to prevent skull fracture, their effectiveness in reducing traumatic injury to soft tissue has gained them popularity in many sports, including ice hockey and American football.

The design, protective capacity and certification of sport helmets is regulated by helmet safety standards which dictate specific impact test protocols to assess helmet performance (Hoshizaki & Brien, 2004). Current helmet designs are sport-specific, with the energy attenuation abilities of the helmet based on impacts common to the sport (Hoshizaki & Brien, 2004). Helmet manufacturers are well aware that their helmets must pass safety standards in order to achieve certification and be marketed to the public. As a

result, if the standards are not incorporating the impact energy ranges inherent to each sport, helmets will continue to be produced that do not necessarily function optimally for their specific activity, but rather that function in a manner that will allow them to be certified (Gilchrist & Mills, 1994). This could have a direct and detrimental effect on athletes wearing helmets which were not designed for the unique behaviours of their sport.

Considerable advancements have been made in the knowledge of brain injury mechanisms over the last few decades however the safety standards themselves have not undergone much change to incorporate this updated information (Halstead, Alexander, Cook, & Drew, 2000). Understanding the characteristics of the various injury mechanisms in sports, and integrating this knowledge in the standards, is crucial if helmets are to be optimized for their intended use.

1.1 PROBLEM STATEMENT

All helmet certification standards currently in use solely prescribe guided freefall apparatuses to test and certify helmets. These drop towers simulate fall-type injuries wherein an athlete's head impacts the ground or other such playing surface. Although this is a common mechanism of brain injury in sport, it is by no means the only way in which athletes are injured. In ice hockey, for example, the principal cause of concussions has been found by numerous researchers to be from collisions with other players (Honey, 1998; Azuelos *et al.*, 2004; Flik *et al.*, 2005). Similar findings have been reported for American football (Guskiewicz *et al.*, 2000). In a study examining concussions in the National Football League (NFL), Pellman and colleagues investigated 174 severe head impacts and concussive injuries, finding that 84% of the cases involved collisions between players, while ground impacts made up the other 16% (Pellman, Viano, Tucker, Casson, &

Waeckerle, 2003). There is a disconnect between what takes place in sport and what is tested in the industry as current sport helmet safety standards fail to account for this second mechanism of head injury commonly encountered by athletes. The dynamics of a head hitting the ground and a head being impacted by another object are fundamentally different and must be considered for proper head protection (Patrick, 1966). Thus, neglecting to address the different injury mechanisms in sports may result in the production of sport helmets that lack essential protective qualities.

1.2 OVERVIEW

Sports-related head and brain injuries are typically the result of two fundamental mechanisms; the first (fall-type) involves a moving head impacting the ground; a stationary surface of large mass, while the second (blow- or collision-type) involves the head being impacted by a moving object of considerably smaller mass, such as another player or object (Yanagida, Fujiwara, & Mizoi, 1989; Honey, 1998; Guskiewicz, Weaver, Padua, & Garrett, 2000; Azuelos, Pearsall, Turcotte, & Montgomery, 2004; Flik, Lyman, & Marx, 2005). While there are a number of variables which characterize these two mechanisms of injury, at the most basic level and for the purpose of this study they are distinguished by mass. For the fall-type impacts, the mass is the area of the ground affected by the impact (dictated by the standards to be no less than 136.1 kg), while for the collision-type impacts, it is the effective mass of the striking object (i.e., another player, stick or other such piece of sporting equipment).

Recently, the National Operating Committee on Standards for Athletic Equipment (NOCSAE) suggested introducing a horizontal linear impactor apparatus into their football helmet standard which is designed to simulate a collision-type impact (NOCSAE, 2006;

Pellman, Viano, Withnall, Shewchenko, Bir, & Halstead, 2006). Although the idea of this novel impact system is held in high regard, little work has been undertaken to characterize the similarities and differences between the new horizontal systems and the conventional vertical drop systems. Many of the past research studies attempting to characterize impact test systems highlighted the severity of drop tests. For instance, research conducted by Gurdjian, Roberts & Thomas (1966) revealed that the energy absorbed by the cadaveric head was maximized in the drop test due to the nature of the immovable impacting surface.

A study by Thom and colleagues (1998) looked at the differences between two drop systems (monorail guided drop and free drop) in the context of motorcycle helmets. They found that the monorail drop systems with restrained head forms consistently generated more severe impacts than the systems that were not guided or that did not have a restrained head form.

In 2000, Pearsall and colleagues compared four international safety standards for ice hockey helmets at each of six prescribed impact locations, using two different impact apparatuses (monorail and guided-wire), on four of the leading ice hockey helmets (Pearsall, Wall & Hoshizaki, 2000). An evaluation of the performance characteristics of the helmets was given in terms of translational acceleration (peak g), and no significant differences were found between test standards. The European standard generated the highest translational accelerations at four of the six impact sites, implying that it is harsher than the North American standards; however, it was concluded that no one standard could be labeled as more or less severe since the interaction produced mixed effects.

Though informative, these past studies neglected to consider the less severe yet more common collision-type impacts seen in sports and reconstructed by horizontal linear

impactors. Overlooking this injury mechanism when designing new sport helmets may actually be placing athletes at risk of suffering minor traumatic brain injury.

In 2003, Pellman and colleagues investigated head impacts in professional football, reconstructing both fall-type and collision-type impacts from video data. A custom guided freefall apparatus was used for each condition, with different impacting surfaces; a simulated, large mass ground consisting of artificial turf and a backing pad resting atop a platform was used in the helmet-to-ground case, while a second helmeted head-neck assembly, this time attached to a freely suspended Hybrid III torso and pelvis (total mass = 46.4 kg), was employed in the helmet-to-helmet trials. To better represent the helmet-to-helmet impacts, additional work was undertaken to create a new linear impact system which would remove the vertical force vector. Unfortunately, no comparisons were made between the helmet-to-ground data and the linear helmet-to-helmet impact data (Pellman, Viano, Withnall, Shewchenko, Bir, & Halstead, 2006).

Understanding the relationship between head injury mechanisms and test apparatus characteristics is imperative, as it gives equipment manufacturers and standard organizations knowledge to improve the development of safety helmets in sports. This study investigated the effects of impact mass, inbound velocity and system compliance on the dynamic response of a human head surrogate; the Hybrid III head form. Two impact test apparatuses used in the development and testing of sports helmets represented the two impact masses. The first system, a vertical drop tower, mimicked the head impacting an immovable object such as the ground, while the second system, a horizontal linear impactor, was representative of a much smaller mass impacting a static head (e.g., collision between two players). The dynamic response was measured in terms of two known

predictors of brain injury: peak translational acceleration and peak rotational acceleration. These response characteristics were examined over a range of inbound velocities and system compliances to assess whether the effects of impact mass varied across different conditions.

1.3 QUESTION

Is there a significant difference in the dynamic response characteristics (in terms of peak resultant translational and rotational acceleration) of a Hybrid III head form impacted with different masses over a range of inbound velocities and system compliances?

1.4 OBJECTIVES

This study was undertaken:

- i. to determine the effects of **impact mass** on the dynamic response characteristics (peak resultant translational acceleration & peak resultant rotational acceleration) of a Hybrid III head form.
- ii. to determine the effects of **impact mass** on the dynamic response characteristics (peak resultant translational acceleration & peak resultant rotational acceleration) of a Hybrid III head form over a range of *inbound velocities*.
- iii. to determine the effects of **impact mass** on the dynamic response characteristics (peak resultant translational acceleration & peak resultant rotational acceleration) of a Hybrid III head form over a range of system compliances.

1.5 NULL HYPOTHESES

- i. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 2.0 m/s.
- ii. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 3.0 m/s.
- iii. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 4.0 m/s.
- iv. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 2.0 m/s.
- v. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 3.0 m/s.
- vi. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the compliant neck at an *inbound velocity* of 4.0 m/s.
- vii. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 2.0 m/s.

- viii. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 3.0 m/s.
- ix. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 4.0 m/s.
- x. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 2.0 m/s.
- xi. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 3.0 m/s.
- xii. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the rigid neck at an *inbound velocity* of 4.0 m/s.
- xiii. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 2.0 m/s.
- xiv. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 3.0 m/s.

- xv. There will be no significant difference in the *peak resultant translational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 4.0 m/s.
- xvi. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 2.0 m/s.
- xvii. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg mass, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 3.0 m/s.
- xviii. There will be no significant difference in the *peak resultant rotational acceleration* of the head form for the two **impact masses** (158.8 kg mass, 17.1 kg) with the unrestrained head and neck at an *inbound velocity* of 4.0 m/s.

1.6 INDEPENDENT VARIABLES

1. **Impact mass** (kg): 158.8 kg, 17.1 kg
2. *Inbound velocity* (m/s): 2.0 m/s, 3.0 m/s, 4.0 m/s
3. System compliance (modified via changes to neck compliance): Rigid neck, compliant neck, unrestrained head and neck

The independent variables in this study were impact mass, inbound velocity and system compliance. In 2008, Gimbel demonstrated that mass has a significant effect on peak translational acceleration. Although there are multiple variables which distinguish

fall-type and collision-type impacts, at the most fundamental level and for the purpose of this study, the difference between them is mass. In the former condition, the moving head contacts a stationary surface of large mass (i.e., the mass of the area of the ground affected by the impact), whereas in the latter, a moving object of smaller mass (e.g., the effective mass of an incoming player) collides with the head (in this case, a stationary head was considered).

In this study, the mass was varied by employing different impact test systems which mimicked the two injury mechanisms. A vertical drop system with an impact mass of 158.8 kg represented the large stationary surface, while a horizontal linear impactor with an impact mass of 17.1 kg represented the effective mass of an opposing player. In helmet safety standards, a mass of at least 136.1 kg is required for vertical drop tests, while a greater mass is often preferred. Accordingly, an impact pedestal with a mass of 158.8 kg was chosen.

In collisions between players, athletes will often instinctively tense their neck muscles, increasing their effective mass and resulting in a decrease in the transferred acceleration (Reid, Epstein, Louis & Reid, 1975; Aubrey, Cantu, Dvorak, Graf-Baumann, Johnston, Kelly, et al., 2001). The mass of 17.1 kg was chosen based on the NFL studies carried out by Viano and Pellman (2005) as well as the linear impactor standard proposed by the NOCSAE (2006). These researchers designed the system to match the kinematics of impacts reconstructed from game-play video data in the NFL.

Inbound velocity and system compliance are also important variables because the level of neck compliance in the system influences the dynamic response of a Hybrid III head form depending on the velocity of the impactor (Rousseau, 2008). Although it is

known that a relationship exists, it has yet to be characterized. Logic dictates that since there is a difference in the dynamic response of the head form under various conditions of compliance and velocity without regard for impact mass, there will likely also be a difference for the two impact masses under the same conditions. A goal of this study was to examine this relationship.

The levels of inbound velocity (2.0 m/s, 3.0 m/s and 4.0 m/s) were chosen because these values are representative of impact velocities seen in sports and they are within the range of velocities historically used in the safety standards (Sim & Chao, 1978; Sim, Simonet, Melton, & Lehn, 1987). Gimbel (2008) recently found that a difference in energy induced by a variety of head form masses and inbound velocities significantly affected the performance of various helmet materials, which ultimately affected the wearer of the helmet. Additional research demonstrated that increasing inbound velocity resulted in increased peak translational and rotational accelerations (Gimbel, 2008; Rousseau, 2008). When considering drop tests, velocity is often controlled via modifications to drop height, as was the case in this study. In terms of horizontal linear impactors, changes to air pressure on a pneumatic system or in draw length on a spring system dictate velocity; the latter was modified.

Although the standards describe specific protocols in terms of test apparatus, impacting surface, head form mass, impact location and drop height (or inbound velocity), control over other variables such as system compliance is also possible on most of the test systems. The amount of compliance in a system directly affects the severity of the test. For example, a helmet mitigates the risk of suffering brain injury by introducing compliance into the system which aids in managing impact forces which would otherwise

be transmitted to the skull and its contents. System compliance can be modified in a number of ways, such as varying the impacting surface (M.E.P. pad vs. steel anvil), changing the head form (NOCSAE vs. vinyl-coated vs. metal alloy), introducing an energy attenuation system (helmet vs. no helmet), and employing different necks (unrestrained head and neck vs. compliant neck vs. rigid neck). This study focused on the latter: modifications to neck compliance.

Various helmet performance standards demand different amounts of compliance in the test systems. At one end of the continuum, there are standards which employ extremely rigid systems using a metal head form attached to a rigid neck impacting a steel surface, while at the other end, there are standards which incorporate the use of a more biofidelic head form impacting a rubber surface. The three levels of system compliance (rigid neck, compliant neck, unrestrained head and neck) were chosen based on the neck compliances used in the current safety standards. Many standards exist for the various sport helmets available to consumers, and each of these individual standards prescribes unique impact specifications (Pearsall, Wall, & Hoshizaki, 2000). Table 1 outlines the impact specifications of the four major hockey helmet certification standards: The American Society for Testing and Materials (ASTM), Committee European de Normalization (CEN), Canadian Standards Association (CSA) and International Organization for Standards (ISO). The differences in the safety standard specifications within the sport of ice hockey are evident, thus the differences that exist between the helmet standards of other sports can be easily inferred.

Table 1. Impact specifications of four hockey helmet certification standards.

	ASTM (F 1045-90a)	CEN (EN 967-1996)	CSA (Z262.1-M90)	ISO (DIS 10256-1996)
Test Apparatus	Monorail	Guided wire	Monorail	Monorail
Inbound Velocity	4.5 m/s \pm 2%	3.96 m/s \pm 0.08	3.96 m/s \pm 0.08	3.96 m/s
Neck Compliance	Rigid	Unrestrained head	Rigid	Rigid
Impacting Surface	MEP	Steel	Steel	Steel

Currently, helmet safety standards in the United States and Canada prescribe the use of rigid (non-compliant) necks in their test protocols while various sport helmet standards in Europe employ free (“unrestrained”) heads. In addition, the proposed NOCSAE standard for football helmets on the horizontal linear impactor recommends using a compliant Hybrid III neck to increase the biofidelity of the test system and to better replicate on field impacts (NOCSAE, 2006). It is clear that there is no consistency in the safety standard test protocols, thus research that attempts to document, compare and understand the dynamic characteristics of helmet test systems is crucial. Such questions as whether or not multiple test apparatuses should be used, and which impact conditions should be included in the design and certification of sport helmets cannot be answered otherwise. The knowledge gained from such studies could lead to a significant and overdue change in helmet safety standards which would ultimately result in an improvement in the performance of helmets.

Previous research has demonstrated that an increase in neck compliance resulted in a decrease in translational acceleration, but an increase in rotational acceleration (Rousseau & Hoshizaki, 2009). Further, Thom and associates (1998) found that the amount of compliance in the system is directly related to the rigour of the test such that impacting a head form with an attached neck is more severe than impacting a head form with an

unrestrained head and neck. An early study conducted on Macaque monkeys indicated that the level of mobility of the head considerably affected the movement of the brain (Pudenz & Sheldon, 1946).

1.7 DEPENDENT VARIABLES

1. Peak resultant translational acceleration (g)
2. Peak resultant rotational acceleration (rad/s^2)

The dependent variables in this study include peak resultant translational acceleration and peak resultant rotational acceleration; terms which describe the dynamic response of a Hybrid III head form. Translational acceleration is defined as the rate of change of linear velocity and is measured in g s, a unit which represents the acceleration due to gravity (Robertson, 2004; Kerr, 2010). Peak translational acceleration is the maximum value on a translational acceleration versus time curve. Rotational acceleration is defined as the rate of change of angular velocity (Robertson, 2004). Peak rotational acceleration is the maximum value on a rotational acceleration versus time curve. Currently peak translational acceleration and the Gadd Severity Index (GSI) are the head injury predictors used in many safety standards. Both of these variables are based on translational acceleration however GSI also incorporates a 2.5 weighting factor (Gadd, 1966). Various researchers, however, have identified rotational acceleration as an important predictor of concussion (Ommaya, Rockoff & Baldwin, 1964; Gennarelli, Ommaya, & Thibault, 1971; Gennarelli & Thibault, 1982; Gennarelli, Thibault, Adams, Graham, Thompson, & Marcincin, 1982). Thus, it is important to understand what can be done to prevent rotational accelerations of the head and brain, and to do this, the test

protocols must be able to elicit and measure rotational acceleration. Most standards constitute drop test protocols which restrict rotational accelerations by using a rigid neck, and certainly fail to measure them.

A recent study conducted in the Neurotrauma Impact Science Lab at the University of Ottawa demonstrated that under different conditions of velocity, there was a significant difference between peak translational and rotational accelerations, however, the other injury indices examined, which included GSI, were all moderately to highly correlated with peak translational acceleration (Rousseau, Walsh, Post, Coulson & Hoshizaki, 2010). As a result, it is believed that simply using peak translational acceleration will capture the same information as combining it with any of the other indices, apart from rotational acceleration, and accordingly, none of the other indices were measured in this study.

1.8 SIGNIFICANCE

Helmets are one of the most effective ways of reducing head and brain injury in sport (Hoshizaki & Brien, 2004; McIntosh & McCrory, 2005). The way in which helmets are designed is a direct reflection of the certification standards which govern their use (Gilchrist & Mills, 1994). It is important that a helmet is designed based on the unique risks associated with a sport if it is to protect players from the injuries inherent to that sport. For example, hockey players are most likely to be injured by one of two behaviours: i) impacting their heads on the ice after a fall, or ii) colliding with other players (Goodman, Gaetz, & Meichenbaum, 2001; Flik *et al.*, 2005). Current ice hockey safety standards worldwide, however, certify helmets solely with the use of drop towers, which neglect to consider the dynamics of a collision-type impact.

As innovative impact testing technologies are introduced, researchers and product developers are faced with decisions regarding which systems to use in the development of safety equipment. Characterization of test apparatuses is essential to gain a better understanding of the dynamic responses associated with the different systems, such that this information can be incorporated into the design of new helmets.

In this study, the relationship between two injury mechanisms was investigated because prior to this, it was not well understood. Research has demonstrated that greater mass impacts are much more severe, however it is not clear which characteristics are responsible for this. A goal of this study was to characterize the dynamic response of this relationship on a human head surrogate. In this case, mass represented the different injury mechanisms and acceleration represented the dynamic response.

It is hoped that the results of this study will help describe impact dynamics relating to standard test methods. With knowledge of how the test variables are related, helmet safety standards could be appropriately modified to more accurately reproduce behaviours specific to the sport being regulated, ultimately enhancing the performance of sport helmets and reducing the risk of sport-related brain injuries.

1.9 LIMITATIONS

Due to the load limitations of the test systems, high-velocity impacts were avoided in this study. In the case of the monorail drop system, inbound velocity values correspond with head form height up the tower; the greater the inbound velocity, the greater the drop height. Since the test was so severe, velocities above 4.0 m/s risked damaging the various parts of the test system, such as the unprotected head form and the guide rail. In terms of the linear impactor, velocity levels were governed by spring tension; the higher the tension

in the springs, the greater the velocity of the impacting arm. Since the head form was unprotected, impacts above 4.0 m/s risked damaging the head form. Since the purpose of varying velocity in this study was not to cover the spectrum of impact velocities an athlete may encounter in a sport, but rather to investigate whether the dynamic response of the head form was affected by velocity, these restrictions did not affect the scope of this study. Thus, three velocity points within the range, but not necessarily at the extremes, of impacts an athlete may encounter should be sufficient to examine this relationship.

In addition, the impact systems were matched as closely as possible to the systems currently used in helmet development labs, with as few modifications as possible. However, to increase reliability in the data, the same head form was used for all impacts and as a result the head-neck support system configuration varied depending upon which impact system was used. This variance may have introduced differences in the data. Since this study was designed to compare the currently used systems, the differences in set-up configurations were unavoidable.

1.10 DELIMITATIONS

This study is delimited to front impacts on a Hybrid III head form at 25.0 ± 2.0 degrees from the horizontal on the monorail drop tower and 25.0 ± 2.0 degrees from the vertical on the horizontal linear impactor, directed through the centre of mass of the un-helmeted head form.

1.11 ASSUMPTIONS

For this study, it was assumed that the Hybrid III head form accurately replicates the reaction of an adult human head subjected to impact.

CHAPTER 2. REVIEW OF LITERATURE

2.1 TRAUMATIC BRAIN INJURIES

One of the leading causes of death in Canada and the United States, traumatic brain injury (TBI) is a serious societal issue, particularly among youths and young adults (The Brain Injury Association of Canada, 2008; Thurman, Alverson, Dunn, Guerrero, and Sniezek, 1999). Nearly 17,000 Canadians were hospitalized in the 2003-2004 fiscal year for TBIs, and the numbers are even greater in the United States, where an estimated 1.6 to 3.8 million Americans suffer sports-related TBIs each year, leaving roughly 2% of the country living with TBI associated disabilities (Canadian Institute for Health Information, 2006; Langlois *et al.*, 2006; Zhang, Yang & King 2004; King, Yang, Zhang, & Hardy, 2003). The consequences of suffering a TBI are not only devastating for the injured party, but also for family-members and society at large, as a sudden, heavy economic and emotional burden are placed upon them (Zhang, Yang & King, 2004; Levine, 1999). In fact, it is estimated that \$3 billion is spent in Canada and \$60 billion is spent in the United States annually on medical bills and lost productivity as a result of TBI (Brain Injury Association of Alberta, 2009; Finkelstein, Corso, & Miller, 2006).

The term TBI broadly encompasses an extensive range of brain injuries, from mild incidences to catastrophic events (Bayly, Cohen, Leister, Ajo, Leuthardt, & Genin, 2005). Of interest in this research is mild traumatic brain injury (mTBI), and although the term implies a minor injury, mTBI (commonly known as concussion) can have devastating consequences including impaired neurocognitive functioning (Barth, Freeman, Broshek, & Varney, 2001). Concussion may be defined as “a complex pathophysiological process affecting the brain, induced by traumatic biomechanical forces” which “may or may not

involve loss of consciousness” (McCrory, Johnston, Meeuwisse, Aubry, Cantu, Dvorak, Graf-Baumann, Kelly, Lovell, & Schamasch, 2005). Some of the acute signs and symptoms associated with concussion may include blurred vision, vertigo, cognitive dysfunction, amnesia, tinnitus, headache, nausea, vomiting, difficulty concentrating, and a balance disturbance, while the delayed signs may be fatigue, lethargy, sleep irregularities, personality changes, and depression (Wojtys, Hovda, Landry, Boland, Lovell, McCrea, & Minkoff, 1999; Zhang, Yang & King, 2001). These signs and symptoms may quickly vanish or last for an extended period of time (Powell, 2001).

Concussion is typically caused by one of the following categories of loading: impulsive (indirect) or impact (direct), the latter of which is important to this study (Bailes & Hudson, 2001; Goldsmith & Plunkett, 2004; Gurdjian, Webster, & Lissner, 1958). Impulsive loading occurs when the head is indirectly placed in motion in response to a sudden force applied to a different body part (whiplash-type motion). In this case, rotation of the head occurs about the cervical spine, and since the brain is not rigidly attached to the skull, the two structures move at different rates causing the bridging veins to stretch and become damaged (Goldsmith & Plunkett, 2004; Gurdjian, Hodgson, Thomas & Patrick, 1968). Impact loading occurs when the head strikes a solid stationary object (a fall-type impact) or when a solid moving object directly strikes the head (a collision-type impact) over less than 200 milliseconds, such that contact phenomenon and wave propagation take place (Bailes & Hudson, 2001; Goldsmith & Plunkett, 2004; Gurdjian *et al.*, 1958). This inevitably results in both translational and rotational accelerations of the brain, causing diffuse axonal shearing in the white matter of the cerebellar and cerebral hemispheres (Biasca, Wirth, & Tegner, 2002; Goldsmith & Plunkett, 2004). The reversibility and

severity of the neurological deficit and associated symptoms are dependent upon the location and number of axons damaged (Biasca, Wirth, & Tegner, 2002).

Several indices have been developed to better understand, quantify and facilitate the diagnosis of traumatic brain injuries, including the Abbreviated Injury Scale (AIS). Initially developed in 1971 as a standardized injury severity categorization system for vehicular crashes, the AIS has undergone many adaptations over the years (Greenspan, McLellan, & Greig, 1985; Gennarelli & Wodzin, 2006). In 1998, the AIS ranked injuries on a six-point ordinal scale for each body region. The AIS categorization for brain injury can be found in Table 2.

Table 2. The Abbreviated Injury Scale for brain injury (Shojaati, 2003).

AIS	Category	Injuries
1	Minor	Light brain injuries with headache, vertigo, no loss of consciousness, light cervical injuries, whiplash, abrasion, contusion
2	Moderate	Concussion with or without skull fracture, less than 15 minutes unconsciousness, corneal tiny cracks, detachment of retina, face or nose fracture without shifting
3	Serious	Concussion with or without skull fracture, more than 15 minutes unconsciousness without severe neurological damages, closed and shifted or impressed skull fracture without unconsciousness or other injury indications in skull, loss of vision, shifted and/or open face bone fracture with antral or orbital implications, cervical fracture without damage of spinal cord
4	Severe	Closed and shifted impressed skull fracture with severe neurological injuries
5	Critical	Concussion with or without skull fracture with more than 12 hours unconsciousness with haemorrhage in skull and/or critical neurological indications
6	Survival not sure	Death, part or full damage of brainstem or upper part of cervical due to pressure or disruption, fracture and/or wrench of upper part of cervical with injuries of spinal cord

In 2005, Yogananden and colleagues derived a relationship between the AIS scores and the concussion grading system proposed by Ommaya and Gennarelli (1974), such that concussion grades 1 through 3 relate to an AIS score of 1 with no loss of consciousness (an injury Yogananden and associates labelled as “mild concussion”), a grade 4 concussion can refer to an AIS score of either 2 or 3 depending on the length of time spent unconscious (less than an hour up to 6 hours; described as “classical” or “severe concussion”), and a grade 5 concussion can apply to AIS 4, 5, or 6, again depending on the amount of time the person loses consciousness (6 hours to greater than 24 hours) but also depending on the degree of physical brain damage (no brain abnormality to decerebration or decortication; classified as “mild”, “moderate”, or “severe diffuse axonal injury”). In the most recent update to the AIS, the section on concussive injuries was removed and replaced with a contemporary set of injury descriptors to enhance the accuracy of coding these injuries (Gennarelli & Wodzin, 2006).

Validated thresholds indicating the levels of impact loading which result in specific brain injuries do not currently exist, however, in 2004 a group of researchers published proposed thresholds for mild traumatic brain injury in terms of translational and rotational accelerations (Zhang, Yang and King, 2004). These thresholds were based on twenty-four reconstructed helmet-to-helmet football collisions gathered by Pellman and colleagues in a series of NFL studies entitled “Concussion in professional football”. The data were then analyzed via finite elements modelling. The proposed thresholds can be found in Table 3.

Table 3. Thresholds for mTBI as proposed by Zhang, Yang and King (2004).

Translational acceleration threshold (g)	Rotational acceleration threshold (rad/s²)	Probability of suffering mTBI (%)
66	4,600	25
82	5,900	50
106	7,900	80

Head and brain injuries are common in sports and recreational activities. In fiscal 2003-2004, these activities were the third leading cause of traumatic head injury admissions in Canadian hospitals (Canadian Institute for Health Information, 2006). Of all serious head traumas suffered by adults aged 20 through 34 years, roughly 16% take place in sports and recreation, the majority of these being attributed to six specific activities: ice hockey, cycling, playground, soccer, football and rugby (Kelly *et al.*, 2001). This poses a serious problem due to the high participation rates in these activities. In addition, the reported number of annual TBIs is likely a sizeable underestimate as many TBIs that take place remain unrecognized and undiagnosed (Delaney, Lacroix, Leclerc, & Johnston, 2000; Delaney, Lacroix, Gagne, & Antoniou, 2001).

Two common mechanisms of head and brain injury in sport include the fall-type where a moving head impacts a stationary surface of large mass such as the ground, and the collision-type which involves the head being impacted by a moving object of smaller mass such as another player or object (Honey, 1998; Azuelos *et al.*, 2004; Flik *et al.*, 2005; Guskiewicz *et al.*, 2000). According to Patrick (1966), there are fundamental dynamic differences between these two injury mechanisms; and accordingly, both mechanisms must be considered in developing protective devices for the head. It should be noted that a quantitative comparison of these two injury mechanisms has yet to be undertaken.

Although it is generally agreed that there is an inherent risk of injury associated with participation in sport and recreation activities (Powell, 2001; Biasca & Simmen, 2004; Miele, Norwig & Bailes, 2006), various rules, regulations, safety standards and pieces of equipment have been developed to reduce this risk. Sport helmets have been especially effective in mitigating the risk of head and brain injury, particularly in contact sports such as American football and ice hockey (Hoshizaki & Brien, 2004).

2.2 PREVENTION OF BRAIN INJURIES

Three of the most effective ways of reducing the risk and incidence of head and brain injuries in sports include: i) enforcing rules and regulations which manage game play; ii) educating and training athletes in defensive skills; and iii) developing personal protective equipment which lessen the amount of energy transferred to the head and brain (McIntosh & McCrory, 2005). The focus of this study is on the latter; specifically, helmet development.

2.2.1 *Head Protection*

Helmets are designed to reduce the risk of suffering head and brain injuries by managing impact forces which would otherwise be transmitted to the skull and its contents (Cairns, 1941; Hoshizaki & Brien, 2004). From their modest beginnings as padded leather head harnesses to the modern, fashionably-assembled helmets composed of a hard plastic shell and a sophisticated energy-attenuating liner, sport helmets have undergone many changes over the years (Biasca, Wirth, & Tegner, 2002; Honey, 1998; Fekete, 1968). Designing effective helmets is an exceedingly complex process, involving the development of test conditions and methodology which accurately reflect the head injury mechanisms

seen in sports (Becker, 1998; Sabelli & Morehouse, 2000). Helmets must also pass certification standards, which assess their performance capacity, prior to being released to the consumer market (Morehouse, 2000). Impact reconstruction apparatuses and safety standards are imperative to the improvement of current helmets and the production of new helmets as they are used to quantitatively evaluate the safety performance of the protective devices. Impact test apparatuses and protocols; however, have not always been as sophisticated and ethical as they are today.

2.2.2 History of Impact Testing

Impact testing on human and animal subjects was initiated in the late 1930s by Siegfried Ruff (Stapp, 1983). Although his research focused primarily on full body impacts and deceleration conditions at the time, it was Ruff who first studied the principles of stress analysis on humans, distinguishing between static and dynamic stresses, and comparing prolonged accelerations with brief impact accelerations (Stapp, 1983). Thus were born the concepts of peak acceleration, impact duration and nature of inertial forces in terms of the human body. Ruff carried out animal experiments on test apparatuses comprising of vertical guide rails, described as being similar to a guillotine, which supported a cart with eight rollers. The system had the ability to be shot upward via stretched bungees until impact with a bumper occurred, or to be dropped 3 meters onto a lower bumper (anvil). Acceleration forces resulting from the drop tests ranged from 50 to 600 g with impact durations between 2 and 15 milliseconds. Other systems explored by Ruff and colleagues included: i) a horizontal monorail carriage propelled by an ejection seat catapult reaching speeds from 4.5 m/s to 14.5 m/s, ii) a pendulum like swinging platform, iii) a portable catapult which accelerated a seat via tension in stretched rubber

landing-gear shock cords, and iv) an air propelled slipper-mounted sled more commonly known as “The Daisy Decelerator” (Stapp, 1983; Ruff, 1950). Each of these apparatuses was tested on human volunteers at the cost of several lives.

The first helmet impact test, called the Brinell test, was developed in the 1940s and consisted of a head form, supported by the arm of what resembled an oversized stapler, pressing a steel ball into a small aluminum impression bar (Becker, 1998). This test underwent many revisions over the years, including the addition of a force transducer and the ability to impact a greater range of locations, and remained in use until the late 1950s.

In the mid-1940s, various researchers carried out impact tests on monkeys, dogs and cats, in an attempt to better understand the effects of head trauma on the brain (Groat, Windle & Magoun, 1945; Gurdjian & Lissner, 1944; Pudenz & Shelden, 1946; Walker, Kollros & Case, 1944). A number of impact apparatuses were employed to induce head trauma such as pendulums, compressed air guns, and hammers. Likewise, a variety of systems for measuring and observing brain trauma were employed including accelerometers, strain gauges, and “pressure plugs” mounted to the skulls of the animals, and large translucent apertures in the skull covered by a lucite calvarium.

In the late 1940s and early 1950s, Lombard and associates impacted human volunteers wearing newly available protective helmets with a six-foot pendulum equipped with a ten pound striking ball. The helmet consisted of a fiberglass shell and a one inch thick cellular cellulose acetate liner. When worn, the protective device allowed subjects to tolerate impacts up to, but not greater than 38 g (a value well below the perceived concussion level), at which point subjects complained of headache, neck pains and apprehension (Stapp, 1983; Lombard, Ames, Roth, & Rosenfeld, 1951). Around the same

time, George Snively of the Snell Memorial Foundation began using the “swing-away” test rig; a wire guided impactor which would horizontally strike a head form equipped with an accelerometer at its centre, and mounted on a pivoting armature, held in place by a glass rod (Becker, 1998). The system allowed researchers to impact a greater number of impact locations than before, however it was mechanically complex. Studies eliciting “experimental concussion” in cats, dogs and monkeys continued during this time (Foltz, Jenkner & Ward, 1953; Gurdjian, Lissner, Latimer, Haddad & Webster, 1953).

The mid-1950s brought research conducted by Lantham which examined the response of the heads and necks of seated human volunteers subjected to sinusoidal shake table oscillations and sledge hammer blows from below. In the late 1950s, Ward examined the cortical activity of freely suspended cats and monkeys subjected to 9.1 m/s pneumatic piston impacts (Ward, 1958), while Evans and colleagues were carrying out free fall and guided drop tests on embalmed human cadaveric heads (Evans, Lissner & Lebow, 1958). Around the same time, engineers began using anthropomorphic test dummies in the form of plastic heads, as opposed to human and animal subjects, in the study of skull fracture (Evans, Lissner & Lebow, 1958).

In the early 1960s, Friede conducted research on anesthetized cats subjected to head impacts, striking the sagittal sutures of the cats with a pneumatic hammer fitted with an impact plate (Stapp, 1983; Friede, 1961). The hammer traveled 3.18 cm, driving the cats’ heads back at speeds of 8.5 to 13 m/s, resulting in typical signs of concussion and brain stem damage. Meanwhile, Gurdjian and colleagues were examining blows to embalmed, silicone-gel filled cadaveric heads using a two pound, 2.86 cm diameter blunt striker to examine the effect of varying pulse duration (Stapp, 1983; Gurdjian, 1966). They found

that as pulse duration was increased, the injury effects became more severe. Ommaya and associates continued to examine the effects of gas-pressure linear impacts on the brains of monkeys, delivering blows to their freely mobile or immobilized heads at magnitudes ranging from 2.7-10.7 m/s (Ommaya *et al.*, 1964). Impacts were delivered to either the occipital region or right temporal region of the monkeys' skulls. This work demonstrated that, with their set-up, high linear accelerations were hard to elicit when the neck was immobilized (i.e. in a rigid neck condition) and that higher accelerations were associated with impacts to the occipital region.

By the mid-1960s, falling head form devices became popular (Becker, 1998), although studies of experimental concussion in live animals continued to take place into the 1980s (Ommaya, Grubb & Naumann, 1971; Ommaya, Corrao & Letcher, 1973; Ono, Kikuchi, Nakamura, Kobayashi & Nakamura, 1980). These systems involved a head form of small mass impacting an appropriately shaped surface of much larger mass from a predetermined height. Acceleration data were collected via an accelerometer mounted at the centre of the head form. To this day, falling head form devices are the test apparatus of choice for helmet safety standards due to the reproducibility and reliability of their data. Recently, however, horizontal linear impactors have gained popularity, as researchers are now able to obtain reliable data on these systems as well.

The early research conducted on human volunteers, human cadavers, and anesthetized animals led to a preliminary understanding of the causes of head and brain injury, setting the foundation for further head impact testing, the construction of appropriate test apparatuses, the development of protective head devices and the establishment of helmet safety standards.

2.2.3 History of Head Protection and Safety Standards

Safety standards organizations have become an indispensable channel through which sports helmets are certified. These establishments govern the design and performance of sport helmets in terms of protective ability, and have played an important role in the design of impact reconstruction test apparatuses (Halstead *et al.*, 2000; Pellman *et al.*, 2006). In collaboration with researchers around the world, helmet safety standards are developed and validated, taking into consideration the mechanisms by which people are becoming injured, and updating the way in which these injury mechanisms are tested for (Becker, 1998).

Dating back to military helmets used in 1300 B.C., helmets have been in existence in one form or another for thousands of years (Becker, 1998). It was not until the early 20th-century, however, that their effectiveness to attenuate the risk of serious head injury was demonstrated (Becker, 1998). In the early 1940s, epidemiological studies on head injuries in motorcyclists demonstrated that those who wore crash helmets were less likely to suffer severe head and brain injuries (Cairns, 1941; Cairns & Holbourn, 1943). This pioneering work of Dr. Cairns was the first to relate the mechanisms of head and brain injury to the performance characteristics of crash helmets, which ultimately brought recognition to crash helmets as a valuable form of head protection (Becker, 1998). Although the results of this epidemiological study must be regarded with caution due to variation in the test conditions, the findings of Cairns and Holbourn led to a large-scale investigation of crash helmets by the Ministry of Transport in Great Britain, and ultimately the first performance standard for protective helmets by the British Standard Institution (BSI) in 1952 (Becker, 1998; BSI, 1960). The test methods attempted to replicate the

dynamics of an impact immediately prior to contact. To achieve this, a 4.5 kg block of hardwood was dropped from a height of 2.7 m onto the helmeted head form and force data were measured from a gauge (Becker, 1998; BSI, 1960). A performance threshold, set at 2268 kg, dictated the maximum output force a helmet could attain and still qualify for certification (Becker, 1998).

In the United States, the evaluation and improvement of crash helmet performance was prompted by the accidental car racing death of William “Pete” Snell in 1956 (Hoshizaki & Brien, 2004; Snell Memorial Foundation [SMF], 2008). In response to the helmet’s failure to protect Snell in what was deemed a survivable crash, medical doctor George Snively who was a friend to Snell and a fellow racer, founded the Snell Memorial Foundation in 1957, named in his honour (Aya, 1965; Becker, 1998; SMF, 2008). Research conducted by Snively and colleagues led to the 1959 publication of “Standards for Racing Crash Helmets”, the first American performance standard designed to regulate the protection capabilities of vehicular crash helmets (Becker, 1998; Snell Memorial Foundation, 1959). In the years that followed, the Snell Memorial Foundation began certifying helmets in addition to developing new standards and continually revising existing ones (Becker, 1998).

In the 1960s and 1970s, a number of other independent organizations and committees concerned with the development of safety standards for protective headgear emerged. The International Organization for Standardization (ISO) formed such a committee in 1960, publishing Recommendation R 1511, Protective Helmets for Road Users in 1970, followed by a draft standard entitled “Headforms for Use in the Testing of Protective Helmets” in 1983 (Becker, 1998). Not a certifying body, ISO remains a non-

governmental organization which publishes standards with the intention of promoting international trade and bridging the gap between the public and private sectors (ISO, 2009).

In 1961, the American National Standards Institute (known as the American Standards Association at the time) followed suit (American National Standards Institute [ANSI], n.d.; Becker, 1998). Similar to ISO, the goal of this group was not to issue helmet certifications, but rather to police the development and revision of consensus standards. The first standard published under this committee was Z90.1-1966, Protective Headgear for Vehicular Users in 1966 (Becker, 1998). Prior to completing the first draft of their football helmet standard, the ANSI subcommittee for headgear in sports disbanded, leading two committee members to voluntarily complete the task (Becker, 1998). Medical doctors A. F. James and H. A. Fenner published their football helmet standard, JF73, in 1973.

The American Society for Testing and Materials (ASTM) is an organization similar to ANSI in that they develop consensus standards but do not issue helmet certifications (Becker, 1998). ASTM published their first standard specification for football helmets in 1989, and currently has a repertoire of helmet standards ranging from pole-vaulting to horseback riding to various types of cycling (ASTM, 2009).

The Consumer Product Safety Commission (CPSC) was formed in 1973 to protect the public “against unreasonable risks of injuries associated with consumer products” (CPSC, n.d.). Working alongside the ASTM headgear subcommittee, the CPSC develops voluntary standards for recreational helmets and headgear. Under the direction of Congress, it became mandatory in 1999 that all bicycle helmets manufactured or imported for sale in the United States pass the CPSC federal safety standard (CPSC, 2006).

The Canadian Standards Association (CSA), a not-for-profit membership-based accredited standards development organization, introduced its first performance standard for sport helmets in 1972 (CSA, 2009; Bishop, 2000). CSA now has standards for ice hockey, cycling (introduced in 1989) and alpine skiing/snowboarding helmets (introduced in 2008).

In 1989, the European Committee for Standardization (CEN) published their first standard for ice hockey helmets (CEN, 2009). To this day, the committee consists of 30 member countries collaborating to develop consensus standards and facilitate trade in Europe (CEN, 2009). CEN also works in parallel with ISO to develop common European and international standards (The CEN Management Centre, 2008). Notably, CEN is currently the only standard organization to require the use of a free drop system for their ice hockey helmets wherein the head form is not restricted to motion along a vertical axis. All other standards prescribe the use of restrained head forms.

In 1969, the National Operating Committee on Standards for Athletic Equipment (NOCSAE) was formed (Newman, Beusenber, Shewchenko, Withnall & Fournier, 2005). The goal of this committee was to reduce the amount of injury in competitive sports by developing standards for protective equipment (NOCSAE, n.d.). Based on reports of severe head injury in football, developing a helmet standard for this sport became priority, and a research program on football helmets commenced in 1971 (NOCSAE, n.d.). At the time, helmets were almost exclusively tested on the ANSI Z.90 metal head form, a mechanical device used by manufacturers which gave reliable data, but was not specifically designed to imitate the physical characteristics of the human head (Hodgson, 1975). A comparison of data from human cadaver heads led researchers to conclude that a more

realistic head model was imperative to the development of their test standard, as the response characteristics of the metal head form were unrealistic and distorting the helmet comparisons (Hodgson, 1975).

Thirteen cadaver heads in the most common helmet size (18.4 – 18.7 cm) were chosen to develop skull models (Hodgson, 1975). Information regarding the mass moment of inertia, the anthropometric measures and the weight of each cadaver head was included in these models, and various static load-deflection tests were carried out on both the models and the cadaver heads for comparison purposes. Ultimately, an 18.7 cm skull model was chosen which fell within the cadaver head load limits, and which included such features as a silicone gel brain, an Endevco triaxial accelerometer mounted at the centre of gravity of the head model, and a silicone rubber skin (Hodgson, 1975). To this day, head forms continue to evolve and researchers now have access to anthropomorphic test dummy head forms which are equipped with nine accelerometers, have the ability to measure accelerations in three dimensions, and are highly reliable. Biofidelity remains an issue as there is often a trade-off between how repeatable and how valid the data is.

The first NOCSAE test standard for football helmets was published in 1973 (Newman *et al.*, 2005; NOCSAE, n.d.). Impacts were conducted using a twin-guided wire system and dropping the head model outfitted with a helmet from a height of 1.52 m (equivalent to an impact velocity of 5.5 m/s) onto a 1.27 cm firm rubber pad (38 durometer) mounted on a rigid pedestal (Hodgson, 1975; Newman *et al.*, 2005). A severity index of 1500 GSI was set as the upper limit pass/fail criteria for helmet certification as it was previously established by Gadd (1966) that this was the impact severity level at which the risk of skull fracture was eliminated when wearing a football helmet.

Over the years, the NOCSAE standard has undergone various revisions, and it remains the gold standard for certifying football helmets. Recently, NOCSAE proposed a standard using a linear impact system in the evaluation of protective headgear performance characteristics (NOCSAE, n.d.). To date, they are the only organization to do so.

In addition to the standard organizations, sport-specific certification bodies also began to appear in the 1970s. For example, in 1978, an independent volunteer group called the Hockey Equipment Certification Council (HECC) was formed (HECC, 2009). The goal of this organization was not to write standards, but rather to validate the performance of sports equipment deemed by the product's manufacturer to meet certification standards, as well as to inform consumers regarding the performance capacity of certain helmets (Morehouse, 2000; HECC 2009). In validating the helmets, HECC decides which safety standard to follow for their certification program. Additionally, ice hockey helmets which do not bear a HECC certification label cannot be worn in leagues mandated by USA Hockey, the National Federation of State High School Associations (NFHS) or the National Collegiate Athletic Association (NCAA).

The process of certifying sports equipment varies across the countries. In Canada the process is simple: based on research examining the parameters of the game of hockey, a standard organization writes the standard (e.g., CSA) which the government then adopts and requires by law that only equipment certified under that standard be sold, then sports associations (e.g., Canadian Amateur Hockey Association) mandate that only said certified equipment can be worn in competition (Morehouse, 2000). In the case of CSA standards, certification is carried out by CSA. The process is similar in most European countries;

however, in the United States the process tends to be slightly more laborious, requiring that helmets be certified by a third party such as HECC (Morehouse, 2000).

Writing a high quality, representative standard is a challenging and time-consuming venture, often taking anywhere from two to ten years (Morehouse, 2000). In addition, some organizations require that each standard be revised at specific intervals (e.g., ASTM requires re-evaluation every five years) to ensure that the standards remain adequate and current with improved technologies and enhanced knowledge of injuries (Becker, 1998).

To this day, most standards constitute drop rig protocols wherein a head form is dropped from a prescribed height onto an impacting anvil of large mass (Bishop, 2000). These drop towers mimic an athlete's head impacting the ground (fall-type impact), and although this is a common mechanism of head injury in sport, it is not the only one. Different apparatuses exist to accomplish this objective, including the monorail drop tower, the NOCSAE twin-guided-wire system, and the free drop system. Each of these apparatuses has a unique set of pros and cons, but all of the systems are limited in that they cannot reconstruct a collision-type impact, meaning that they fail to account for the various injury mechanisms athletes may encounter in any given athletic exposure. Fall-type and collision-type impacts represent two fundamentally different impact dynamics, and both must be considered in head protection (Patrick, 1966). New technologies are now emerging which examine non-vertical impact vectors, such as the linear impactor machine, which better mimic collision-type impacts (Pellman *et al.*, 2006).

First used as a helmet testing device in the 1950s by George Snively, this system was designed to deliver a horizontal impact between two moveable bodies; an interaction that differed significantly from the one-body impact systems previously used and the drop

systems which were later adopted (Becker, 1998). Modern day linear impactors are pneumatically powered, and have the ability to reach velocities of 15 m/s (Pellman *et al.*, 2006). The linear impactor system employed by Viano & Pellman (2005) was designed to match the kinematics of impacts they reconstructed from NFL game-play video data. Similarly, the system proposed by the NOCSAE developed theirs to more closely reproduce the on-field impacts which are believed to cause mTBI (NOCSAE, 2006). Low-velocity, spring loaded linear impactors also exist which allow researchers to reach velocities as low as 1 m/s.

As technologies continue to advance, more sophisticated, biofidelic, and injury-representative test systems, undoubtedly equipped with head forms outfitted with multiple internal accelerometers, will almost certainly become the norm in helmet standards.

2.2.4 Comparisons of Helmet Testing Systems

In 1966, Gurdjian and colleagues compared the protective indices of cadaveric impacts from drop tests and rotary hammers. The protective index, which is based on the ratio of impact velocities required to produce a given acceleration for a protected and unprotected head, gives an indication of the protection against accelerative forces afforded by the safety device under investigation. It was found that the energy attenuated by the protective device and its contents was maximized during the drop test due to the immovable nature of the impacting surface.

Thom and colleagues (1998) examined the differences between two drop systems (monorail guided drop versus free drop) and two head form varieties (Department of Transportation [DOT] versus ISO) in the context of motorcycle helmets. Their results showed that the monorail drop rig set-up with a restrained head form was a consistently

more severe test than the set-ups that were not guided or that did not have a restrained head form. Although an important finding, this study gives little information regarding the characteristics of another impact apparatus, the linear impactor, which is designed to simulate an athlete being hit by another player or object.

In 2000, a comparison of four international safety standards for ice hockey helmets was published by Pearsall, Wall & Hoshizaki. The differences between standards CAN/CSA-2262.1-M90, ASTM F1045-90a, EN 967-1996 and ISO DIS 10256-1996 were examined on four of the leading ice hockey helmets available on the market at each of six prescribed impact locations (front, front boss, side, rear boss, rear, and crown), using two different impact apparatuses (a monorail drop tower for CSA, ASTM, and ISO standards, and a guided wire freefall system for the CEN standard). The performance characteristics of the helmets were evaluated in terms of translational acceleration (peak g). Overall, there were no significant differences between testing standards, but significant differences were found between test standards across impact location at four sites (front boss, rear, rear boss, and crown). The CEN standard generated the highest translational accelerations at four of the six impact sites, implying that the CEN standard is harsher than the others; however, since the interaction produced mixed effects, no individual standard was labeled as more or less severe. Ultimately, the researchers found that due to variations in the standards in terms of head form composition, impact surface, test apparatus, and drop height, comparison of safety standards is a highly difficult and complex endeavour. A recommendation was put forth that a single standard protocol able to produce a reliable performance criterion should be adopted.

The rigor of drop tests and the associated danger of fall-type impacts are emphasized in each of these studies, while the less severe but more common collision-type impact is overlooked. Failure to consider this second head and brain injury mechanism when designing new sport helmets may result in the production of helmets that lack essential protective qualities and ultimately put athletes at risk of suffering mTBI. Therefore, it is imperative that research be undertaken which may help identify the risks associated with each impacting condition.

In 2003, Pellman and colleagues began publishing a six-year longitudinal project assessing head impacts in professional football, reconstructing helmet-to-ground (fall-type) and helmet-to-helmet (collision-type) impacts from video data. The same custom guided drop rig was used for both conditions, with different impacting surfaces; a simulated ground composed of artificial turf and a backing pad resting atop a platform was used in the helmet-to-ground case, while a second helmeted head-neck assembly, this time attached to a freely suspended Hybrid III torso and pelvis was employed in the helmet-to-helmet trials. The major difference between the two test set-ups was impact mass: a head form of small mass impacting an immovable surface of large mass in the helmet-to-ground trials versus the same head form impacting a 46.4 kg movable object for the helmet-to-helmet impacts. In an effort to remove the vertical force vector and better represent the helmet-to-helmet impacts, work was undertaken to create a new linear impact system. A failed attempt at developing a versatile pendulum system resulted in modifications to the Wayne State University (WSU) newly designed linear, pneumatic impactor (Pellman *et al.*, 2006). Comparisons were later made between the initial helmet-to-helmet vertical drop data and the newer helmet-to-helmet horizontal impact data. The researchers found that for impacts

to the front quadrant of the head (which encompassed up to 45° to the left and right of the eyes), the horizontal impacts were less severe than the vertical impacts (Pellman *et al.*, 2006). Although this gives important information regarding the two systems evaluated by Pellman and colleagues, no comparisons were made between the helmet-to-ground and horizontal helmet-to-helmet impact data, and there remains a gap in this area of research.

It is important to understand the relationship between head injury mechanisms and helmet test apparatus characteristics, as this knowledge directly affects the way equipment manufacturers and researchers develop sport helmets. Due to the unethical nature of carrying out impact testing on human subjects, injury prediction variables are used to quantify the dynamic response of human head surrogates in this developmental process.

2.3 INJURY PREDICTION VARIABLES

The variable most commonly used in the prediction of brain trauma is translational acceleration. In the 1940s, via analysis of hammer blows to the heads of anesthetized dogs, Gurdjian and Lissner concluded that acceleration of the head can lead to deformation of the skull and changes in intracranial pressure; two afflictions highly associated with traumatic brain damage (Gurdjian & Lissner, 1944). It was further highlighted that peak intracranial pressure occurred simultaneously with maximum acceleration. At the time, researchers were unsure of the relative contribution of the two forms of acceleration: translational and rotational. A separate study in 1955 led Gurdjian and colleagues to conclude that translational acceleration played a more significant role in the production of intracranial trauma than rotational acceleration (Gurdjian, Webster & Lissner, 1955).

Over the years, various researchers corroborated Gurdjian's findings; most notably Ommaya and colleagues in the early 1970s with their experiments on impact and impulse

loading the heads of monkeys, and Ono and associates in the early 1980s with their study on the separation of translational and rotational accelerations applied directly and indirectly to the heads of monkeys (Ommaya, Hirsch & Martinez, 1966; Ono *et al.*, 1980).

To this day, peak translational acceleration remains the input variable of choice for all helmet safety standards. Although peak rotational acceleration has yet to be adopted in the helmet standards, various researchers worldwide have reported that it is more indicative of mild traumatic brain injury than translational acceleration.

In 1943, Holbourn reported that based on the physical properties of the brain (large bulk modulus versus small modulus of rigidity), shear-strains in the brain should be responsible for injury, not compression or rarefaction strains. Using brain and skull models, his research demonstrated that rotational acceleration forces produce large shear-strains in the brain, whereas translational accelerations produce minimal shear-strains, leading to the conclusion that rotational acceleration is the main predictor of brain injury (Holbourn, 1943). In a follow-up study, Holbourn emphasized the idea that translational acceleration is negligible in every incident when compared to rotational acceleration. This is because during translational acceleration, no part of the brain lags behind the motion of the skull, whereas in rotational acceleration, the brain – not being a rigid structure – moves relative to the skull, causing brain distortion and injury (Holbourn, 1945).

In the early 1970s, Gennarelli and colleagues induced cerebral concussion in squirrel monkeys, comparing the translational and rotational motions of the head. They found that none of the monkeys subjected to translational accelerations exhibited signs of cerebral concussion, while all of the animals exposed to rotational accelerations did

(Gennarelli, Ommaya & Thibault, 1971). This finding led them to conclude that rotational acceleration is the better predictor of mTBI.

2.4 SUMMARY

Traumatic brain injury is a major societal issue, particularly in sporting arenas. Sport helmets were designed to mitigate the risk of suffering brain trauma; however, the helmet development process is not without flaws. Safety standards, which govern the design and protective capacity of sport helmets, solely prescribe the use of vertical drop systems in their certification process. These test apparatuses mimic fall-type impacts, wherein an athlete's head impacts the playing surface. Although this is a common injury mechanism in sport, it is certainly not the only way in which athletes are injured. Collision-type impacts are the primary cause of mild traumatic brain injury in contact sports such as ice hockey and American football, yet these types of impacts are not accounted for in today's safety standards. This oversight may have a detrimental effect on the athletes wearing helmets which were not optimally designed.

Understanding injury mechanism is incredibly important as it will dictate how researchers and equipment manufacturers measure risk. Currently, there are mixed opinions regarding how to best quantify risk. Various researchers contend that translational acceleration is the best brain injury predictor, while others argue that rotational acceleration is more indicative of trauma. The goal of this research is to evaluate the effect of impact mass on the dynamic response of a Hybrid III head form in terms of peak resultant translational and rotational acceleration, over a range of inbound velocities and system compliances. It is hoped that the information generated in this study will improve the way in which athletes are protected in sport.

CHAPTER 3. RESEARCH METHODOLOGY

3.1 DATA COLLECTION

The dynamic response of a Hybrid III head form impacted on two distinct helmet testing systems with different masses (158.8 kg, 17.1 kg) was examined in this study. The head form was impacted at three inbound velocities (2.0 m/s, 3.0 m/s, 4.0 m/s) and with three system compliances (rigid neck, compliant neck, unrestrained head and neck). Impacts took place between the head form and a 1” modular elastomer programmer (M.E.P.) pad impacting surface (Durometer 60 ± 5 , Shore A). Eighteen impact conditions were evaluated, and three sets of three impacts were carried out per test condition, for a total of 162 impact trials. Peak resultant translational and rotational acceleration data were collected for each trial.

3.1.1 Test Equipment

Monorail Guided Drop Apparatus

The monorail guided drop apparatus consisted of a 4.7 m vertical drop tower with a single track running its length, along the anterior surface. The tower was secured to the wall 3.7 m from the floor by a metal arm, ensuring that the tower remained fixed and stable throughout testing. A drop carriage which ran smoothly along the track via ball bearings supported an arm holder (700 ± 1 g) which delivered an impact with negligible friction. A custom connection piece slid into the arm holder (1102 ± 1 g), acting as the attachment point for the compliant and rigid necks. The carriage was used to raise and lower the head-neck complex to the chosen height (and corresponding inbound velocity) and a release mechanism on the carriage liberated the arm holder, dropping the head form. At the base of the drop tower was a supporting block for the M.E.P. impacting surface. A quick-

release mechanism held the impact pedestal in place. Upon impact, the head form contacted the M.E.P. pad in a vertical force vector. The monorail guided drop apparatus for the compliant and rigid necks can be seen in Figure 1, and the neck attachment configuration can be seen in Figure 2.

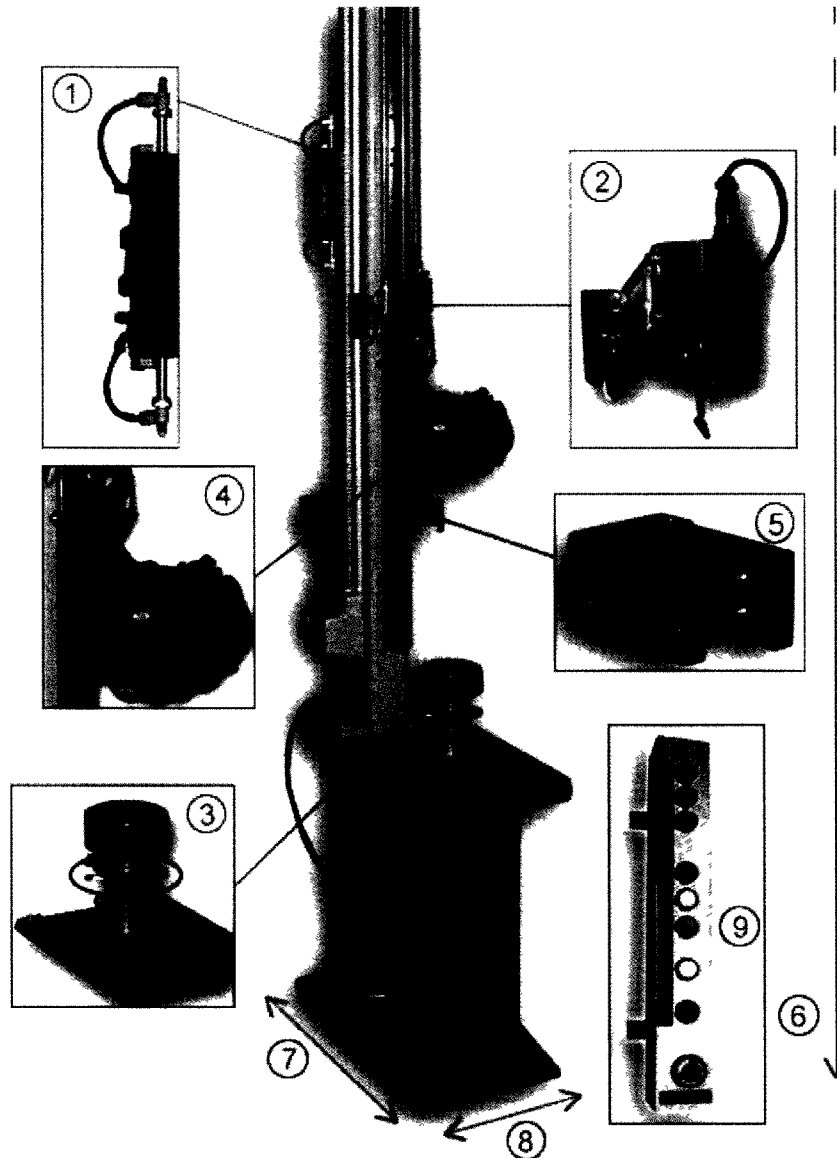


Figure 1. Lower portion of monorail guided drop system set up for compliant and rigid necks. The numbers correspond to specific components: 1) auto-retractable nylon rope brake system, 2) drop carriage with release mechanism, 3) quick release support clamp for impact pedestal, 4) head form attached to arm holder, 5) time gate, 6) direction of force vector, 7) depth of the base = 63.5 cm, 8) width of the base = 31.8 cm, and 9) drop control panel (Cadex, 2009b)



Figure 2. Set-up for the rigid and compliant neck trials on the monorail guided drop apparatus. The Hybrid III head form is attached to the neck which in turn is connected to the monorail ball arm holder. The impacting surface consists of a modular elastomer programmer pad mounted on an anvil.

The unrestrained head and neck impacts required a slightly different set-up. In this case, the neck was present for the purpose of maintaining consistency in mass across the impact conditions. As opposed to being attached to the arm holder, the Hybrid III neck was attached to a nylon rope which caught the head form upon impact. A hollow basket was inserted into the arm holder, which cradled the head form as it fell toward the impacting surface. The drop carriage raised the basket to the appropriate drop height (based on inbound velocity) and the same release mechanism as above freed the arm holder. The head form impacted the M.E.P. pad while the basket continued its downward motion, falling past the impacting anvil. Post-impact, the head form had the ability to move freely, until caught by the nylon rope. Under this condition, the head form was not driven into the impact surface. The monorail drop apparatus for the unrestrained head and neck can be seen in Figure 3 and the head form support system can be seen in Figure 4.

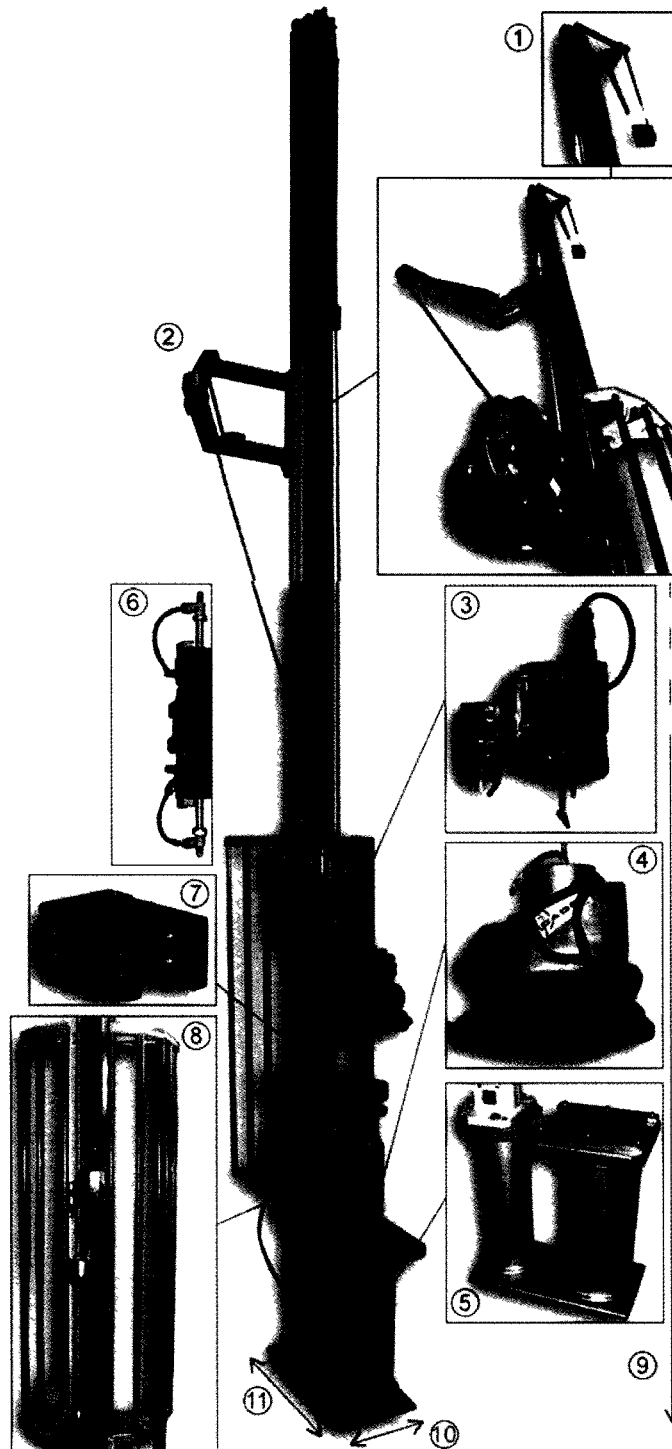


Figure 3. Monorail guided drop system set up for unrestrained head and neck. The numbers correspond to specific components: 1) portion of auto-retractable nylon rope brake system, 2) metal support arm, 3) drop carriage with release mechanism, 4) hollow basket supporting head form, 5) base, 6) auto-retractable nylon rope brake system 7) time gate, 8) safety cage, 9) direction of force vector, 10) width of the base, and 11) depth of the base (Cadex, 2009a)

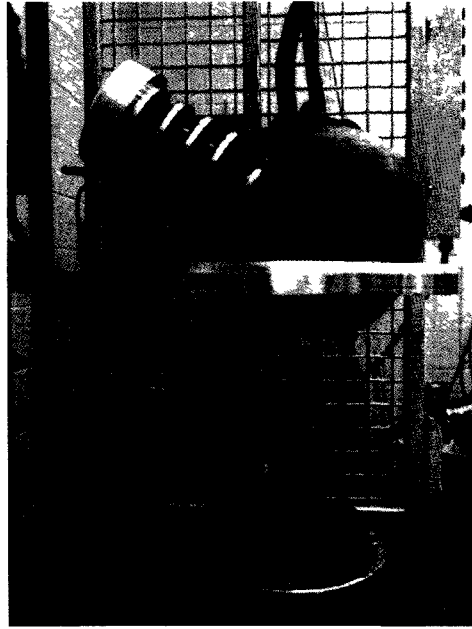


Figure 4. Set-up for the unrestrained head and neck trials on the monorail guided drop system. A hollow basket attached to the monorail arm holder cradles the Hybrid III head form which is tethered to the drop tower via nylon rope at the base of the compliant neck. The impacting surface consists of a modular elastomer programmer pad on a raised anvil.

The monorail drop tower was connected to a computer equipped with Cadex Software (Cadex Inc., St-Jean-sur-Richelieu, QC); a multi-functional program used to set up the impact parameters. The desired inbound velocity of the impact was entered into the software program, which then triggered the carriage to be raised to the corresponding drop height. The velocity of the head form upon impact was calculated automatically by the software as a small time gate flag of 0.2525 m passed in front of an infrared beam. The software then displayed the velocity values for recording purposes. An error of $\pm 2\%$ was accepted for each inbound velocity, as per the safety standards. Velocity values were previously validated in the Neurotrauma Impact Science Laboratory at the University of Ottawa, with the use of a high-speed video camera.

Linear Impactor Apparatus

The linear impactor test apparatus was designed to deliver impacts in the horizontal plane, removing the vertical force vector. The system consisted of an impacting arm (mass = 17.1 ± 0.1 kg) that was propelled forwards by a custom spring system (Figure 5). An electromagnet attached to a crank system pulled the impacting arm backwards, engaging the springs. The distance the springs were stretched corresponded with specific velocities, and desired impact velocities could be achieved by simply adjusting the tension in the system. Once the desired tension in the springs was obtained, the electrical feed to the electromagnet was severed, driving the impacting arm toward the Hybrid III head form. The impacting surface of the arm was consistent with the impacting surface of the monorail: a 1" M.E.P. pad. The linear impactor apparatus can be seen in Figure 5.



Figure 5. Linear impactor system (left); the plunger arm was thrust toward the stationary head form via spring loading (right).

During the compliant and rigid neck trials, the necks were connected directly to a sliding table, which rested atop a base of support and was free to move along the same axis as the impacting arm. A spring loaded brake system safely arrested the sliding table

following a displacement of 0.54 ± 0.01 m. In contrast to the vertical drop trials, the M.E.P. impacting surface struck the stationary head form, as opposed to the head form striking the stationary impacting surface (Figure 6). The connection between the neck and table allowed for control over impact location in five degrees of freedom: fore-aft (x), lateral (y), and up-down (z) translation, as well as fore-aft (y) and axial (z) rotation of the neck base. Although a number of positions were possible, head form and neck positioning remained constant throughout testing.

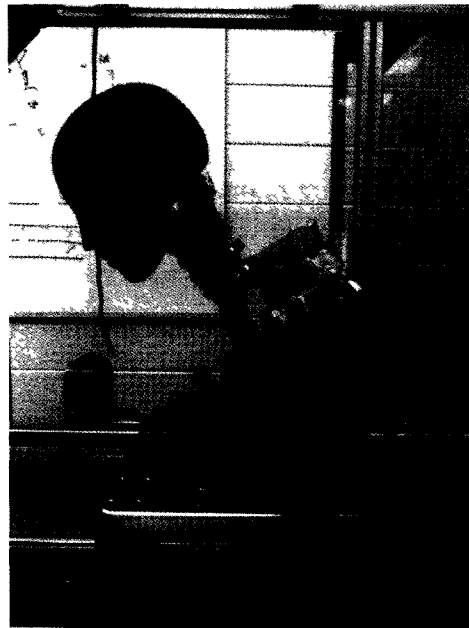


Figure 6 Set-up for the compliant and rigid neck trials on the linear impactor system. The Hybrid III head form and neck were attached to a sliding table. The impacting surface, which consisted of a modular elastomer programmer pad, was attached to a plunger arm which thrust it toward the stationary head form via spring loading.

For the unrestrained head and neck conditions, the Hybrid III neck was attached to the head form but not to the sliding table; the neck simply rested on the table against a wooden wedge. A nylon rope tied around the base of the neck caught the head form shortly after impact took place. Again, the neck was used to maintain a constant mass

across the impact conditions. This set-up allowed the head form to move freely upon impact (Figure 7).



Figure 7. Set-up for the unrestrained head and neck trials on the linear impactor system. The Hybrid III head form and neck were not attached to the sliding table. The impacting surface, which consisted of a modular elastomer programmer pad, was attached to a plunger arm which thrust it toward the stationary head form via spring loading.

The velocity of the system was calculated a few millimeters prior to impact by measuring the amount of time taken for a small flag of 0.2525 m to pass in front of an infrared beam. The length and time were then used to calculate velocity. Time data were gathered by Measurement and Automation Explorer (National Instruments, Austin, TX), a software program on the computer to which the linear impactor was connected. Velocity data were processed in BioProc2 (Robertson, 2009), and an error of $\pm 2\%$ was accepted for each inbound velocity.

Hybrid III Head Form

A 50th-percentile Hybrid III head form equipped with nine single-axis accelerometers was used in this study (mass = 4.59 ± 0.01 kg; First Technology Safety

Systems [FTSS], Plymouth, MI). The head form is a mechanical device used in car crash and other impact testing. It was designed to replicate the response of the average adult male human head for forehead impacts and is the most commonly used and most sophisticated device of its nature available (Mertz, 1985). It is composed of a hollow aluminum core, and a vinyl skin-like covering complete with facial features. A photograph of the Hybrid III head form can be seen in Figure 8.

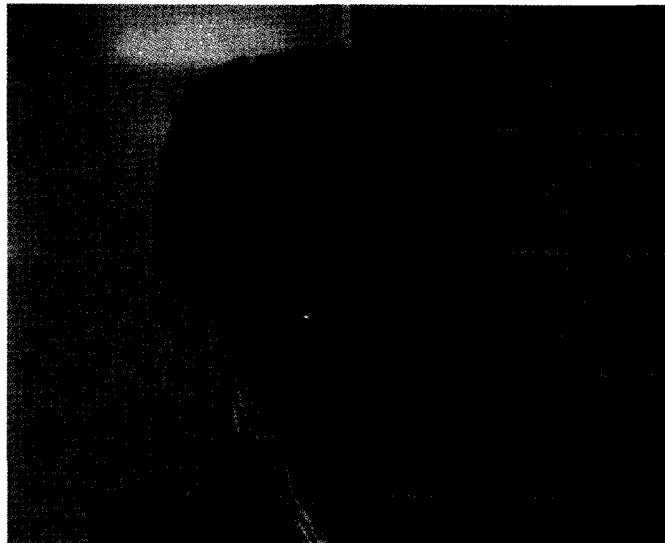


Figure 8. The Hybrid III head form attached to a Hybrid III compliant neck.

Compliant & Rigid Necks

In the impact conditions where the head form was attached to a compliant neck, a human-like Hybrid III mechanical neck composed of segmented rubber and aluminum discs was used (mass = 1.54 kg; Denton ATD, Milan, OH). The head-neck complex provided an accurate simulation of the dynamic response of the head undergoing flexion and extension (FTSS, 2009). A photograph of the neck can be seen in Figure 9.

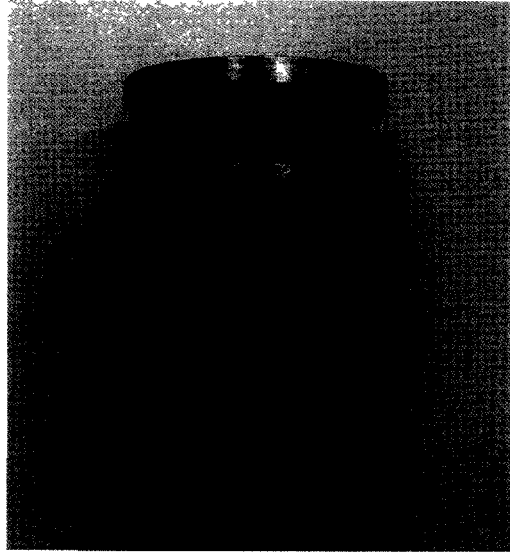


Figure 9. The Hybrid III compliant neck.

For the rigid neck conditions, a custom neck made of aluminum by the University of Ottawa Machine Shop was used (mass = 1.55 ± 0.01 kg). This device was not designed to replicate a human neck but rather to reproduce the neck parameters used in the North American standards. It was designed to match the Hybrid III neck in terms of height, base circumference and mass. A photograph of the rigid neck can be seen in Figure 10.

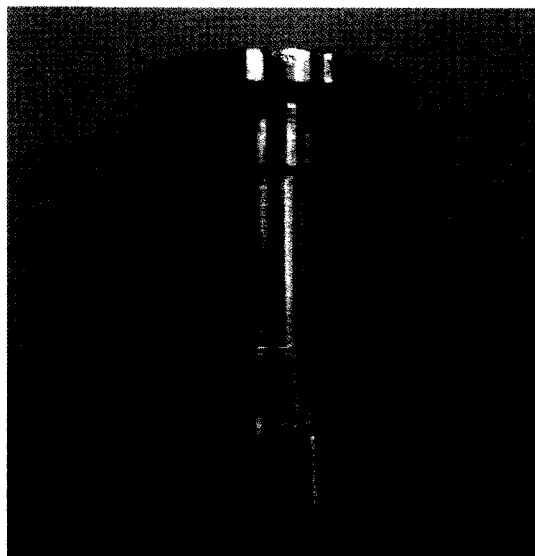


Figure 10. The custom aluminum rigid neck.

Accelerometers

As can be seen in Figure 11, nine single-axis Endevco 7264C-2KTZ-2-300 accelerometers were mounted inside the core of the head form in the 3-2-2-2 array (Padgaonkar, Krieger, & King, 1975). This set-up allowed for measurement of the three-dimensional motion of the head form. Both translational and rotational acceleration of the head form were collected upon impact at 20 kHz by a TDAS Pro Lab module (DTS, Seal Beach, CA) and filtered through a low pass 300 Hz filter, according to SAE J211-1 (2007). To maintain calibration specifications, the accelerometers are returned to the manufacturer yearly for calibration by a shaker table.

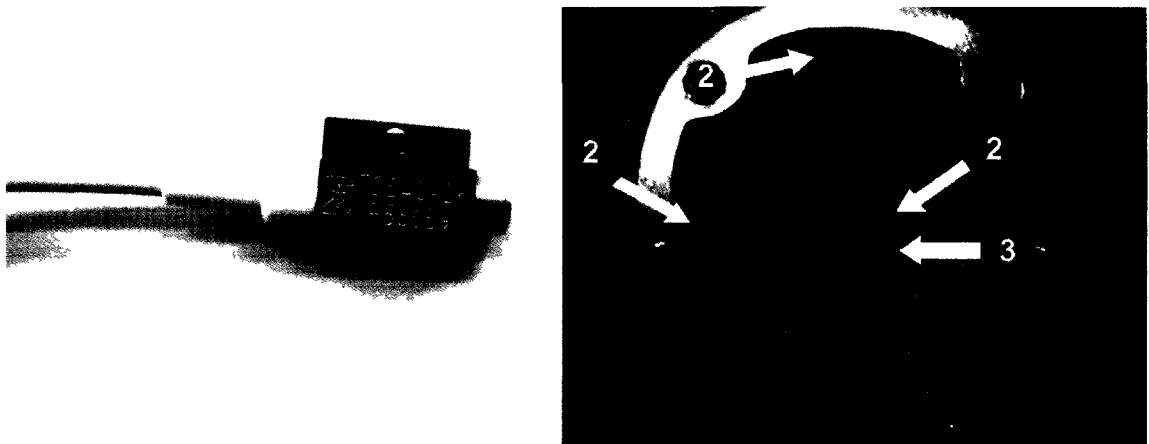


Figure 11. Endevco accelerometer (left); Accelerometers mounted in the 3-2-2-2 array in the core of the Hybrid III head form (right).

Modular Elastomer Programmer Pad

A modular elastomer programmer pad composed of a circular 1" (25 mm) thick and 6" (152 mm) diameter polyurethane rubber disc was used as the impacting surface in this study. M.E.P. pads are known for their resiliency and material property stability (Becker, 1997; Code of Federal Regulations, 2007). The same M.E.P. pad, which is returned to the manufacturer annually for calibration, was used for all testing conditions to maintain

consistency. Prior to and following data collection, the M.E.P. pad was impacted ten times with a 5 kg spherical impactor at 5.44 m/s, following the same protocol used by Spyrou, Pearsall & Hoshizaki (2000). This conditioning was carried out to determine whether deterioration occurred in the M.E.P. pad. Although the pad attenuated a small portion of the impact energy, it was a necessary component to protect the equipment since helmets, which increase the compliance of the system, were not used. Safety helmets were avoided because they have been shown to increase measurement variance.

High-Speed Camera

High-speed video data was collected at 2 kHz using a Fastcam 512 PCI video camera (HSI, Winnipeg, MB). The camera was set up orthogonal to the impact, with a resolution of 520 x 520. Spot lights aimed at the impact site were used to enhance the image quality and clarity. Video data were recorded and analyzed via Photron Motion Tools software (San Diego, CA) on the computer to which the camera was connected. The video data were synchronized with the acceleration data, and a trigger was used to initiate the action of the camera.

The video data were used to better visualize the impacts, and to determine the energy transfer taking place between the impacting surface and head form. Markers were placed on the necks to facilitate tracking. Digitization of the video data was carried out using Photron Motion Tools to calculate the velocity of the head form and impactor, both upon and immediately following the impact. These values, combined with mass, were then substituted into the formula for kinetic energy:

$$KE = \frac{1}{2} mv^2$$

where, KE is the kinetic energy of the system at a particular instant in time, m is the mass of the moving object and v is the velocity of the moving object.

3.2 RESEARCH DESIGN

The ultimate goal of this study was to compare resultant translational and rotational accelerations generated during impacts with mass A1 (158.8 kg) and mass A2 (17.1 kg) across the conditions of inbound velocity (B) and system compliance (C). The relationship between these three variables was investigated according to the following fully crossed design:

A1 - Large Mass (158.8 kg)				A2 - Small Mass (17.1 kg)			
	B1	B2	B3		B1	B2	B3
C1	B1C1	B2C1	B3C1	C1	B1C1	B2C1	B3C1
C2	B1C2	B2C2	B3C2	C2	B1C2	B2C2	B3C2
C3	B1C3	B2C3	B3C3	C3	B1C3	B2C3	B3C3

where,

A1 = 158.8 kg

A2 = 17.1 kg

B1 = 2.0 m/s

B2 = 3.0 m/s

B3 = 4.0 m/s

C1 = Rigid neck

C2 = Compliant neck

C3 = Unrestrained head and neck

3.3 TEST PROTOCOL

All of the testing was conducted under ambient conditions, with humidity, temperature and pressure values recorded. Over the course of testing, the humidity ranged from 21-30%, the temperature values were between 24-27°C and the pressure varied between 981-1003 mbHPa; values well within the operating range of the accelerometers (Endevco, 2011). Prior to collecting research data, all electronic equipment was allowed to “warm up” for 15 minutes to ensure thermal stability within the system components. In

addition, the M.E.P. pad was conditioned as outlined above. Data collection commenced immediately following the conditioning period, and no more than 7 minutes was allowed to pass between impacts to avoid variations in thermal stability and surface conditioning.

The test apparatuses were set up as shown in Figures 2, 4, 6, and 7. Three drops were carried out at each of three predetermined velocities on both of the test apparatuses, and acceleration data were collected to examine the severity of the impacts. Upon completion of the first data set, two more sets were subsequently collected. Pilot data indicated that order did not have an effect on the treatment outcomes. Accordingly, since the impact conditions were independent of each other, trial randomization was not necessary.

All impacts were directed through the head form's centre of mass in the front position, 30 ± 1 mm above the reference plane of the un-helmeted head form. The head form was set at an angle of $25 \pm 2^\circ$ from the axis of impact, following the neck position employed by Pellman and associates (2006).

3.4 DATA EXCLUSION CRITERIA

Throughout the data collection, certain trials were deemed unacceptable, and were disregarded. The only reasons for disregarding data were: i) there was a technological problem with the testing equipment and/or systems, and ii) the set-up was not consistent with previous trials.

3.5 FILTERING

In this study, processing algorithms were used to calculate rotational acceleration from the translational acceleration data. Noisy signals can cause the

algorithms to misinterpret the data (Withnall & Keown, 2003). To determine which filter to apply to the data, a series of tests were conducted. First, a frequency spectral analysis was performed which examined the frequency content of the signal to determine where the natural head form resonance was occurring. For each of the nine accelerometer channels, the majority of the frequency content (95th percentile) was below 150 Hz. A further power analysis indicated that 99% of frequency content was below 80 Hz. BioProc2 was then used to determine the effects of filtering at such low frequencies. It was found that at both 80 Hz and 150 Hz, the peak data was highly attenuated. Ultimately, a 300 Hz low pass filter was applied to the data, according to SAE J211-1 (2007), because at this frequency the noise was removed with minimal effect on the peaks.

3.6 ANALYSIS

To evaluate the influence of impact mass, inbound velocity and system compliance on the dynamic response of a Hybrid III head form, two separate 2 x 3 x 3 analyses of variance (ANOVAs) were carried out: one for peak translational acceleration and one for peak rotational acceleration. This allowed for examination of the relationships between the independent variables for each dependent variable. Statistical analyses were performed using the predictive analysis software package SPSS 16.0 (SPSS Inc., Chicago, IL).

CHAPTER 4. RESULTS

Two three-way repeated measures ANOVAs were conducted in the investigation of the effects of mass, inbound velocity and system compliance on the dynamic response of a Hybrid III head form subjected to impact. The dependent variables were analyzed separately based upon the results of a bivariate Pearson Correlation which tested for independence of the dependent variables. The correlation revealed a strong positive relationship between peak resultant translational and peak resultant rotational acceleration ($r = 0.871$, $p < 0.01$) indicating that the variables were correlated but not completely independent in the conditions examined.

Two levels of impact mass (17.1 kg and 158.8 kg), three levels of inbound velocity (2.0 m/s, 3.0 m/s and 4.0 m/s) and three levels of system compliance (rigid neck, compliant neck and unrestrained head and neck) were examined, giving a total of 18 impact conditions. Mean peak resultant translational and rotational accelerations were assessed using a between subjects (i.e., head forms) design, and it was hypothesized that no significant difference would exist between the impact masses across each level of inbound velocity and system compliance. A significance level of 0.05 was used for all analyses. The translational acceleration values along with standard deviations (SD) for each impact condition can be found in Table 4. Similar descriptive statistics for rotational acceleration can be found in Table 5.

Table 4. Mean peak resultant translational acceleration (g) and standard deviation values for two impact masses (17.1 kg and 158.8 kg) across three inbound velocities (2.0, 3.0, and 4.0 m/s) and three system compliances (rigid neck, compliant neck, unrestrained head and neck).

System Compliance (neck)	Inbound Velocity (m/s)	17.1 kg (Linear Impactor)		158.8 kg (Monorail)	
		Mean Peak Resultant a (g)	SD	Mean Peak Resultant a (g)	SD
Rigid	2.0	57.08	0.61	74.53	2.14
	3.0	99.70	0.26	120.61	2.27
	4.0	146.34	1.23	172.11	1.61
Compliant	2.0	59.51	0.46	76.79	0.66
	3.0	105.47	0.44	136.78	0.60
	4.0	152.27	0.78	190.44	1.71
Unrestrained	2.0	59.61	0.86	76.83	1.14
	3.0	104.76	0.68	131.07	1.21
	4.0	151.73	0.62	188.41	1.26

Table 5. Mean peak resultant rotational acceleration (rad/s²) and standard deviation values for two impact masses (17.1 kg and 158.8 kg) across three inbound velocities (2.0, 3.0, and 4.0 m/s) and three system compliances (rigid neck, compliant neck, unrestrained head and neck).

System Compliance (neck)	Inbound Velocity (m/s)	17.1 kg (Linear Impactor)		158.8 kg (Monorail)	
		Mean Peak Resultant α (rad/s ²)	SD	Mean Peak Resultant α (rad/s ²)	SD
Rigid	2.0	1851.44	60.60	2431.89	61.57
	3.0	2969.11	182.81	4210.22	84.93
	4.0	3854.67	91.65	5953.22	128.72
Compliant	2.0	2998.11	37.65	3749.22	45.19
	3.0	5469.11	55.52	6751.44	90.28
	4.0	7717.33	59.04	9023.11	62.32
Unrestrained	2.0	3102.56	63.65	3916.56	51.10
	3.0	5617.67	36.56	6785.56	127.88
	4.0	8018.22	50.07	9893.11	101.90

The analysis of variance statistical test functions under the assumption that the error variances of the dependent variables are equal across all groups investigated. Equality of error variances is evaluated by Levene's test, such that when $p < 0.05$, the assumption is violated. Upon further analysis, if the data are normally distributed, the analysis can be accepted as valid, regardless of the equality of variances violation (George

& Mallery, 2007). In this analysis, Levene's test revealed that the mean peak resultant translational acceleration ($F(17,144) = 7.618, p < 0.001$) and mean peak resultant rotational acceleration ($F(17,144) = 3.386, p < 0.001$) lacked homogeneity across groups. Further evaluation of the descriptive statistics revealed that both peak resultant translational acceleration and peak resultant rotational acceleration were positively skewed, with values of 0.178 and 0.435, respectively, while both accelerations had negative kurtoses, measuring -1.180 and -0.927, respectively. Each of these measures fell within the ± 2.0 value accepted for normally distributed data (George & Mallery, 2007). Therefore, normality within the distribution of the data was assumed, and despite the results of Levene's test, the results of the ANOVA were accepted as valid.

Analyses of variance revealed significant main effects for impact mass on peak resultant translational acceleration ($F(1,144) = 19,460, p < 0.001, \text{partial } \eta^2 = 0.993$) and on peak resultant rotational acceleration ($F(1,144) = 8,402, p < 0.001, \text{partial } \eta^2 = 0.983$). Main effects were also significant for inbound velocity on peak resultant translational acceleration ($F(2,144) = 97,390, p < 0.001, \text{partial } \eta^2 = 0.999$) and on peak resultant rotational acceleration ($F(2,144) = 35,590, p < 0.001, \text{partial } \eta^2 = 0.998$) as well as for system compliance on peak resultant translational acceleration ($F(2,144) = 807.691, p < 0.001, \text{partial } \eta^2 = 0.918$) and on peak resultant rotational acceleration ($F(2,144) = 15,950, p < 0.001, \text{partial } \eta^2 = 0.996$). Significant two-way interactions were found between impact mass and inbound velocity for peak resultant translational acceleration ($F(2,144) = 649.046, p < 0.001, \text{partial } \eta^2 = 0.900$) and peak resultant rotational acceleration ($F(2,144) = 500.788, p < 0.001, \text{partial } \eta^2 = 0.874$). Similarly, two-way interactions were significant between impact mass and system compliance for peak resultant translational acceleration

($F(2,144) = 148.244, p < 0.001, \text{partial } \eta^2 = 0.673$) and peak resultant rotational acceleration ($F(2,144) = 20.714, p < 0.001, \text{partial } \eta^2 = 0.223$) as well as between inbound velocity and system compliance for peak resultant translational acceleration ($F(4,144) = 108.144, p < 0.001, \text{partial } \eta^2 = 0.750$) and peak resultant rotational acceleration ($F(4,144) = 1,280, p < 0.001, \text{partial } \eta^2 = 0.973$). Finally, significant three-way interactions were found between all three independent variables for peak resultant translational acceleration ($F(4, 144) = 45.499, p < 0.001, \text{partial } \eta^2 = 0.558$) and peak resultant rotational acceleration ($F(4, 144) = 46.257, p < 0.001, \text{partial } \eta^2 = 0.562$). The observed power for each of these main effects and interactions was 1.000, computed using an alpha value of 0.05 and indicating that in any sample population with the same sample size the chance of finding significant differences between each group was 100%.

Post hoc Bonferroni pairwise comparisons were conducted to determine where significant differences existed within each variable when significant interactions were found. All analyses were examined separately.

TWO-WAY INTERACTIONS: IMPACT MASS AND INBOUND VELOCITY

For the two impact masses, the highest mean peak resultant translational accelerations were generated during the 4.0 m/s impacts, followed by the 3.0 m/s impacts and ending with the 2.0 m/s impacts. Post hoc tests using the Bonferroni adjustment revealed significant differences for Impact Mass A (158.8 kg), between all inbound velocities ($p < 0.001$). Similarly, significant differences existed between all inbound velocities for Impact Mass B (17.1 kg; $p < 0.001$).

The same trend was found for mean peak resultant rotational accelerations; the greatest values were generated during the 4.0 m/s impacts, again followed by the 3.0 m/s

impacts and ending with the 2.0 m/s impacts for both impact masses. Post hoc tests using the Bonferroni adjustment revealed that for both Impact Mass A (158.8 kg) and Impact Mass B, significant differences existed between all inbound velocities ($P < 0.001$). The relationship between impact mass, inbound velocity and mean peak resultant translational acceleration can be found in Figure 12, along with the relationship between impact mass, inbound velocity and mean peak resultant rotational acceleration.

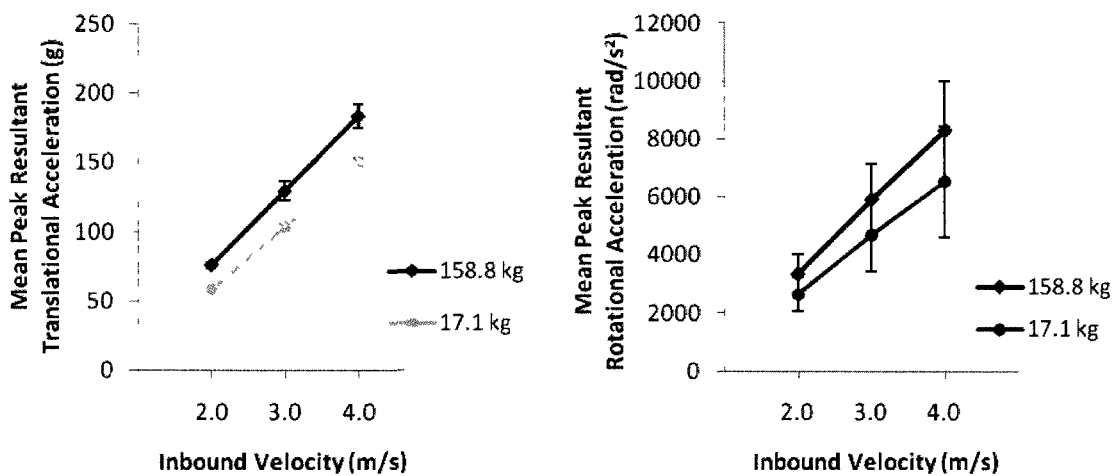


Figure 12. Comparison of the mean peak resultant translational acceleration (g) and mean peak resultant rotational acceleration (rad/s²) values for Impact Mass A (158.8 kg) and Impact Mass B (17.1 kg) impacted at inbound velocities of 2.0, 3.0, and 4.0 m/s across all conditions of system compliance (rigid neck, compliant neck, unrestrained head and neck), where $n = 27$ for each impact mass and inbound velocity interaction. Error bars represent standard deviations.

TWO-WAY INTERACTIONS: IMPACT MASS AND SYSTEM COMPLIANCE

The two-way interaction trends between impact mass and system compliance were the same for both impact masses: the highest mean peak resultant translational accelerations were generated with the compliant neck, followed by the unrestrained head and neck, and ending with the rigid neck. Post hoc analyses using the Bonferroni adjustment revealed significant differences between all system compliances for Impact

Mass A (300 kg; $p < 0.001$). Conversely, significant differences were found between all system compliances ($p < 0.001$) for Impact Mass B (17.1 kg), except between the compliant neck and unrestrained head and neck condition ($p = 0.157$).

A different trend was found for mean peak resultant rotational accelerations wherein the greatest values were generated with the unrestrained head and neck, followed by the compliant neck and ending with the rigid neck, for both impact masses. Again, separate post hoc Bonferroni tests were conducted for each impact mass to determine where significant differences existed. For Impact Mass A (158.8 kg) and Impact Mass B (17.1 kg), significant differences existed between all system compliances ($P < 0.001$). The relationship between impact mass, system compliance and mean peak resultant translational acceleration, as well as the relationship between impact mass, system compliance and mean peak resultant rotational acceleration can be found in Figure 13.

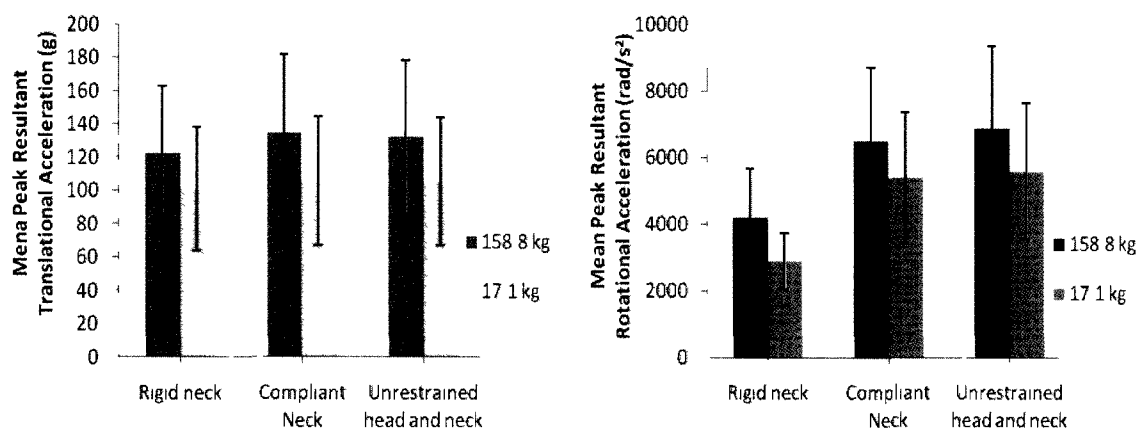


Figure 13. Comparison of the mean peak resultant translational acceleration (g) and mean peak resultant rotational acceleration (rad/s²) values for Impact Mass A (158.8 kg) and Impact Mass B (17.1 kg) with system compliances ranging from rigid neck to compliant neck to unrestrained head and neck, across all conditions of inbound velocities of (2.0, 3.0, and 4.0 m/s), where $n = 27$ for each impact mass and system compliance interaction. Error bars represent standard deviations.

TWO-WAY INTERACTIONS: SYSTEM COMPLIANCE AND INBOUND VELOCITY

For all three system compliances, the highest mean peak resultant translational accelerations were generated during the 4.0 m/s impacts, followed by the 3.0 m/s impacts and ending with the 2.0 m/s impacts. Post hoc Bonferroni analyses revealed significant differences between all inbound velocities ($p < 0.001$) for System Compliance A (rigid), System Compliance B (compliant) and System Compliance C (unrestrained).

The same trend was found for mean peak resultant rotational accelerations such that the highest values were generated at 4.0 m/s, and the acceleration values decreased with decreasing velocity for all three system compliances. Post hoc tests using a Bonferroni adjustment indicated that for System Compliance A, System compliance B and System Compliance C, significant differences existed between all inbound velocities ($P < 0.001$). The relationship between system compliance, inbound velocity and mean peak resultant translational acceleration, along with the relationship between system compliance, inbound velocity and mean peak resultant rotational acceleration can be found in Figure 14.

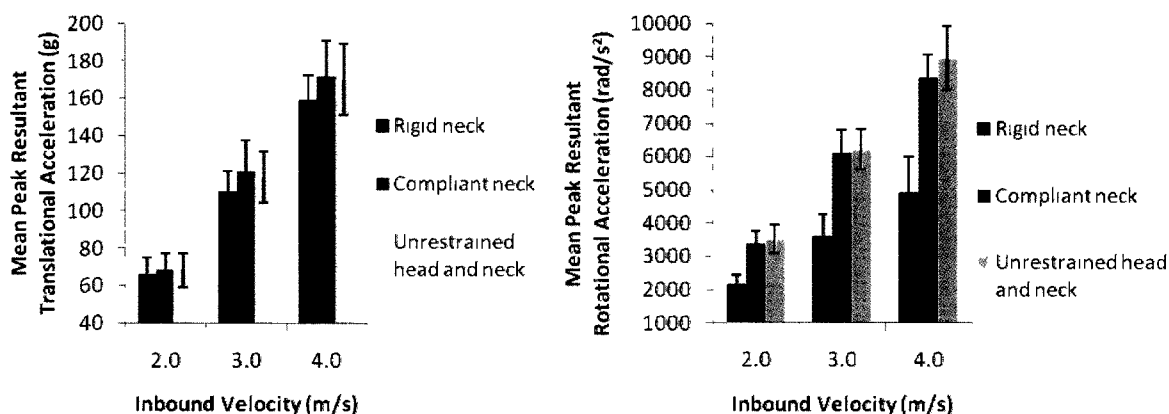


Figure 14. Comparison of the mean peak resultant translational acceleration (g) and mean peak resultant rotational acceleration (rad/s²) values for System Compliance A (rigid neck), System Compliance B (compliant neck) and System Compliance C (unrestrained head and neck) impacted at inbound velocities 2.0, 3.0, and 4.0 m/s, across both conditions of impact mass (158.8 kg, 17.1 kg), where $n = 18$ for each interaction. Error bars represent standard deviations.

THREE-WAY INTERACTIONS: IMPACT MASS, INBOUND VELOCITY, SYSTEM COMPLIANCE

Separate post hoc Bonferroni analyses were conducted for each impact mass to determine where significant differences existed across the three inbound velocities and the three system compliances for mean peak resultant translational acceleration. Significant differences were found between the two impact masses ($p < 0.001$) across all three levels of inbound velocity (2.0, 3.0 and 4.0 m/s) and system compliance (rigid neck, compliant neck and unrestrained head and neck), as can be seen in Table 6 below.

Table 6. Significant differences in mean peak resultant translational acceleration (denoted by an asterisk (*)) between Impact Mass A (MA) and Impact Mass B (MB) across all conditions of inbound velocity (2.0, 3.0 and 4.0 m/s) and system compliance (rigid neck, compliant neck and unrestrained head and neck), such that $p < 0.001$.

	2.0 m/s	3.0 m/s	4.0 m/s
Rigid neck	MA – MB*	MA – MB*	MA – MB*
Compliant neck	MA – MB*	MA – MB*	MA – MB*
Unrestrained head and neck	MA – MB*	MA – MB*	MA – MB*

To determine where significant differences existed in terms of peak resultant rotational acceleration, separate post hoc tests using a Bonferroni adjustment were once again conducted between the two impact masses across all three inbound velocities and system compliances. Significant differences were found between the two impact masses for each condition ($P < 0.001$), as can be seen in Table 7 below.

Table 7. Significant differences in mean peak resultant rotational acceleration (denoted by an asterisk (*)) between Impact Mass A (MA) and Impact Mass B (MB) across all conditions of inbound velocity (2.0, 3.0 and 4.0 m/s) and system compliance (rigid neck, compliant neck and unrestrained head and neck), such that $p < 0.001$.

	2.0 m/s	3.0 m/s	4.0 m/s
Rigid neck	MA – MB*	MA – MB*	MA – MB*
Compliant neck	MA – MB*	MA – MB*	MA – MB*
Unrestrained head and neck	MA – MB*	MA – MB*	MA – MB*

The effect of the three-way interaction between impact mass, inbound velocity and system compliance on mean peak resultant translational acceleration and mean peak resultant rotational acceleration can be seen graphically in Figures 15 and 16.

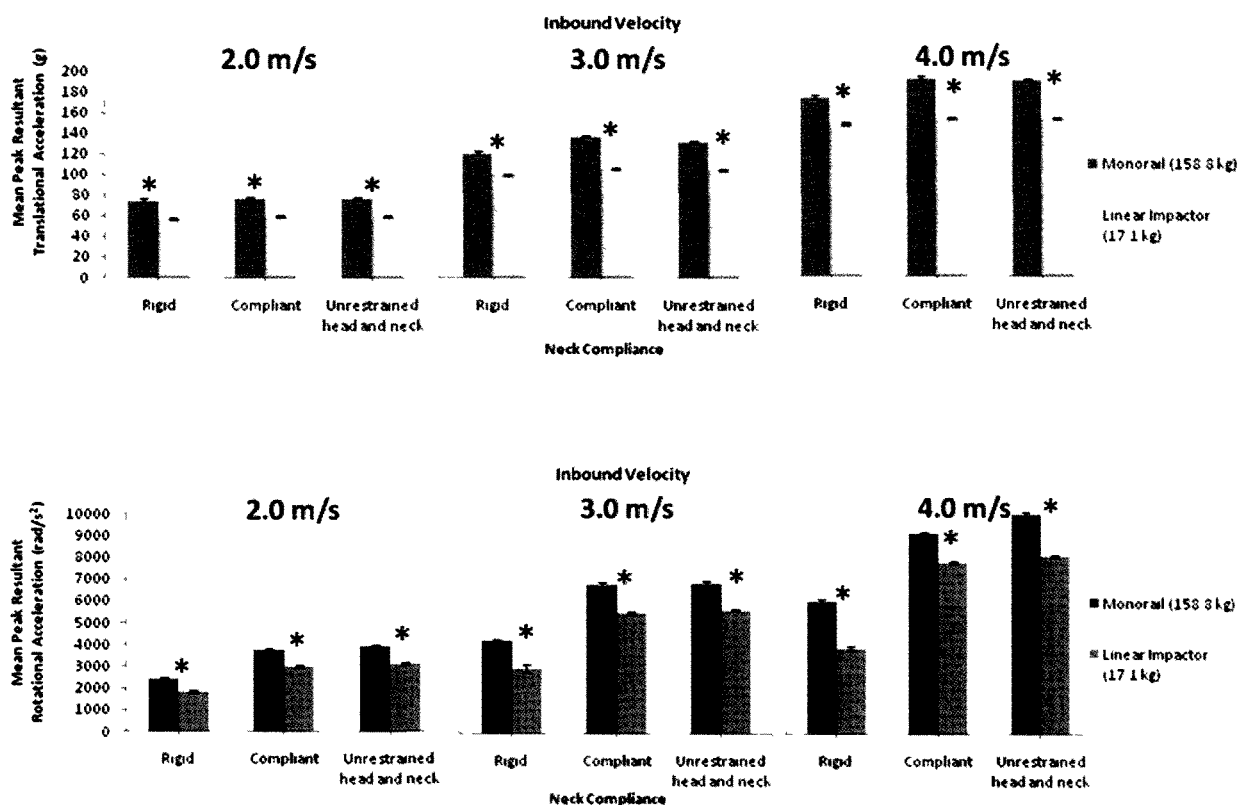


Figure 15. Comparison of the mean peak resultant translational acceleration (g) and mean peak resultant rotational acceleration (rad/s²) values for Impact Mass A (158.8 kg) and Impact Mass B (17.1 kg) impacted using three conditions of system compliance (rigid neck, compliant neck and unrestrained head and neck) across inbound velocities of 2.0, 3.0, and 4.0 m/s, where n = 9 for each three-way interaction. Error bars represent standard deviations; asterisks denote significance.

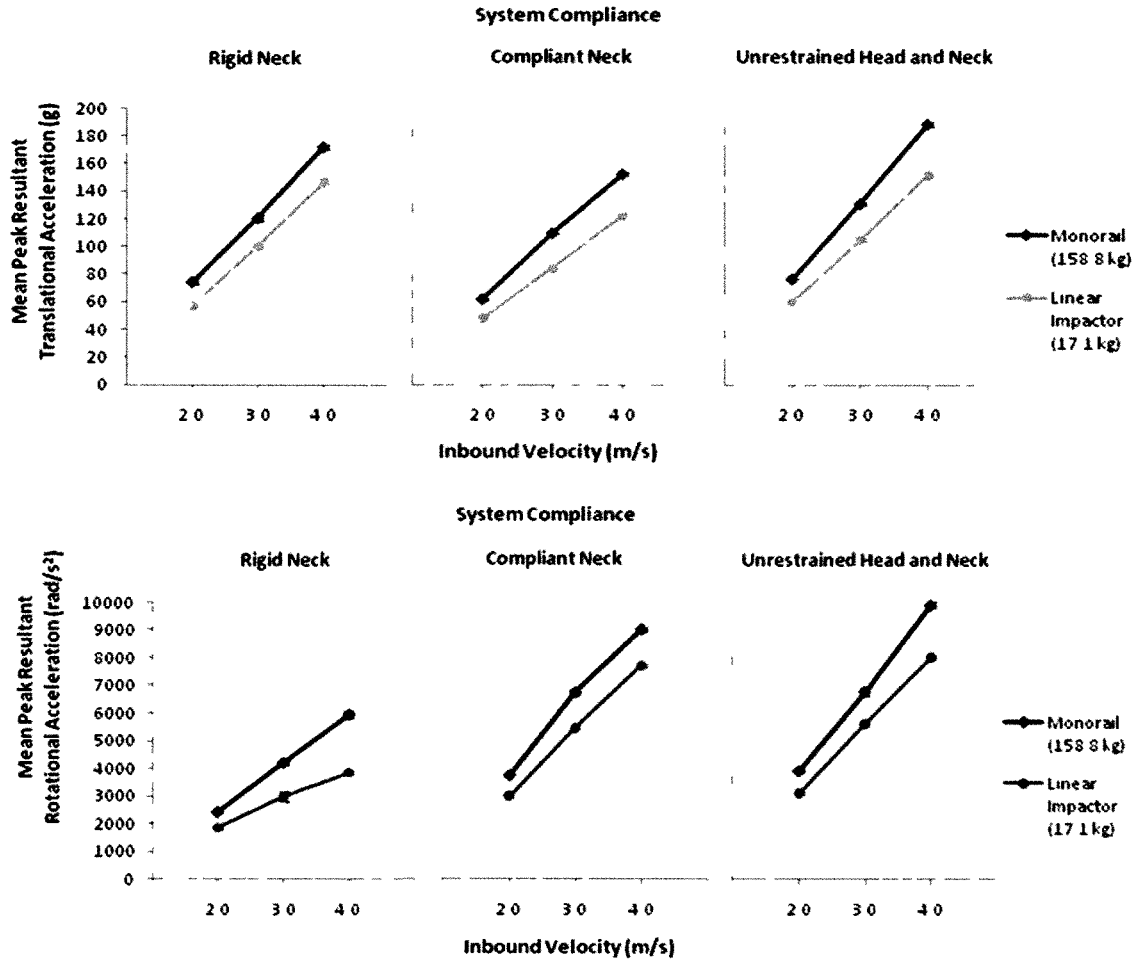


Figure 16. Comparison of the mean peak resultant translational acceleration (g) and mean peak resultant rotational acceleration (rad/s²) values for Impact Mass A (158.8 kg) and Impact Mass B (17.1 kg) impacted at inbound velocities 2.0, 3.0, and 4.0 m/s, across conditions of system compliance (rigid neck, compliant neck and unrestrained head and neck), where $n = 9$ for each three-way interaction. Error bars represent standard deviations.

In summary, under the same conditions of inbound velocity and system compliance, the monorail vertical drop tower led to greater acceleration values both translationally and rotationally in all cases. In addition, as inbound velocity was increased, both peak resultant translational and rotational accelerations also increased. Finally, the rigid neck conditions always generated the lowest translational and rotational accelerations, when compared to the compliant neck and unrestrained head and neck conditions.

CHAPTER 5. DISCUSSION

The purpose of this study was to examine the effects of impact mass, inbound velocity and system compliance on the dynamic response of a Hybrid III head form. According to the literature, this study was the first of its kind to analyze the relationship between these variables as they are related to the improvement of helmet safety standards.

The results presented in Chapter 4 indicate that impact mass, inbound velocity and system compliance all had a significant influence on the dynamic response of a human head surrogate. In addition, the two-way and three-way interactions between all the variables significantly affected the head form's dynamic response. Consequently, to define the relationship between these factors as it is associated with the performance of sport helmets, each relationship must be considered.

Significant differences were found between the two impact masses for each condition of inbound velocity and system compliance, leading to a rejection of all eighteen null hypotheses listed above. This indicates that the two test systems which were represented by the masses are indeed fundamentally different. In each case, as impact mass was increased (from 17.1 kg to 158.8 kg), the resulting peak translational and rotational accelerations also increased. This may be explained by the differences in the kinetic energies of the systems. Kinetic energy (KE) in any object is based on the mass (m) and velocity (v) of the object, such that $KE = \frac{1}{2}mv^2$. To examine the kinetic energy values of each condition investigated in this study, the inbound and outbound velocities of the impactor and head form were digitized from high-speed video data.

The kinetic energy analysis revealed that for the horizontal linear impactor system, the amount of energy transfer varied with each neck compliance and inbound velocity

condition. At 2.0 m/s, energy was transferred according to the following trend: the greatest amount of energy was transferred away from the impactor during the rigid neck trials, followed by the compliant neck trials, and the least amount of energy was transferred when the unrestrained head and neck system was employed. At 3.0 and 4.0 m/s, the trend was different than at 2.0 m/s: again, the greatest amount of energy was transferred away from the impactor during the rigid neck trials; however, this was followed by the unrestrained head and neck system and then the compliant neck trials. The inbound and outbound energy data for one trial of each condition on the linear impactor can be found in Table 8, while the energy transfer values can be found in Table 9.

Table 8. Inbound and outbound kinetic energy values for the head form and impacting arm under each impact condition on the horizontal linear impactor (Mass A – 17.1 kg).

		Kinetic Energy (J)			
		Inbound Impactor	Outbound Impactor	Inbound Head form	Outbound Head form
2.0 m/s	Rigid neck	36.15	0.1090	0	8.744
	Compliant neck	38.31	7.757	0	2.188
	Unrestrained head and neck	37.14	9.284	0	17.85
3.0 m/s	Rigid neck	80.14	0.1099	0	13.36
	Compliant neck	74.32	15.83	0	11.68
	Unrestrained head and neck	95.92	16.65	0	38.74
4.0 m/s	Rigid neck	143.80	0	0	20.88
	Compliant neck	131.52	18.58	0	18.91
	Unrestrained head and neck	120.20	15.69	0	50.33

Table 9. Kinetic energy transfer data for the head form and impacting arm under each impact condition on the horizontal linear impactor (Mass A – 17.1 kg).

		Kinetic Energy (J)			
		Total Inbound	Total Outbound	Total Energy Transfer (%)	Energy Transferred to head-neck system (%)
2.0 m/s	Rigid neck	36.15	0.1090	99.7	24.49
	Compliant neck	38.31	7.757	79.75	25.96
	Unrestrained head and neck	37.14	9.284	75.00	73.07
3.0 m/s	Rigid neck	80.14	0.1099	99.86	16.81
	Compliant neck	74.32	15.83	78.70	37.02
	Unrestrained head and neck	95.92	16.65	82.64	57.75
4.0 m/s	Rigid neck	143.80	0	0	14.52
	Compliant neck	131.52	18.58	85.87	28.51
	Unrestrained head and neck	120.20	15.69	86.95	54.93

According to the Law of Conservation of Energy, the total amount of energy in a closed system must remain constant. Thus, the remaining percentage of energy unaccounted for by the impactor and the head form must have gone elsewhere within the system. Upon each impact, there was a loud “bang” sound indicating that a portion of the energy was converted to sound energy. In addition, the video data demonstrated that the compliant neck absorbed a portion of the energy. This indicates that some of the energy was converted to deformation energy. At 2 m/s, the impact caused a bending in the neck at an angle of 32°. With increasing velocities, the angle of neck extension also increased: at 3 m/s the neck bent at an angle of 42° and at 4 m/s the neck extension angle was 56°. Therefore, as velocity increased, it is possible that more energy was absorbed by the compliant neck. For the unrestrained head and neck, the head form was able to move freely post-impact, and although the neck was not tied down at its base, it underwent a measureable degree of extension. At 2.0 m/s, 7° of extension occurred in the neck,

followed by 11° and 13° at 3.0 m/s and 4.0 m/s, respectively. Again, with increasing velocity, the amount of energy absorbed by the neck may have increased.

In the case of the rigid neck, no bending took place, however a large spike occurred in the data following the impact, implying that a large torque was placed on the system. This torque occurred in the sliding table to which the head-neck complex was attached, as confirmed by the video data. It is clearly visible that the resulting motion was not simply linear translation along an axis parallel to the motion of the impacting arm: instead, there was an upward component included in the motion, which accounts for a portion of the energy. In addition, the system was not completely rigid. Slight compliances were introduced by the nodding joints and by the vinyl skin covering the head form, therefore it is likely that some of the unaccounted for energy was relocated to these areas as well.

Conversely, on the monorail drop tower, the energy was entirely transferred out of the head form, as indicated by a rebound. Immediately prior to impact, the head form had a certain velocity in the downward direction. Upon impact, the velocity was reduced to zero momentarily as the head form contacted the M.E.P. pad, and was followed by a bounce caused by elastic deformation of the impact surface and the head form. The energy wasn't entirely transferred to the impacting surface, however. A portion of the energy was also converted to sound energy, while another portion went into the neck, although the exact amounts are unknown. Further, since the head form was released from a different height than it ended at, the impacts on the monorail also experienced a change in potential energy.

In scrutinizing the acceleration data, it is clear that the resulting translational accelerations were less sensitive to the changes in the independent variables than rotational acceleration. When the conditions were varied, the changes in peak resultant translational

acceleration were not as pronounced as those seen in rotational acceleration. The differences in sensitivity between the dependent variables can be seen in Figure 17. While the translational acceleration only changes by a few dozen gs between conditions, the rotational acceleration changes by thousands of units. The mild traumatic brain injury thresholds proposed by Zhang, Yang and King (2004) have been added to the figures to highlight the brain injury risk associated with the impacts examined in this study. From this graph it is clear that low-velocity impacts are well within the range of causing concussion.

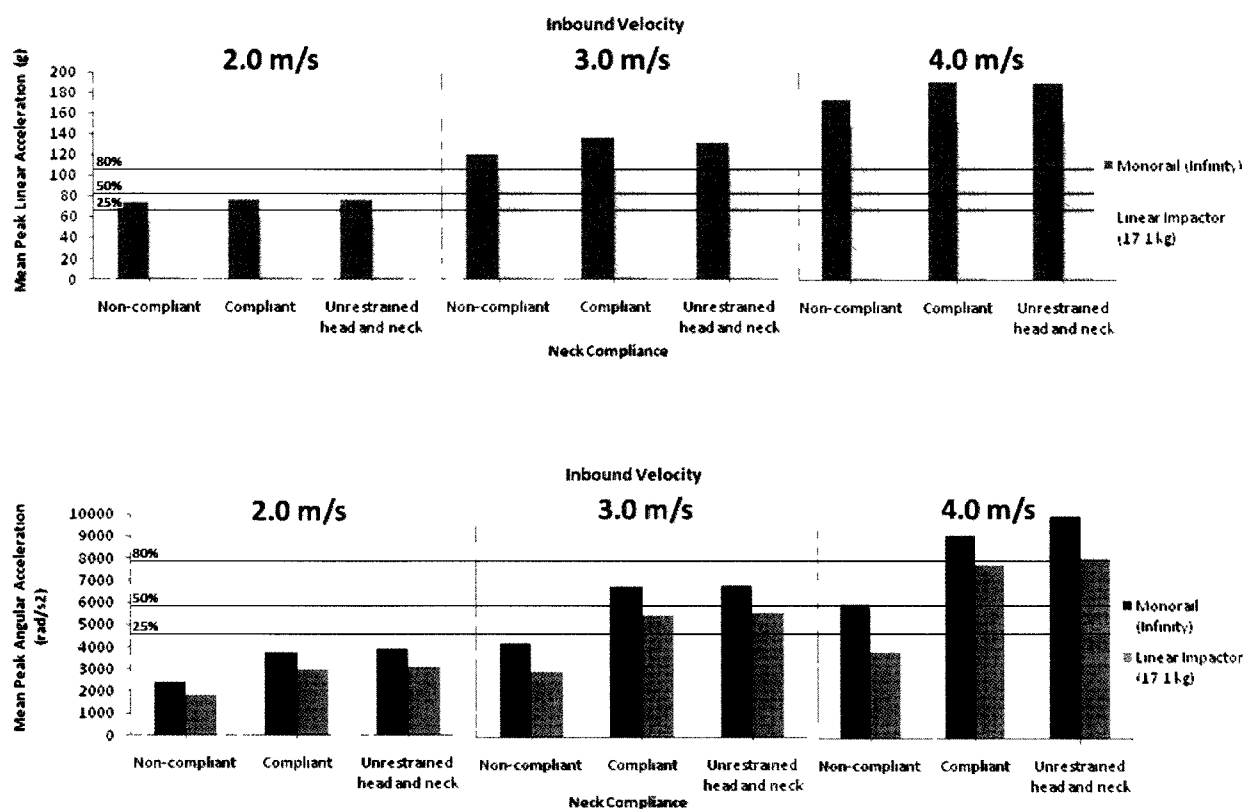


Figure 17. Comparison of the mean peak resultant translational accelerations (g) and mean peak resultant rotational accelerations (rad/s²) between three inbound velocities (2.0, 3.0, and 4.0 m/s), on two impact test systems (Monorail - 158.8 kg; Linear impactor - 17.1 kg) impacted using three different neck conditions (rigid neck, compliant neck and unrestrained head and neck), where n = 9 for each condition. The lines represent 25%, 50% and 80% probability of sustaining an mTBI.

In terms of translational acceleration, there is a 25% chance that every neck condition on the monorail (denoted by the darker green colour) will lead to a concussion at 2.0 m/s. On the contrary, the impacts conducted on the linear impactor system (light green) at 2.0 m/s had less than 25% probability of leading to mTBI. At 3.0 m/s, all of impact conditions reached the 80% probability threshold, except the rigid neck on the linear impactor. At 4.0 m/s, every impact on both systems was well above the 80% probability threshold for translational acceleration, indicating that the risk of suffering traumatic brain injury increases with increasing velocity.

With respect to rotational acceleration, none of the 2.0 m/s impacts fell within the 25% probability range for suffering mTBI. At 3.0 m/s, the rigid neck conditions generated accelerations below the 25% probability threshold, while the compliant neck and unrestrained head and neck conditions surpassed the 25% threshold on the linear impactor, and the 50% probability threshold on the monorail. At 4.0 m/s, when the rigid neck was employed, the associated mTBI risk was minimized. On the linear impactor, the risk was the lowest, with less than 25% chance of suffering mTBI. On the monorail, there was a 50% probability that an impact would lead to concussion. For the compliant neck impacts at 4.0 m/s, the results on the monorail imply that such an impulse will lead to brain injury 80% of the time, while similar impacts on the horizontal linear impactor have slightly lower odds. For the unrestrained head and neck trials on both impact systems, the 80% probability threshold was surpassed. Again, these results imply that at greater velocities, the risk of suffering mTBI is increased. The results imply that the chances of suffering mTBI are reduced when a stiff neck is employed.

It is important to recognize that only front impacts were examined in this study. Extrapolation of these findings to other impact locations is discouraged, as research has shown that direction of impact has a considerable effect on the intracranial response (Kleiven, 2003).

In addition to sensitivities observed in the acceleration data, this study also revealed that inbound velocity has a significant effect on the dynamic response of a hybrid III head form. As velocity was increased (from 2.0 to 3.0 to 4.0 m/s), both mean peak resultant translational and mean peak resultant rotational acceleration also increased. The results demonstrate the importance of including low-velocity impacts in the certification of sport helmets, since the risk of suffering traumatic brain injury at these speeds is high.

These results were expected based on the findings of various researchers. In 2008, Gimbel demonstrated that on a monorail vertical drop system, as velocity increased so did the mean peak resultant translational acceleration of a head form. Although a few differences existed between Gimbel's protocol and the current protocol in terms of head form (magnesium vs. Hybrid III), accelerometers (one uniaxial vs. 3-2-2-2 array of 9 accelerometers) and impacting surface (foam vs. M.E.P. pad), the similarity in the trends is consistent.

Rousseau & Hoshizaki (2009) demonstrated similar findings on a horizontal linear impactor system. As with the current protocol, both translational and rotational acceleration were measured on a Hybrid III head form. Once again, the impacting surface varied from the one employed in the current study: a nylon cap outfitted with foam was used as opposed to an M.E.P. pad.

The current study also demonstrated that system compliance has a significant effect on the dynamic response of a hybrid III head form; however, this relationship was more complex. It was expected that as system compliance was increased (from rigid neck to compliant neck to unrestrained head and neck), there would be a trade-off between the accelerations such that mean peak resultant translational acceleration would decrease while mean peak resultant rotational acceleration would increase, as demonstrated by Rousseau (2008). This is because when a rigid (or non-compliant) head-neck system is impacted, the resulting motion of the head form should be entirely linear, as no mobility is possible in the neck joint. Minimal rotational accelerations may result if the system is not completely rigid. When the neck is switched to a compliant neck with the ability to bend, the resulting translational acceleration should decrease (since it was at a maximum in the previous condition, it can only decrease), and the rotational acceleration should increase (since it was at a minimum in the initial condition). Finally, when an unrestrained head-neck system is employed, the head form has the freedom to move in any direction following impact. This should lead to a further increase in rotational acceleration, and a further decrease in translational acceleration, as the head form is no longer restrained to motion in one plane.

This relationship was demonstrated by Rousseau (2008) at high velocities (5, 7, 9 m/s) using necks with different compliances than those employed by this study. Rousseau reported that increases in neck compliance resulted in decreases in translational acceleration and increases in rotational acceleration, indicating an inverse relationship between translational and rotational acceleration.

In this study, the following, non-linear trend was observed for mean peak translational acceleration: the rigid neck generated the lowest acceleration values, followed by the unrestrained head and neck, and ending with the compliant neck creating the greatest translational accelerations. Although the finding that translational acceleration was reduced when a stiff neck was employed is contrary to what Rousseau (2008) found, it is congruent with the findings of other researchers (Cantu & Mueller, 2003; Levy, Ozgur, Berry, Aryan & Apuzzo, 2004). This unexpected, non-linear trend may have been due to the equipment used and the protocol followed. For instance, the rigid system was not completely rigid. To match the rest of the conditions in all respects other than neck compliance, minor compliances were inevitably introduced into the system at the nodding joint and via the vinyl skin covering the head form.

Conversely, rotational accelerations followed the expected trend, such that as compliance increased, mean peak resultant rotational acceleration also increased. Although, it should be noted that the differences between the compliant neck and unrestrained head-neck values were minimal.

Researchers and helmet developers are faced with attempting to refine the performance capabilities of helmets to include mild traumatic brain injury. In order to do so, more sophisticated means of testing may be required; which could include different testing devices, such as horizontal linear impactors. Therefore, it is important to characterize the various mechanisms of injury seen in sports and recreational activities to ensure that the safety standards are accommodating these mechanisms. Until the safety

standards undergo a refinement process, the performance capabilities of the helmets will not improve.

Looking back at the means of reducing the risk of head and brain injuries in sport, as outlined by McIntosh and McCrory (2005), it is expected that this study will provide information regarding the last of these methods: developing personal protective equipment which lessens the amount of energy transferred to the head and brain. Specifically, a deeper understanding of the effects of impact mass, inbound velocity and system compliance on the dynamic response of head form will lead to a better description of test protocols which could ultimately contribute to the development of more protective sport helmets.

CHAPTER 6. CONCLUSION & FUTURE STUDIES

The goal of this paper was to determine the influence of impact mass on the dynamic response of a Hybrid III head form subjected to impacts over a range of system compliances and inbound velocities. The results indicate that an increase in impact mass leads to an increase in peak translational and peak rotational accelerations for all conditions. In addition, increases in inbound velocity also resulted in increases in the acceleration values. Conversely, increases in system compliance led to a non-linear response for translational acceleration: at the lowest compliance, translational acceleration was at a minimum, but at the next level of increased compliance, translational acceleration was at a maximum, and the middle value for acceleration occurred with the most compliant system. The response was however, linear for rotational acceleration: as compliance increased, so too did rotational acceleration. It is hoped that this new information will be incorporated into helmet certification standards in an attempt to further mitigate the risk of suffering mild traumatic brain injury.

6.1 CONCLUSION

Overall, the null hypotheses in this study were rejected. Impact mass had a significant effect on the dynamic response of a Hybrid III head form over a range of system compliances, impacted across three levels of velocity.

6.2 FUTURE STUDIES

In this study, all tests were conducted using a Hybrid III head form. A future study should apply a similar test methodology using the head forms currently prescribed in the helmet standards. This would give a better characterization of the systems currently used

in the certification of sports helmets and would assess whether or not the results of the current study can be generalized to other head forms. This information could lead to major changes in the protocols currently used in the evaluation of helmet performance, and ultimately to the reduction of head and brain injury in sports.

Further research could also be carried out which examines situations where both the impacted and impacting objects are in motion. This would introduce additional inbound energies and momentums, thereby changing the dynamics of the impact.

The results of this study could also be inputted into a finite elements model of the head and brain to gain a better understanding of the effects of these independent variables on the brain.

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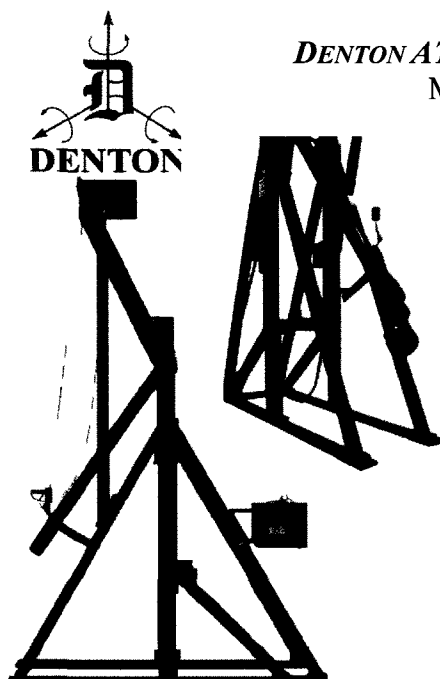
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APPENDIX A: NECK CALIBRATION PROTOCOL



DENTON ATD, INC. NECK PENDULUM TEST STAND

Model Number: TF-200-0000

INTRODUCTION

The TF-200 Neck Pendulum Test Stand is a complete system used for the calibration and testing of the neck component for the Hybrid III family, Hybrid II 50th, SA-106C, SA-103C, BioSID, EuroSID-1, ES-2, SID-HIII and SID-IIs. The neck can be tested in both the flexion and extension modes. The neck performance specifications for these tests are velocity at impact, pendulum acceleration, total rotation of the head/neck system, moment about the occipital condyle, and force.

The neck pendulum is in compliance with the specifications as written in the United States Code of Federal Regulation, Title 49, Part 572 relating to weight, center-of-gravity, moment of inertia, and mounting location of the pendulum accelerometer.

DESCRIPTION

The neck pendulum test is accomplished by releasing the pendulum and allowing it to fall freely to achieve a given impact velocity. The aluminum honeycomb stops the pendulum with a specified acceleration versus time pulse.

Velocity measurements are made by means of an infrared velocity measurement system. This system is made up of an infrared emitter, detector, and a precision slotted vane. The slotted vane is attached to the pendulum, and when dropped, the vane passes between the infrared emitter and detector generating a series of on/off pulses that can be converted into velocity by the data acquisition system.

Honeycomb material is positioned on four (4) aluminum dowel pins mounted on the face of the back stop which locates this material in the proper position between the pendulum striker plate and the back stop.

An upper neck load cell and one uniaxial accelerometer required for the test are not included.

Dimensions	Fixtures	Work Area
Length	11.0 ft	13.0 ft
Width	2.5 ft	5.0 ft
Height	12.9 ft	—

Standard Equipment

- * 'A' frame welded steel structure
- * Pendulum assembly; Part 572 compliant
- * Structural back stop
- * Infrared velocity measurement system
- * Mechanical Angle Indicator
- * Torque wrench 0 - 30 in-lb.
- * Calibration unit for potentiometers
- * Condyle Pin Removal Fixture
- * Scissors

Optional Equipment

- * Neck Mounting Adaptation packages are available for the following dummies:
 - * Hybrid III 50th/Hybrid III 95th
 - * Hybrid III 5th
 - * Hybrid III 10 Y.O.
 - * Hybrid III 6 Y.O.
 - * Hybrid III 3 Y.O.
 - * CRABI 12-Month-Old
 - * Hybrid II 50th
 - * SA-106C
 - * SA-103C
 - * SID-HIII
 - * SID-IIs
 - * EuroSID-1
 - * ES-2
 - * Hybrid III 5th/Hybrid III 6 Y.O. Adapter Plate for condyle pin removal fixture
 - * Aluminum honeycomb 4 ft x 8 ft x 6 inches
 - * Aluminum honeycomb 4 ft x 8 ft x 3 inches
 - * Electric hoist
 - * Electro-mechanical quick release
 - * Electronic inclinometer with display
 - * Pendulum rope hoist with snap release mechanism

Denton ATD, Inc. 10317 U.S. Highway 250 North, Milan, Ohio 44846-9570
 Tel (419) 625-5200 * Fax (419) 625-5335 * email: info@dentonatd.com * www.dentonatd.com

APPENDIX B: 50th PERCENTILE HYBRID III (78051-90) NECK CALIBRATION RESULTS

Test Name: Neck Flexion

Dummy type: Hybrid III 50th

Test ID: 144804

Test Number: 4682-1

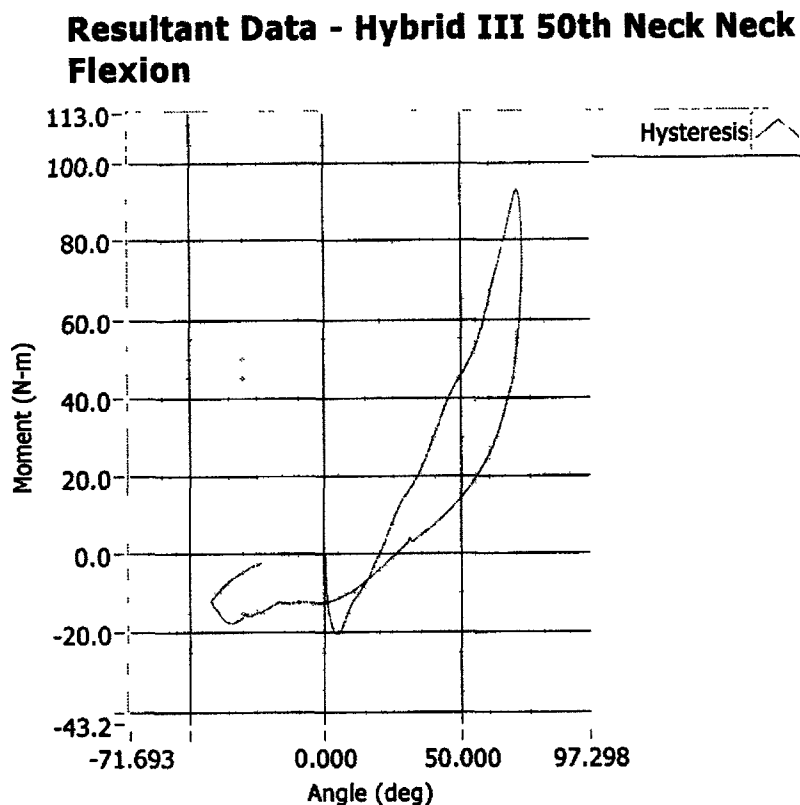
Test Date: 5/15/2007

Test time: 10:51 AM

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.20 deg C	P
Humidity		10 – 70	33 %RH	P
Velocity		6.89 – 7.13	7.05 m/s	P
Pendulum Deceleration at	10 ms	22.5 – 27.5	23.90 g	P
Pendulum Deceleration at	20 ms	17.6 – 22.6	20.19 g	P
Pendulum Deceleration at	30 ms	12.5 – 18.5	14.08 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 29.0	13.90 g	P
Decel Time to	5 g	34.0 – 42.0	40.20 ms	P
D Plane Rotation		-78.0 – -64.0	-72.42 degrees	P
Time at Max Rotation		57.0 – 64.0	60.40 ms	P
Rotation Decay to Zero		113.0 – 128.0	119.50 ms	P
Moment About Occipital Condyle		88.1 – 108.4	93.03 Nm	P
Time at Max Moment		47.0 – 58.0	53.90 ms	P
Moment Decay to Zero		97.0 – 107.0	102.70 ms	P

Test Supervisor: JR

Company: First Technology Safety Systems



Test Name: Neck Flexion

Dummy type: Hybrid III 50th

Test ID: 144812

Test Number: 4682-2

Test Date: 5/15/2007

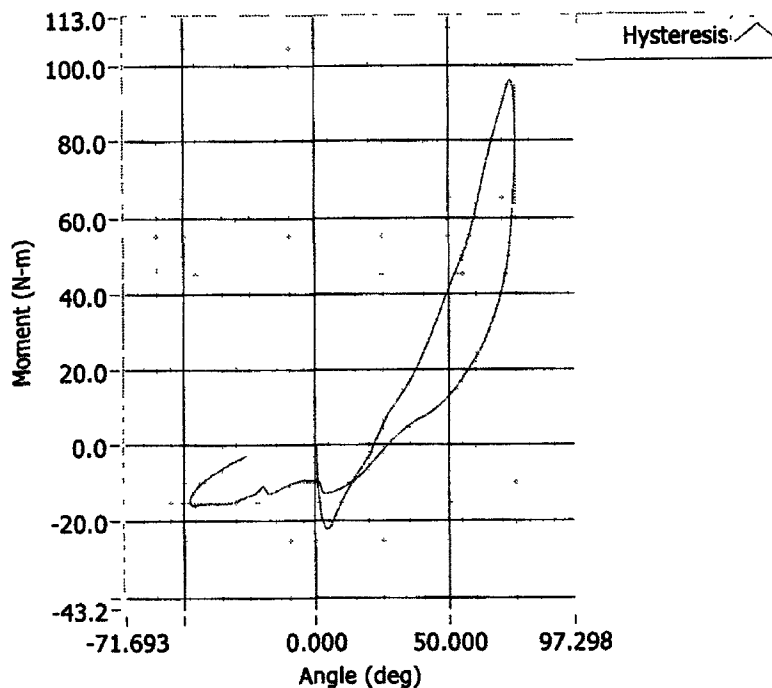
Test time: 11:32 AM

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.30 deg C	P
Humidity		10 – 70	33 %RH	P
Velocity		6.89 – 7.13	6.96 m/s	P
Pendulum Deceleration at	10 ms	22.5 – 27.5	23.16 g	P
Pendulum Deceleration at	20 ms	17.6 – 22.6	19.98 g	P
Pendulum Deceleration at	30 ms	12.5 – 18.5	13.86 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 29.0	13.68 g	P
Decel Time to	5 g	34.0 – 42.0	40.00 ms	P
D Plane Rotation		-78.0 – -64.0	-75.24 degrees	P
Time at Max Rotation		57.0 – 64.0	60.50 ms	P
Rotation Decay to Zero		113.0 – 128.0	119.30 ms	P
Moment About Occipital Condyle		88.1 – 108.4	96.18 Nm	P
Time at Max Moment		47.0 – 58.0	54.00 ms	P
Moment Decay to Zero		97.0 – 107.0	102.50 ms	P

Test Supervisor: JR

Company: First Technology Safety Systems

Resultant Data - Hybrid III 50th Neck Neck Flexion



Test Name: Neck Extension

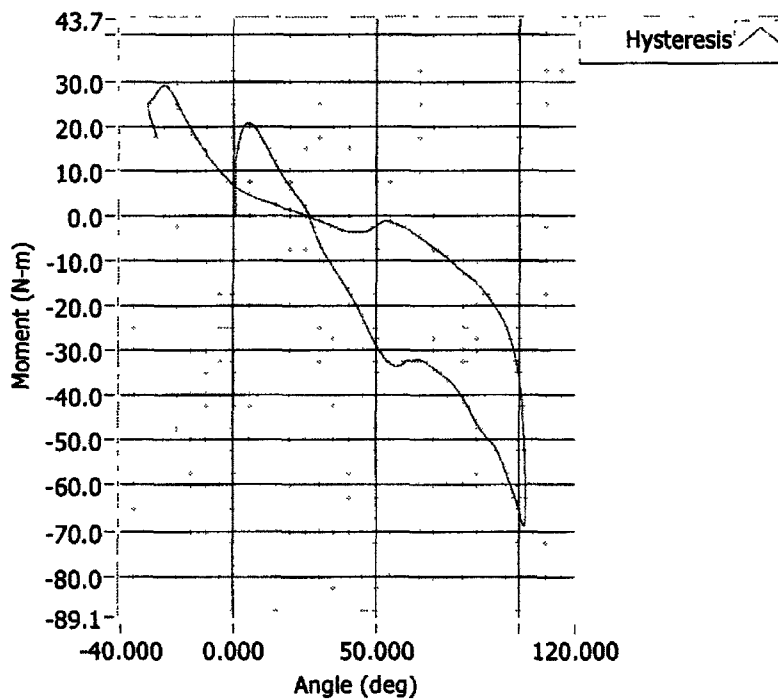
Dummy type: Hybrid III 50th
 Test ID: 144813
 Test Number: 4682-1

Test Date: 5/15/2007
 Test time: 12:12 PM

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.30 deg C	P
Humidity		10 – 70	33 %RH	P
Velocity		5.94 – 6.19	6.18 m/s	P
Pendulum Deceleration at	10 ms	17.2 – 21.2	20.26 g	P
Pendulum Deceleration at	20 ms	14.0 – 19.0	17.82 g	P
Pendulum Deceleration at	30 ms	11.0 – 16.0	13.40 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 22.0	13.26 g	P
Decel Time to	5 g	38.0 – 46.0	38.90 ms	P
D Plane Rotation		81.0 – 106.0	102.53 degrees	P
Time at Max Rotation		72.0 – 82.0	76.90 ms	P
Rotation Decay to Zero		147.0 – 174.0	164.90 ms	P
Moment About Occipital Condyle		-80.0 – -52.9	-69.00 Nm	P
Time at Max Moment		65.0 – 79.0	72.80 ms	P
Moment Decay to Zero		120.0 – 148.0	146.60 ms	P

Test Supervisor: JR
 Company: First Technology Safety Systems

Resultant Data - Hybrid III 50th Neck Neck Extension



Test Name: Neck Extension

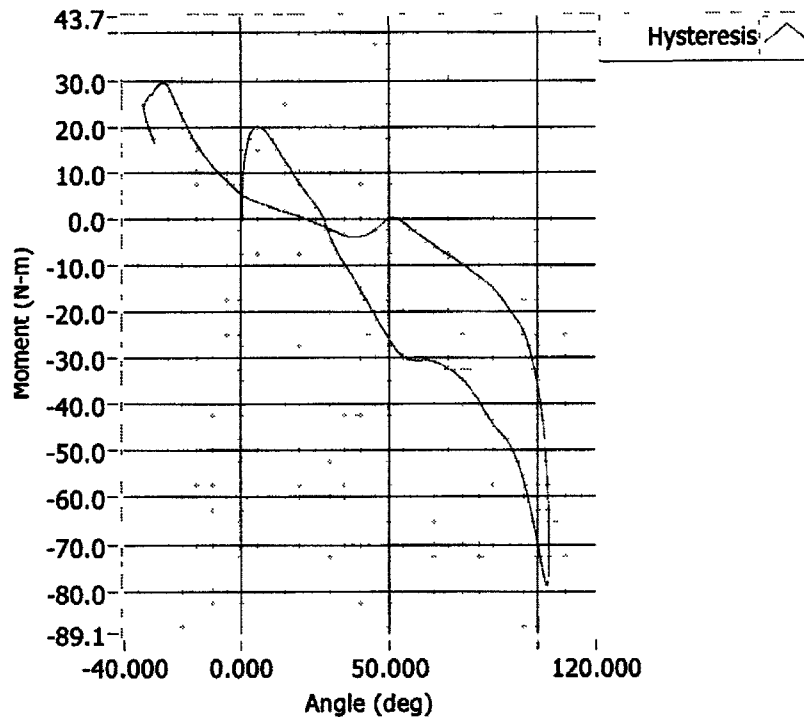
Dummy type: Hybrid III 50th
 Test ID: 144831
 Test Number: 4682-2

Test Date: 5/15/2007
 Test time: 12:56 PM

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.20 deg C	P
Humidity		10 – 70	33 %RH	P
Velocity		5.94 – 6.19	6.19 m/s	P
Pendulum Deceleration at	10 ms	17.2 – 21.2	21.20 g	P
Pendulum Deceleration at	20 ms	14.0 – 19.0	18.47 g	P
Pendulum Deceleration at	30 ms	11.0 – 16.0	12.58 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 22.0	14.03 g	P
Decel Time to	5 g	38.0 – 46.0	38.20 ms	P
D Plane Rotation		81.0 – 106.0	104.07 degrees	P
Time at Max Rotation		72.0 – 82.0	76.00 ms	P
Rotation Decay to Zero		147.0 – 174.0	161.80 ms	P
Moment About Occipital Condyle		-80.0 – -52.9	-78.96 Nm	P
Time at Max Moment		65.0 – 79.0	72.10 ms	P
Moment Decay to Zero		120.0 – 148.0	124.50 ms	P

Test Supervisor: JR
 Company: First Technology Safety Systems

Resultant Data - Hybrid III 50th Neck Neck Extension



Test Setup Details:

Channel	Name	Sensor SN	Axis	Cal Due Date	Gain	Filter	Class
1	Velocity	NeckVel2	None	06/13/07	1	3000	None
2	Moment, My	226	My	11/16/07	500	3000	600
3	Force, Fx	226	Fx	11/16/07	500	3000	1000
6	Pendulum Pot	3PP1	None	10/18/07	2	3000	60
7	Head Pot	3HP1	None	10/18/07	2	3000	60
5	Pendulum Accel.	EP11	None	08/17/07	100	3000	60

APPENDIX C: Soft Hybrid III (1820) NECK CALIBRATION RESULTS

Test Name: Neck Flexion

Dummy type: Hybrid III 50th

Test ID: AA1820-2

Test Number: 2

Test Date: 8/28/2007

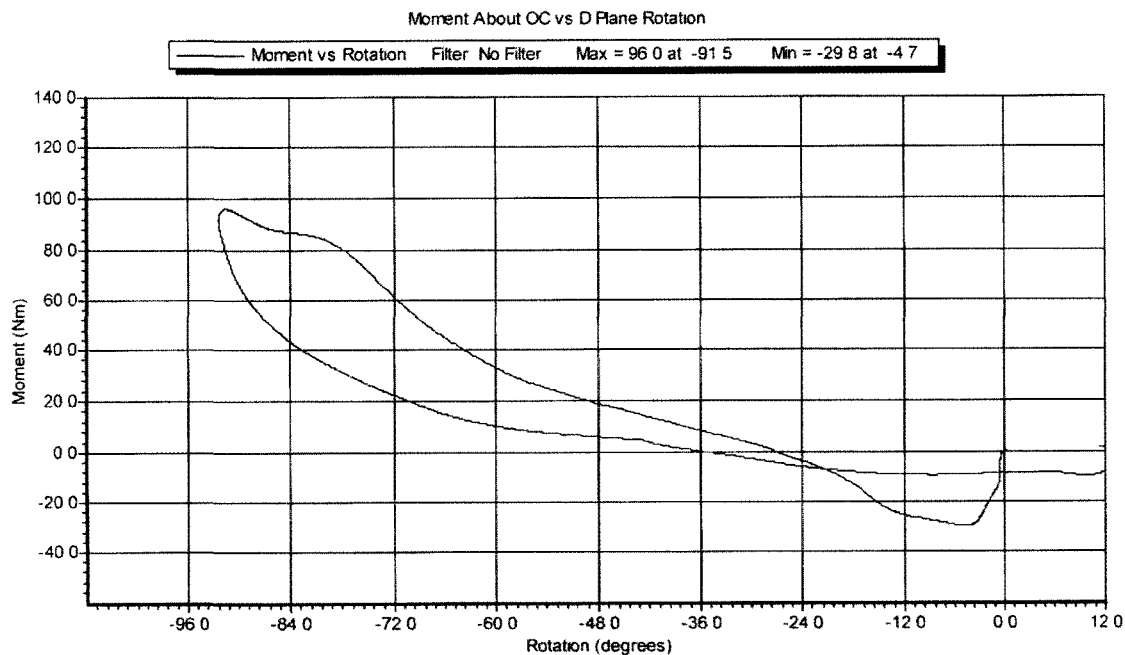
Test time: 10:22:55 AM

Comments: Soft rubber.

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.5 deg C	P
Humidity		10 – 70	54 %RH	P
Velocity		6.89 – 7.13	7.00 m/s	P
Pendulum Deceleration at	10 ms	22.5 – 27.5	25.7 g	P
Pendulum Deceleration at	20 ms	17.6 – 22.6	21.9 g	P
Pendulum Deceleration at	30 ms	12.5 – 18.5	17.0 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 29.0	17.0 g	P
Decel Time to	5 g	34.0 – 42.0	36.9 ms	P
D Plane Rotation		-78.0 – -64.0	-92.4 degrees	F
Time at Max Rotation		57.0 – 64.0	76.5 ms	F
Rotation Decay to Zero		113.0 – 128.0	129.9 ms	F
Moment About Occipital Condyle		88.1 – 108.4	96.0 Nm	P
Time at Max Moment		47.0 – 58.0	61.3 ms	F
Moment Decay to Zero		97.0 – 107.0	108.7 ms	F

Technician: GS

Company: Denton ATD, Inc.



Test Name: Neck Flexion

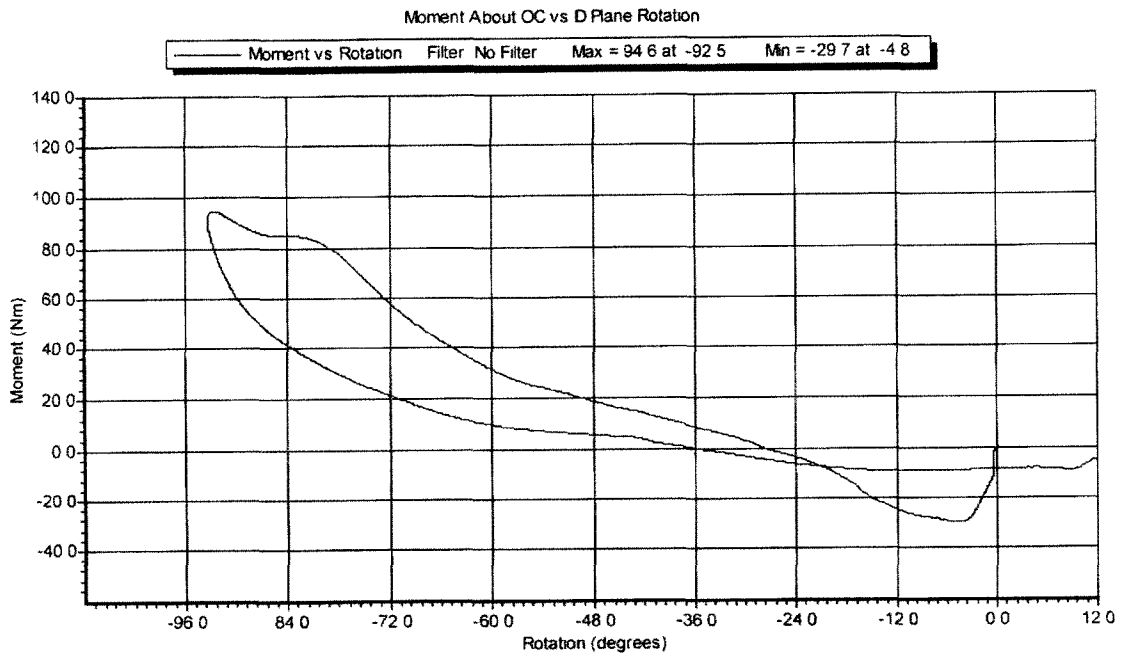
Dummy type: Hybrid III 50th
 Test ID: AA1820-3
 Test Number: 3

Test Date: 8/28/2007
 Test time: 11:32:59 AM

Comments: Soft rubber.

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.3 deg C	P
Humidity		10 – 70	56 %RH	P
Velocity		6.89 – 7.13	7.00 m/s	P
Pendulum Deceleration at	10 ms	22.5 – 27.5	24.6 g	P
Pendulum Deceleration at	20 ms	17.6 – 22.6	21.0 g	P
Pendulum Deceleration at	30 ms	12.5 – 18.5	17.6 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 29.0	17.6 g	P
Decel Time to	5 g	34.0 – 42.0	37.4 ms	P
D Plane Rotation		-78.0 – -64.0	-93.3 degrees	F
Time at Max Rotation		57.0 – 64.0	66.3 ms	F
Rotation Decay to Zero		113.0 – 128.0	131.2 ms	F
Moment About Occipital Condyle		88.1 – 108.4	94.6 Nm	P
Time at Max Moment		47.0 – 58.0	62.1 ms	F
Moment Decay to Zero		97.0 – 107.0	109.7 ms	F

Technician: GS
 Company: Denton ATD, Inc.



Test Name: Neck Extension

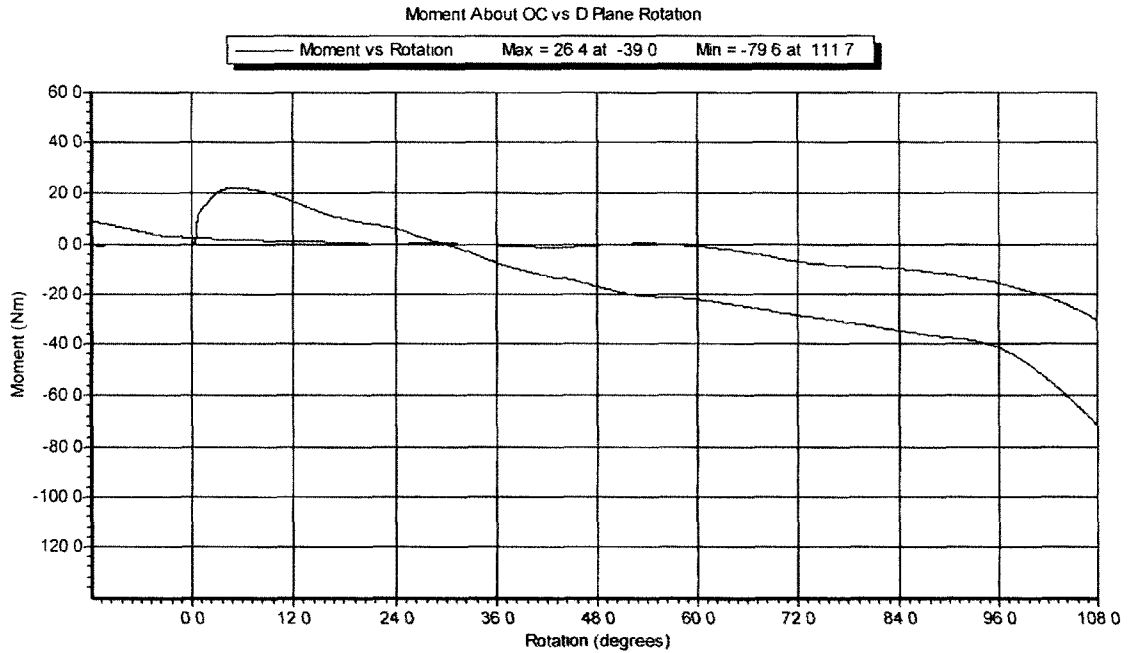
Dummy type: Hybrid III 50th
 Test ID: AC1820-1
 Test Number: 1

Test Date: 8/28/2007
 Test time: 1:26:33 PM

Comments: Soft rubber. Head touched back of bracket during test.

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.5 deg C	P
Humidity		10 – 70	57 %RH	P
Velocity		5.94 – 6.19	6.02 m/s	P
Pendulum Deceleration at	10 ms	17.2 – 21.2	19.9 g	P
Pendulum Deceleration at	20 ms	14.0 – 19.0	17.6 g	P
Pendulum Deceleration at	30 ms	11.0 – 16.0	15.3 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 22.0	15.3 g	P
Decel Time to	5 g	38.0 – 46.0	38.5 ms	P
D Plane Rotation		81.0 – 106.0	113.3 degrees	F
Time at Max Rotation		72.0 – 82.0	79.6 ms	P
Rotation Decay to Zero		147.0 – 174.0	170.3 ms	P
Moment About Occipital Condyle		-80.0 – -52.9	-79.6 Nm	P
Time at Max Moment		65.0 – 79.0	73.9 ms	P
Moment Decay to Zero		120.0 – 148.0	132.9 ms	P

Technician: GS
 Company: Denton ATD, Inc.



Test Name: Neck Extension

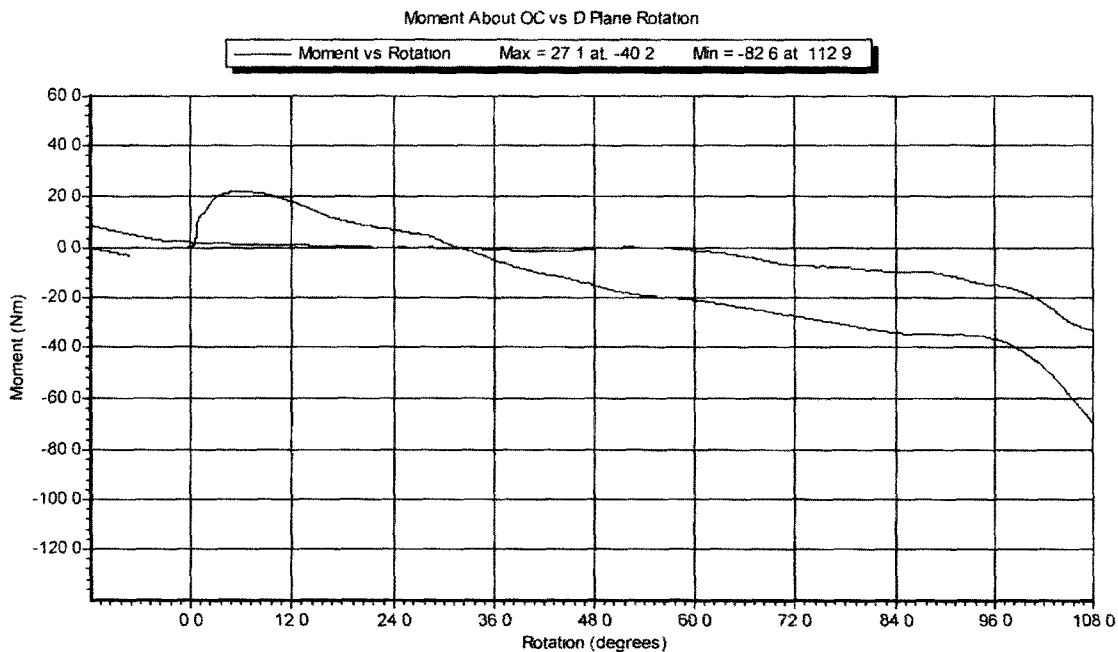
Dummy type: Hybrid III 50th
 Test ID: AC1820-2
 Test Number: 2

Test Date: 8/28/2007
 Test time: 2:42:48 PM

Comments: Soft rubber. Head touched back of bracket during test.

Test Parameters	Test Specifications	Test results (avg)	
Temperature	20.6 – 22.2	21.6 deg C	P
Humidity	10 – 70	58 %RH	P
Velocity	5.94 – 6.19	5.99 m/s	P
Pendulum Deceleration at 10 ms	17.2 – 21.2	19.7 g	P
Pendulum Deceleration at 20 ms	14.0 – 19.0	18.4 g	P
Pendulum Deceleration at 30 ms	11.0 – 16.0	15.2 g	P
Max Pendulum Deceleration after 30 ms	0.0 – 22.0	15.2 g	P
Decel Time to 5 g	38.0 – 46.0	37.8 ms	F
D Plane Rotation	81.0 – 106.0	114.3 degrees	F
Time at Max Rotation	72.0 – 82.0	76.8 ms	P
Rotation Decay to Zero	147.0 – 174.0	166.1 ms	P
Moment About Occipital Condyle	-80.0 – -52.9	-82.6 Nm	F
Time at Max Moment	65.0 – 79.0	73.1 ms	P
Moment Decay to Zero	120.0 – 148.0	130.9 ms	P

Technician: GS
 Company: Denton ATD, Inc.



Test Name: Neck Extension

Dummy type: Hybrid III 50th

Test ID: AC1820-4

Test Number: 4

Test Date: 8/28/2007

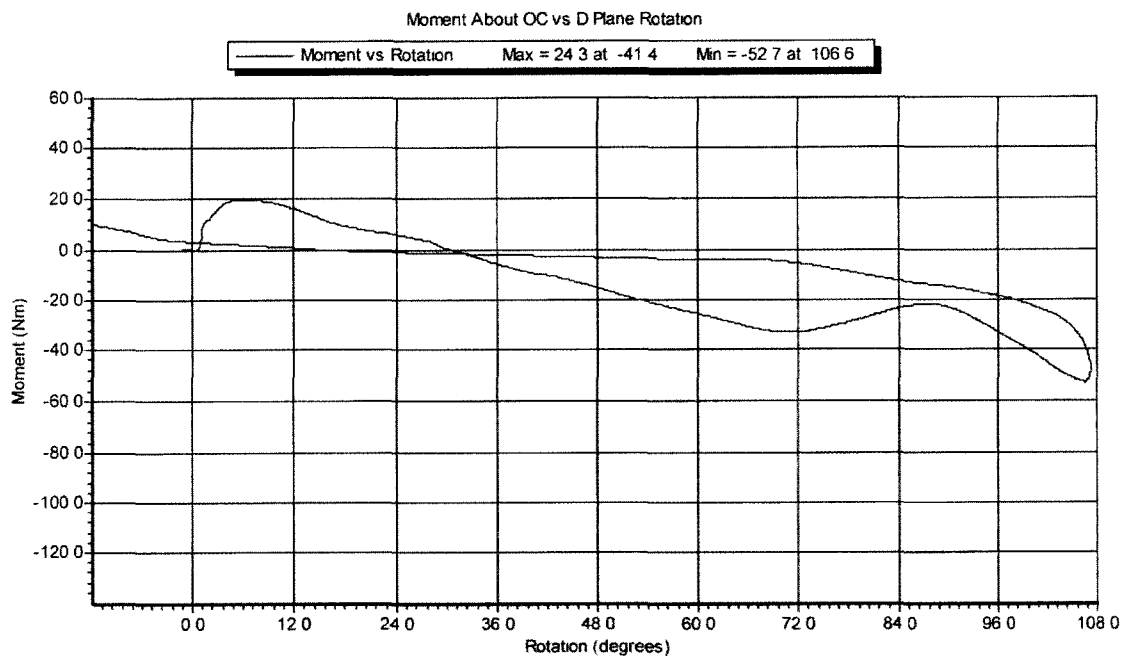
Test time: 4:02:56 PM

Comments: Soft rubber. Modified velocity.

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.4 deg C	P
Humidity		10 – 70	52 %RH	P
Velocity		5.94 – 6.19	4.91 m/s	F
Pendulum Deceleration at	10 ms	17.2 – 21.2	18.4 g	P
Pendulum Deceleration at	20 ms	14.0 – 19.0	16.6 g	P
Pendulum Deceleration at	30 ms	11.0 – 16.0	12.7 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 22.0	12.7 g	P
Decel Time to	5 g	38.0 – 46.0	33.6 ms	F
D Plane Rotation		81.0 – 106.0	107.2 degrees	F
Time at Max Rotation		72.0 – 82.0	87.4 ms	F
Rotation Decay to Zero		147.0 – 174.0	182.5 ms	F
Moment About Occipital Condyle		-80.0 – -52.9	-52.7 Nm	F
Time at Max Moment		65.0 – 79.0	82.7 ms	F
Moment Decay to Zero		120.0 – 148.0	172.2 ms	F

Technician: GS

Company: Denton ATD, Inc.



Test Name: Neck Extension

Dummy type: Hybrid III 50th

Test ID: AC1820-5

Test Number: 5

Test Date: 8/28/2007

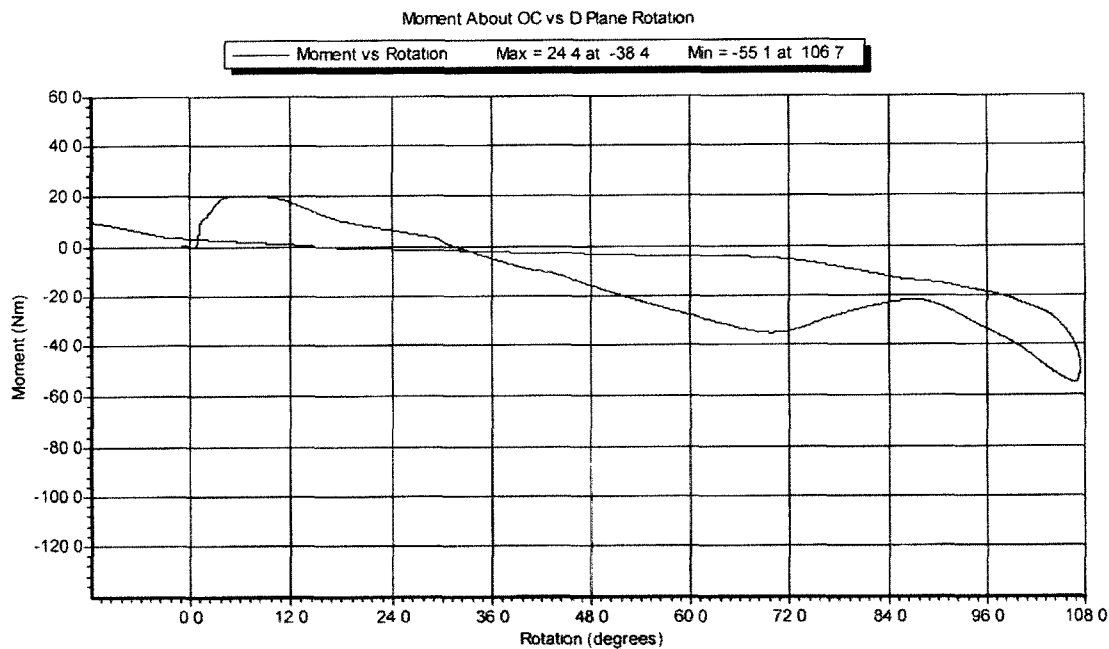
Test time: 4:32:26 PM

Comments: Soft rubber. Modified velocity.

Test Parameters		Test Specifications	Test results (avg)	
Temperature		20.6 – 22.2	21.3 deg C	P
Humidity		10 – 70	50 %RH	P
Velocity		5.94 – 6.19	4.91 m/s	F
Pendulum Deceleration at	10 ms	17.2 – 21.2	19.7 g	P
Pendulum Deceleration at	20 ms	14.0 – 19.0	17.0 g	P
Pendulum Deceleration at	30 ms	11.0 – 16.0	10.7 g	P
Max Pendulum Deceleration after	30 ms	0.0 – 22.0	10.7 g	P
Decel Time to	5 g	38.0 – 46.0	31.7 ms	F
D Plane Rotation		81.0 – 106.0	107.4 degrees	F
Time at Max Rotation		72.0 – 82.0	86.1 ms	F
Rotation Decay to Zero		147.0 – 174.0	181.0 ms	F
Moment About Occipital Condyle		-80.0 – -52.9	-55.1 Nm	P
Time at Max Moment		65.0 – 79.0	81.4 ms	F
Moment Decay to Zero		120.0 – 148.0	170.6 ms	F

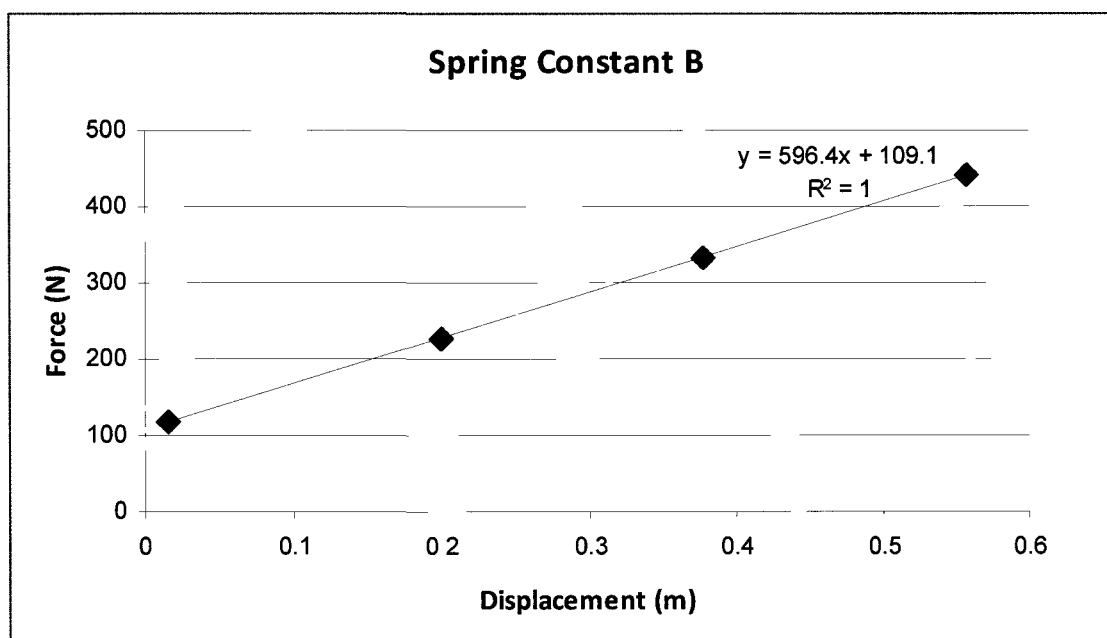
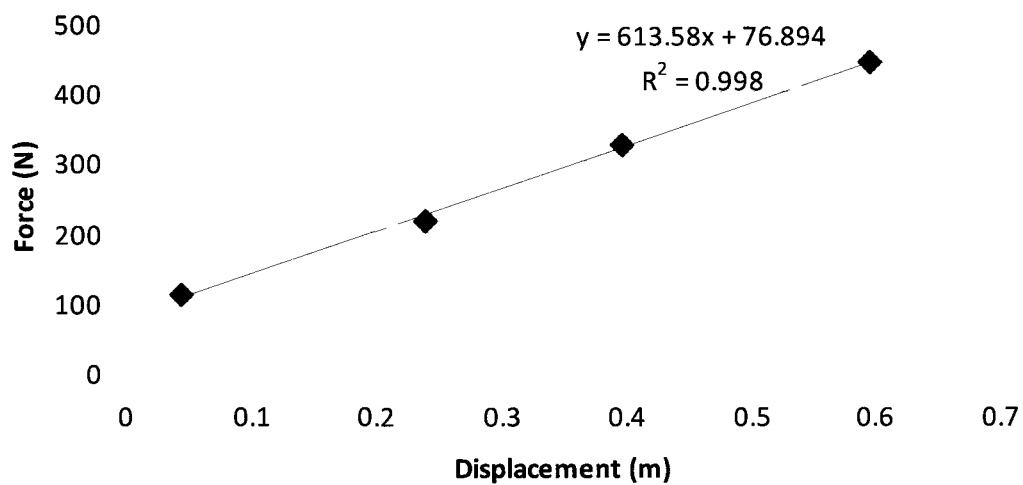
Technician: GS

Company: Denton ATD, Inc.



Reference equipment:

<u>Manufacturer</u>	<u>Model</u>	<u>Serial Number</u>	<u>Calibration Date</u>
Endevco	7231-750T	C17826	9/5/2006
Denton ATD	78051-342	7921-0538	1/2/2007
Denton ATD	78051-342	7921-0466	1/22/2007
Denton	1716A	1029	3/13/2007
Denton	1716A	1029	3/13/2007

APPENDIX D: SPRING CONSTANT DETERMINATION**Spring A Constant**

**APPENDIX E: VELOCITIES CORRESPONDING TO DRAW LENGTHS ON THE LIONEAR
IMPACTOR SPRING SYSTEM**

