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The Influence of Neck Compliance and Head Displacement on Impact Dynamics of a Hybrid III Head

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The Influence of Neck Compliance and Head Displacement on Impact Dynamics of a Hybrid III Head

Philippe Rousseau

A thesis submitted to
The Faculty of Graduate and Postdoctoral Studies of the University of Ottawa
In partial fulfillment of the requirements for the degree for the
Master of Science in Human Kinetics.

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October, 2008

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University of Ottawa

Your file Votre référence
ISBN: 978-0-494-46497-7
Our file Notre référence
ISBN: 978-0-494-46497-7

Abstract

The objective of this study was to determine the influence of impact deflection and neck compliance on peak linear, peak angular accelerations and the Gadd severity index. To replicate direct impacts to the head, a pneumatic linear impactor was used to impact a Hybrid III head and neck attached to a sliding table. The headform was translated 0cm, 3.88cm, 7.75 cm and 11.63cm laterally to identify the effects of impact deflection. The effects of neck compliance were tested using a Hybrid III 50^{th} percentile neck as well as two specially engineered Hybrid III necks. Impacts away from the centre of gravity of the headform recorded lower values for peak linear and angular accelerations and GSI. This means that impact deflection can effectively reduce brain injury risk. Neck compliance had a significant effect on linear acceleration and angular acceleration. An increase in neck compliance caused a decrease in linear acceleration; however, it also resulted in a decrease in angular acceleration. A comparison to published injury thresholds revealed that a decrease in neck compliance could significantly decrease the risk of brain injury. The data collected provided important insight into the importance of impact avoidance and deflection on injury prevention as well as the effect of the neck on the dynamic response of the head during impact.

Acknowledgment

J'aimerais dédier cette thèse à mes parents, Christian et Danièle. Vous m'avez encouragé tout au long de mes études et appuyé, peut importe mes décisions. Cette thèse est aussi dédiée à ma sœur, Frédérique, mon frère, Justin, et Stacey. Votre amour m'est très important et est nécessaire à ma carrière. Savoir que vous êtes tous derrière moi, prêt à m'appuyer me permettra de faire de grande chose tout au long de ma vie.

I would like to thank my research supervisor, Dr. Thomas Blaine Hoshizaki for his guidance and support. You have been, and continue to be, a source of inspiration for me. I have learned much from you and am looking forward to working many more years under your supervision. Your guidance allowed me to become a stronger researcher and develop my critical skills.

J'aimerais remercier les membres de mon comité de thèse, professeurs Yves Lajoie et Mario Lamontagne. Vos commentaires m'ont permis de renforcir ma thèse et d'en faire une œuvre dont je suis fier. J'espère avoir la chance de continuer à vous côtoyer et peut-être aurons-nous la chance un jour de travailler ensemble.

I would like to thank my colleagues, Andrew, Genille, Leslie, Natalie and Scott. Being a graduate student is a life changing experience and I believe that you have contributed into making be a better researcher as well as a person. I highly value your friendship and will always be available if you need me.

Finally, I would like to thank Xenith Inc. for supporting the Neurotrauma Impact Science Lab. Jeff, Mike, Bobby and Vin, I enjoyed your visits, which allowed me to work on product development. This allowed me to see another aspect of research and I believe it made me a more complete researcher. I am looking forward to many more years of collaboration.

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Where F_x is the contact force, F is the inbound force of the striker and Ψ is the impact angle.

Equation 2: $\boldsymbol{F = ma}$

Where F is impact force, m is effective mass and a is peak linear acceleration.

Equation 3: $\boldsymbol{M = I\alpha}$

Where M is moment, I is moment of inertia and α is peak rotational acceleration.

Equation 4: $\boldsymbol{SI = \int_0^t a^{2.5}\,dt}$ (Gadd, 1966)

Where a is peak acceleration and t is the duration of the impact.

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Where α is the angular acceleration for component i (x,y,z), A_{ij} is the linear acceleration for component i (x, y, z) along orthogonal arm j (1, 2, 3), and ρ_{x2} is the length of orthogonal arm j acting along the component i.

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Where α is the angular acceleration for component i (x,y,z), A_{ij} is the linear acceleration for component i (x, y, z) along orthogonal arm j (1, 2, 3), and ρ_{x2} is the length of orthogonal arm j acting along the component i.

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Where α is the angular acceleration for component i (x,y,z), A_{ij} is the linear acceleration for component i (x, y, z) along orthogonal arm j (1, 2, 3), and ρ_{x2} is the length of orthogonal arm j acting along the component i.

CHAPTER 1

INTRODUCTION

Among the head impacts incurred in sport, those that are unanticipated can have particularly devastating consequences. In fact, mild traumatic brain injuries (mTBI) caused by unexpected collisions are a common occurrence (Pellman, Powell et al., 2004; Viano & Pellman, 2005). While on the field, it is very difficult for athletes to know the exact position of all opponents, making them vulnerable to accidental or deliberate collisions. For instance, in American football, a quarterback attempting to complete a pass to the left of the field does not see a linebacker sprinting towards him from the right side creating a blindside tackle. In hockey, a player fighting for control of the puck behind the net may not see an opponent skating at full speed to administer a body-check. In baseball, a catcher waiting for a throw from the right field can not prepare for a collision with a runner trying to take home base. Similar situations also happen in other contact sports such as Australian rules football, basketball, lacrosse, rugby, and soccer (Bailes & Cantu, 2001; Toth, McNeil, & Feasby, 2005).

Unable to properly execute protective techniques, such athletes are at a higher risk of suffering a head injury. Conversely, athletes aware of impending collision prepare themselves to absorb, deflect or avoid forces generated by an impact (Reid, Epstein, Louis, & Reid, 1975).

Currently, there are few studies investigating the biomechanics of collisions in sport. While a number of studies have attempted to document collisions in America football, the current state of knowledge remains limited (Newman, Beusenberg, Shewchenko, Withnall, &

Fournier, 2005; Pellman, Viano, Tucker, & Casson, 2003; Pellman, Viano, Tucker, Casson, & Waeckerle, 2003; Viano et al., 2005; Viano & Pellman, 2005). In this thesis, two techniques commonly used by athletes to protect themselves against head impact injuries were examined: impact deflection and neck compliance. Characterizing the vulnerability of athletes during unanticipated impacts is expected to help us better understand the mechanism of injury and how to decrease risk of injury.

1.1 STATEMENT OF THE PROBLEM

Due to the nature of the sport, American football was identified as the sport with the highest number of head injuries (Powell & Barber-Foss, 1999; Toth et al., 2005). Studies reported that 3% to 19 % of all American football athletes suffered from an mTBI (Covassin, Swanik, & Sachs, 2003; Gerberich, Priest, Boen, Straub, & Maxwell, 1983; Guskiewicz, Weaver, Padua, & Garrett, 2000; Powell & Barber-Foss, 1999). The disparity between the studies can be attributed to factors such as population (high-school vs. college), the period of the study and the criteria used to construct questionnaires.

Recently, several studies have been omitting the word "concussion," which was too often perceived as trivial. Using mTBI symptoms as a substitute, Langburt and colleagues (Langburt, Cohen, Akhthar, O'Neill, & Lee, 2001) identified 47% of the participating American football players as having previously suffered from mTBI. Studies conducted in Canada showed similar trends by identifying 44.8% of Canadian Football League players and 70.4% of Canadian Interuniversity Sports football players as having reportedly suffered from symptoms related to mTBI (Delaney et al., 2000; Delaney et al., 2002). This implies that football athletes are at a higher brain injury risk than once thought, and actions need to be undertaken to increase their protection.

The resulting mechanical stress stemming from violent collisions puts players at a high injury risk when tackling and blocking (Boden, Tacchetti, Cantu, Knowles, & Mueller, 2006; Gerberich et al., 1983; Pellman, Powell et al., 2004; Powell & Barber-Foss, 1999; Torg, Guille, & Jaffe, 2002). Upon impact, forces are transferred between the athletes, subjecting their heads to both linear and rotational accelerations. Being composed of several types of tissue, the head does not react as a solid object to the change of velocity (Prange & Margulies, 2002). This disparity causes mechanical strain on structures traversing the subdural space and axons. Severe impacts generate sufficient strain and can cause an mTBI (Bayly et al., 2005; Kleiven, 2003; Zhang, Yang, & King, 2001).

Athletes have the capacity to diminish the injury risk by either deflecting a portion of the force transferred or by reducing the accelerations stemming from the impact. When two bodies collide, the force transmitted is in part related to the angle of transmitted force vector during contact (Fig. 1).

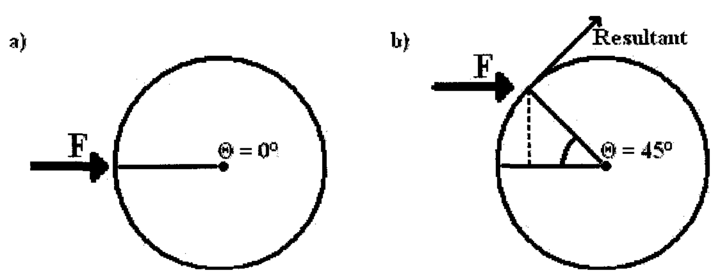

Figure 1: More force is transmitted to the head upon contact for impacts through the center of gravity (a) than eccentric impacts (b) due to the effect of impact angle.

The relationship between force and angle is assumed to be the following:

$$F_x = F\,cos\theta \tag{1}$$

Where F_x is the contact force, F is the inbound force of the striker and θ is the impact angle. By convention, 0° designates an impact through the center of gravity of the head. Athletes who are not expecting an oncoming collision are not able to either avoid or deflect the resulting blow. As a result, the impacts are closer to the center of gravity of their head and therefore more severe. Conversely, athletes anticipating the hit are able to either avoid the hit altogether or deflect the blow by adjusting their head position; thus, reducing the force transmitted to the head (Reid et al., 1975). Thus, it is proposed in this thesis that a displacement of the head will deflect the impact force sufficiently to reduce head injury risk

When forces are transferred from one body to another, the resulting accelerations can be reduced by increasing the effective mass of the receiving body (Aubry et al., 2002). This principle is described by Newton's second law of motion, which can be summarized as follow:

$$F = ma \tag{2}$$

$$M = I\,\alpha \tag{3}$$

Where F is impact force, m is effective mass and a is peak linear acceleration. The second equation is a modified version applicable to rotational acceleration, where M is moment, I is moment of inertia and α is peak rotational acceleration. Athletes unaware of an impending collision are not prepared to resist the shock; thus, the force transmitted through the impact is absorbed solely by the mass of the head. In contrast, athletes anticipating the hit have sufficient time to contract their neck muscles, allowing them to resist the impact with the mass of their head, neck and upper torso. Assuming that the impact forces sustained by both players are equal, the lower impact mass of the initial player will result in larger head peak acceleration post-impact (Viano & Pellman, 2005). Linear and rotational acceleration of the head are common predictors of mTBI; hence, when they increase, there is an associated increase in risk of injury. Thus, it is proposed in this thesis that a stiff or noncompliant neck will mitigate head injuries.

1.2 SIGNIFICANCE

Training methods (strength and training), administrative intervention (rules and regulations) and protective equipment design (helmet) all benefit from a better understanding of the mechanism of injury (Cantu & Mueller, 2003b; McIntosh & McCrory, 2005).

Previous studies suggested that training neck muscles to improve strength may help to prevent mTBI; nevertheless, the recommendations were based on observations and theoretical work (Bailes & Cantu, 2001; Barthe et al., 2001; Johnston, McCrory, Mohtadi, & Meuwisse, 2001; McIntosh & McCrory, 2005; Patel, Shivdasani, & Baker, 2005). This thesis will be the first to measure the effects of neck compliance on brain injury indicators. Moreover, this thesis will be the first to measure the effects of impact deflection on brain injury indicators, which is an important strategy used by all players (Reid et al., 1975).

Establishing and enforcing rules to protect players in vulnerable positions are common methods to prevent injury (McIntosh & McCrory, 2005). Furthermore, stringent rules protect reckless players who have a false impression of invulnerability caused by modern protective technologies (Cantu & Mueller, 2003a; Finch, McIntosh, & McCrory, 2001).

Finally, it is believed that these results may provide pertinent information to the design of helmets. If impact deflection does reduce peak linear and angular accelerations, it will become important to monitor size and geometry when designing helmets.

1.3 OBJECTIVE

The objective of this thesis was to determine the influence of impact deflection and neck compliance on head impact dynamics, which are peak linear acceleration, peak angular acceleration and Gadd Severity Index (GSI) during front impacts to a Hybrid III head.

1.4 RESEARCH HYPOTHESES

1.4.1 Impact deflection

1. An increase in impact deflection will result in a decrease in peak linear acceleration.
2. An increase in impact deflection will result in a decrease in peak angular acceleration.
3. An increase in impact deflection will result in a decrease in GSI.

1.4.2 Neck compliance

1. A decrease in neck compliance will result in a decrease in peak linear acceleration.
2. A decrease in neck compliance will result in a decrease in peak rotational acceleration.
3. A decrease in neck compliance will result in a decrease in GSI.

1.5 VARIABLES

1.5.1 Independent variables

1. Lateral translation: 0cm (centre of gravity), 3.88 cm, 7.75 cm and 11.63 cm.
2. Neck compliance: Soft, median and stiff.

1.5.2 Dependent variables

1. Peak linear acceleration (g)
2. Peak angular acceleration (rad/sec^2)
3. Gadd Severity Index ($SI = \int_0^t a^{2.5}\, dt$)

1.6 DELIMITATIONS

Helmets were not used in this study. Although football athletes wear a helmet while on the field, it introduces unnecessary variation to the measurement protocol. Variance is introduced through headgear fitting, shell mechanics and foam durability.

Three male Hybrid III necks, one standard and two modified, were used to represent different neck compliances. The modified necks were specifically engineered to represent -30% and +30% of the compliance of the standard 50th percentile male Hybrid III neck (Appendix B).

Four impact sites (centre of gravity, 3.875 cm, 7.75 cm and 11.625 cm) were used in this study. It is understood that athletes withstand impacts at a vast number of locations; however, the sites were determined based on the 50th percentile male Hybrid III head's width.

1.7 LIMITATIONS

A 50th percentile male Hybrid III head and a 50th percentile male Hybrid III neck were used as human surrogates for this thesis. Although the Hybrid III is the most widely used mannequin, it is not biofidelic, meaning that it does not imitate the head's exact dynamic properties (Deng, 1989; Seemann, Muzzy III, & Lustick, 1986). Furthermore, the necks were only calibrated for rotation around the y axis. Despite these limitations, the Hybrid III dummy remains the industry standard and has a multi-axis neck capable of withstanding the forces generated by high-velocity impacts.

The lowest velocity tested was 5 m/s, which represents the lowest velocity the linear impactor could be set, while the other two velocities, 7 m/s and 9 m/s, were chosen to reflect common impact velocities. Although some impacts have been recorded as high as 11.7 m/s (Pellman, Viano, Tucker, Casson et al., 2003; Pellman et al., 2006), the players were protected by their helmets, which was not the case for the Hybrid III headform. Nonetheless, it is believed that a similar study using helmets will yield the same relationship between impact deflection and neck compliance on peak linear and angular accelerations.

The headform was attached to a sliding table (12.78 kg), which allowed the dummy to slide backwards after the impact. The table offered very little resistance to motion and its effect on impact mechanics has not yet been studied.

CHAPTER 2

REVIEW OF LITERATURE

An estimated 1.6 to 3.8 million sport or recreational related traumatic brain injuries occur in the United States every year, most being mild events which are not treated in a hospital (Langlois, Rutland-Brown, & Wald, 2006). Furthermore, brain injury-related fatalities accounted for 69% of all on-field deaths among American football players between 1945 and 1999 (Cantu & Mueller, 2003a). The introduction and perfection of helmets dramatically decreased the number of traumatic brain injuries since 1980; however, the incidence of mild traumatic brain injuries (mTBI) has remained stable (Cantu & Mueller, 2003a; Covassin et al., 2003; Delaney et al., 2000; Delaney et al., 2002; Gerberich et al., 1983; Guskiewicz et al., 2000; Powell & Barber-Foss, 1999). To protect themselves against such injuries, football players have developed defensive strategies such as impact deflection and increasing the receiving mass involved in the impact. A player's natural reaction is to dodge the hit, causing it to be a glancing blow. This strategy reduces the efficiency of the impact, allowing for only a fraction of the transmitted force (Reid et al., 1975). When avoiding the hit is impossible, players will increase their effective mass by tensing the muscles in their neck. This will offer a greater resistance to the force and reduce the resulting accelerations (Reid et al., 1975). The injury mechanism, a description of traumatic brain injuries and common injury predictors will be discussed in this chapter. The effect of neck compliance on head and neck injury will also be discussed.

2.1 INJURY MECHANISM

High energy impacts in American football result in a variety of injuries. Tackling manoeuvres are the main cause of injury in football and have the potential to be catastrophic (Cantu & Mueller, 2003a; Gerberich et al., 1983; Guskiewicz et al., 2000). Upon contact, the head is subjected to linear and angular accelerations, both being the product of a dynamic application of forces to the head (Ommaya, Goldsmith, & Thibault, 2002). Linear acceleration, or translational acceleration, is the result of forces being applied directly inline with the centre of gravity of the head. Angular acceleration, or rotational acceleration, is a result of tangential forces applied elsewhere on the head, causing it to rotate. Linear and angular accelerations have both been positively correlated with head injuries, as well as commonly used brain injury predictors (Gennarelli, Ommaya, & Thibault, 1971, 1972; Gennarelli & Thibault, 1982; Gennarelli et al., 1982; Gurdjian & Lissner, 1944; Gurdjian, Lissner, Latimer, Haddad, & Webster, 1953; Gurdjian & Webster, 1943; Gurdjian, Webster, & Lissner, 1955; Holbourn, 1943; Lowenheilm, 1975; Ommaya & Hirsch, 1971; Ommaya, Hirsch, & Martinez, 1966; Ono, Kikuchi, Nakamura, Kobayashi, & Nakamura, 1980; Pellman, Viano, Tucker, Casson et al., 2003).

Following a traumatic event, the brain can suffer both primary and secondary injuries. Primary injuries are damage to the neural tissue as a consequence of the impact. These are generally classified as either focal or diffuse injuries (Ommaya & Gennarelli, 1974). Focal injuries are localized damage often caused by macroscopic lesions to the cerebrum, such as haemorrhages, contusions and haematomas (Nolan, 2005). Diffuse injuries are widespread and often follow a severe mTBI involving a loss of consciousness (Strich, 1961). These types of injuries are also associated with microscopic lesions characterized by the shearing of white matter tracts (Bailes & Cantu, 2001).

Secondary injuries are physiological responses to the cerebral lesions triggered by the impact. These types of injuries are divided into intracranial and systemic responses (Bailes & Cantu, 2001; Ommaya & Gennarelli, 1974). Common intracranial responses are cerebral hypertension, ischemia, herniation and vasospasm (Doberstein, Hovda, & Becker, 1993; Fisher, Kistler, & Davis, 1980). Systemic insults, such as hypoxia and hypotension, are a response to the impairment of the neural system. Hypoxia is thought to be caused by brainstem trauma and is often triggered by hypoventilation (Doberstein et al., 1993). Hypotension reduces cerebral perfusion, causing cerebral ischemia and infarction (Doberstein et al., 1993). Anaemia, hypoglycaemia, hyperthermia and electrolyte disturbance are less common but still potential systemic insults (Doberstein et al., 1993).

2.2 MAJOR TRAUMATIC BRAIN INJURIES

Major traumatic brain injuries, i.e. haematomas and contusions, have dramatically decreased since 1980, yet remain a major concern in American Football (Cantu & Mueller, 2003a).

Epidural haematomas are an accumulation of blood between the dura and the skull (Fig. 3), which are often accompanied by skull fractures (Bailes & Hudson, 2001). Although they are located outside of the brain, epidural haematomas disrupt brain activity once they reach a certain size (Jamieson & Yelland, 1968; Lobato et al., 1988).

Subdural haematomas are an accumulation of blood between the dura and the arachnoid (Fig. 2). They are caused by tears in subdural veins or lesions on the cortical surface and may cause brain herniation and cerebral ischemia (Logan, Bell, & Leonard, 2001). Subdural haematomas have been the leading cause of brain injury-related deaths (86%) in football between 1945 and 1999 (Cantu & Mueller, 2003a).

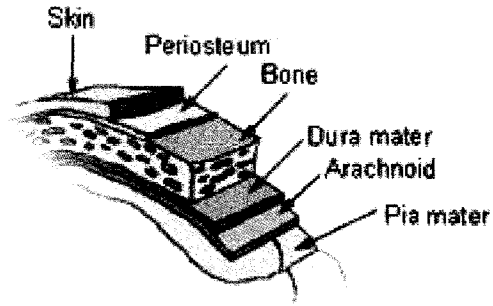

Figure 2: Cranial meninges ("The central nervous system," 2008).

Intracerebral haematomas are an accumulation of blood within the brain. They usually cause focal neurological disruption but can progress and affect a much wider sector. This, ultimately, leads to brain herniation, coma and potential death (Bailes & Hudson, 2001).

Cerebral contusions refer to haemorrhages, cerebral infarctions, necrosis and oedemas. They are similar to intracerebral haematomas; however, blood does not accumulate in a specific area (Bailes & Hudson, 2001). Contusions typically occur when the skull comes in contact with the brain (Nolan, 2005).

2.3 MINOR TRAUMATIC BRAIN INJURIES

Mild traumatic brain injuries, commonly referred to as concussion, are the most widespread head injuries and are widely documented in the literature (Bailes & Cantu, 2001; Covassin et al., 2003; Delaney et al., 2000; Delaney et al., 2002; Gerberich et al., 1983; Guskiewicz et al., 2000; Langburt et al., 2001; Pellman, Powell et al., 2004; Powell & Barber-

Foss, 1999). Mild traumatic brain injuries have been described as, "trauma-induced alterations in mental status that may or may not involve loss of consciousness" (Kelly et al., 1991). Historically these traumas have erroneously been dismissed as benign due to a lack of visible lesions to the brain (Levy, Ozgur, Berry, Aryan, & Apuzzo, 2004; Wojtys et al., 1999). New epidemiological researches have established the gravity of mTBI, which may occasionally lead to permanent disabilities and death (Bailes & Cantu, 2001; Centers for Disease Control and Prevention, 1997; Gerberich et al., 1983; Levy et al., 2004). They are diffuse injuries and consist of microscopic tears in the white matter without visible damage to the cortex (Strich, 1956). It is believed that shear stress and strain caused by the forces applied to the head stretches and tears nerve fibres (Strich, 1961).

The mechanism of mTBI is a complex sequence of events (Gaetz, 2004). The initial reaction, occurring within minutes following the impact, is a swelling of the axons. This swelling amplifies in the hours following the impact, resulting in disconnected axons (Grady et al., 1993; Pettus, Christman, Giebel, & Povlishock, 1994; Povlishock, Becker, Cheng, & Vaughan, 1983). Three to four days following the injury, certain axons have intact axolemmas while others show multiple segmented swellings and start to degenerate (Povlishock & Becker, 1985). Close to a week after the initial injury, the swollen axons show varied reactions: some remain unchanged, some have a disrupted axollema, and others start to regenerate (Povlishock & Becker, 1985). Two weeks post-injury, two responses are possible: a continued degeneration or a slow regeneration (Povlishock & Becker, 1985).

Degeneration of the axons is believed to be caused by a cascade of events starting with an axonal stretch, followed by a calcium (Ca^{2+}) influx (Gaetz, 2004). Axonal stresses cause an ionic imbalance characterised by an influx of sodium (Na^{+}), Ca^{2+} and chloride (Cl^{-}), and an efflux of potassium (K^+). This inhibits the generation and propagation of action potentials, and if severe enough, may lead to axonal death (Gennarelli, 1996).

Loss of consciousness (LOC), a symptom typically associated with concussions, is not an adequate indicator of trauma (Aubry et al., 2002). However, it is believed to be correlated with the depth of the lesion (Jenkins, Teasdale, Hadley, Macpherson, & Rowan, 1986) and should be considered as an expression of cerebral dysfunction, like memory and cognitive dysfunctions (Bailes & Cantu, 2001).

2.4 Post Concussion and Second Impact Syndromes

Athletes suffering from mTBI may be subjected to post concussion syndrome (Gerberich et al., 1983; Guskiewicz et al., 2000; Pellman, Powell et al., 2004). A wide variety of symptoms exists, which are listed in table 1. The severity and the number of symptoms varies from one athlete to another, usually disappearing two to four weeks after the event (Delaney et al., 2000; Delaney et al., 2002; Pellman, Viano et al., 2004).

Table 1: Symptoms for concussed players in NFL games (Pellman, Powell et al., 2004).

Category	Most common symptoms
General symptoms	Headaches; neck pain; nausea; syncope; vomiting
Cranial nerve symptoms	Dizziness; blurred vision; vertigo; photophobia
Memory problems	Retrograde amnesia; information processing problems
Cognitive problems	Immediate recall; time, place, persons disorientation
Somatic complaints	Fatigue; anxiety; personality changes, irritability

It is essential that athletes be well recovered before resuming training or competition (Canadian Academy of Sport Medicine Concussion, 2000). Players still experiencing post concussion symptoms may collapse and become comatose shortly after a mild impact (Bailes & Cantu, 2001). This is known as a second impact syndrome and is often lethal (Saunders & Harbaugh, 1984).

Following the first mTBI, the brain is in a vulnerable state. While the cells are not damaged mechanically, they require a large amount of energy to counteract the ionic imbalance (Doberstein et al., 1993). Subsequent impacts will further disrupt the weakened homeostasis and cause great damage to the brain. Rest is necessary to correct the cellular alterations and normal functions are resumed with time (Doberstein et al., 1993).

2.5 IMPACT MECHANICS

Research in head injury started at a time when measurement techniques were limited, which made it impossible to directly measure the effects of an impact on the brain. Consequently, other parameters had to be established to quantify the probability of brain trauma. Peak linear acceleration quickly became accepted as the best predicted measurement; however, some believed that peak angular acceleration was more appropriate (King, Yang, Zhang, & Hardy, 2003). Over time, the research community set aside the direct injury causes and primarily attributed acceleration as the main role in brain trauma. This led to studies that recreated purely linear or angular acceleration events, even though these instances are practically non-existent outside of a laboratory (King et al., 2003).

2.5.1 Peak linear acceleration

Sixty years ago, Gurdjian and his associates studied head injuries using anaesthetised dogs. They discovered that acceleration and deceleration present in typical head injuries were almost always associated with compression. This led to skull deformations as well as changes in intracranial pressure. Both were highly linked with brain trauma. In addition, time duration of the pressure and acceleration of the head were identified as important factors in the mechanics of concussion (Gurdjian & Lissner, 1944; Gurdjian et al., 1953; Gurdjian & Webster, 1943). The contribution of angular acceleration and linear acceleration remained undetermined until 1955, when an article by Gurdjian and his colleagues reported minimal contribution by angular acceleration in many concussive events; thus, making linear acceleration a more important contributor (Gurdjian et al., 1955).

Gurdjian's theory was supported by Ommaya and his associates (1971; 1966) who hypothesized that angular acceleration alone could not be the cause of mTBI. Their studies, conducted on monkeys, consisted of a comparison between direct impact and impulse loading (whiplash). They concluded that although angular acceleration caused by whiplash can cause a concussion, it requires higher levels of acceleration then that of direct impacts, making it a lesser contributor in the mechanism of injury.

Ono and colleagues (1980) led their own study with the hope of determining which form of acceleration was more detrimental to one's life. To solve this issue, they dissociated linear and angular acceleration by using four devices capable of producing pure linear or angular acceleration through direct impacts and impulse loading. The results showed a good correlation between concussions and peak linear acceleration but none with peak angular acceleration of the head. This observation led them to theorise that mTBI were mostly caused by linear acceleration from direct impacts.

There is currently no precise injury threshold; however, peak linear accelerations were reported to be 66, 82, and 106 G for a 25%, 50%, and 80% probability of sustaining an mTBI, respectively (Zhang, Yang, & King, 2004).

2.5.2 Peak angular acceleration

Holbourn (1943) was the first to propose that the level of injury was proportional to the amount of angular acceleration. He based his theory on the brain's physical properties. The brain has a high modulus of incompressibility and a small modulus of rigidity. This means that pressures arising from head injuries are too low to cause any harm through compression and rarefaction. He thus assumed that shear-stress, through elongation of the particles composing the brain, was the main source of injury. Holbourn observed that shear-stress was influenced by angular acceleration and he went on to identify it as the main injury predictor. This allowed him to conclude that angular acceleration, and not linear acceleration, was responsible for concussions, haemorrhages and contre-coup injury.

Using a mathematical model of the brain, Lowenheilm (1975) studied the role of angular acceleration in gliding contusions. He observed that the injuries were caused by strain on the surrounding tissues of the brain. This implied that angular acceleration was responsible for brain deformation and shear-strain in the subcortical matter.

Gennarelli and his colleagues (1971; 1972; 1982) conducted a series of studies, using monkeys and physical models, to determine the role of acceleration in impacts involving no contact. The results showed a predominant role of angular acceleration in the mechanism of concussion, diffuse axonal injury and subdural haematoma. This led to the conclusion that angular acceleration alone was capable of causing sufficient shear stress which led to cause brain damage (Gennarelli et al., 1971, 1972; Gennarelli & Thibault, 1982; Gennarelli et al., 1982).

Similar to linear acceleration, no precise injury threshold exist for angular acceleration. Peak angular accelerations were reported to be 4,600, 5,900, and 7,900 rad/s^2 for a 25%, 50%, and 80% probability of sustaining an mTBI, respectively (Zhang et al., 2004).

2.5.3 Gadd Severity Index

The Gadd severity index is used to measure head insult in enclosed environments such as sport helmets. Gadd (1966) proposed the following equation:

$$SI = \int_0^T a^{2.5}\, dt \tag{4}$$

Where a is the peak acceleration of the headform's centre of gravity expressed in g and t is the duration of the impact in seconds. The 2.5 as a weighing factor is based on the slope of the Wayne State Tolerance Curve (WSTC) between 4 and 50 ms. The WSTC (Fig. 3) is a plot of effective head acceleration vs. impact duration created by Lissner and his colleagues (1960; Patrick, Lissner, & Gurdjian, 1963). The curve was drawn using six data points taken from cadaveric tests. The curve was originally used to predict cranial fractures; yet, concussion onset was inferred, following the observation that concussions were associated with skull fractures in 80% of the patients admitted in hospitals. Further data points were added from tests conducted on animals using a pressure pulse to prolong the curve (Zhang et al., 2001).

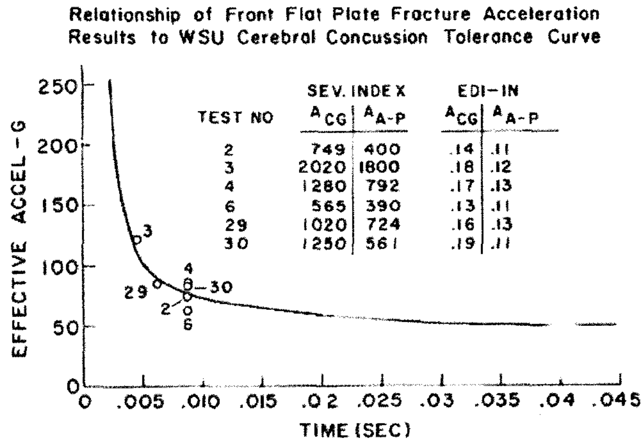

Figure 3: Wayne State Tolerance Curve (WSTC).

NFL studies conducted by Pellman and colleagues (2003) showed that GSI was strongly correlated to mTBI (R = 0.75). An mTBI tolerance level of 300 GSI was proposed for helmet impacts following laboratory reconstructions based on video analysis of concussive impacts in the NFL recorded between 1996 and 2001 (Pellman, Viano, Tucker, Casson et al., 2003).

2.6 THE NECK

2.6.1 Anatomy

The human neck is composed of seven vertebrae, called cervical, joined by ligaments and supported with muscle tissue (Fig. 4). Its peculiar morphology is ideal for the wide range of movement required during daily activities.

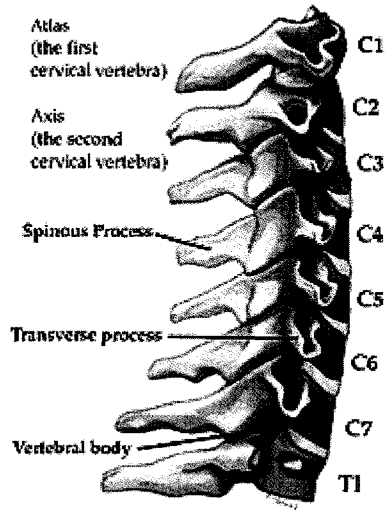

Figure 4: Human cervical spine.

The first vertebra (C1), also known as atlas, possesses a distinct shape allowing it to cradle the occipital. The main movement at the atlanto-occipital joint is flexion/extension, yet axial rotation and lateral flexion are possible when forced (Werne, 1959). The second vertebra, called the axis, has an odontoid process (dens) which projects upward. Its primary responsibility is to support the weight of the head while allowing the most axial rotation (Bogduk & Mercer, 2000). The remaining five vertebrae have similar shapes and are meant to protect the spinal cord and the meninges enclosed within (Moore & Dalley, 1999). They are all capable of flexion/extension, axial rotation and lateral flexion. The maximal range is largely limited by the vertebrae structures and the ligaments connecting them (Bogduk & Mercer, 2000).

2.6.2 Musculature

Motion at the neck is made possible by the many neck muscles acting upon it (Fig. 5). The sternocleidomastoid, longus capitis & colli, and the scalenus anterior, medius & posterior are responsible for neck flexion. The splenius capitis & cervicis, spinalis capitis & cervicis, semispinalis capitis & cervicis, longissimus capitis & cervicis and trapezius are responsible for neck extension.

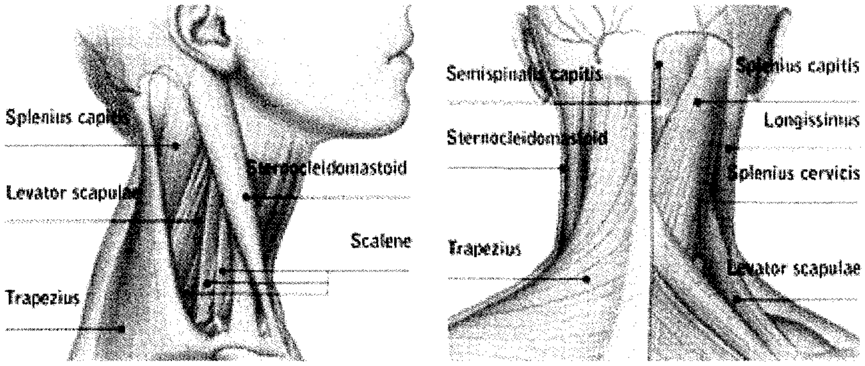

Figure 5: Superficial neck muscles.

Many studies have attempted to measure the strength of the neck muscles. Using force transducers, Choi (2003) measured peak moments of the neck at C4/5 level. Ten healthy males participated and performed near maximum, isometric, voluntary and ramp efforts in extension and flexion. The mean peak moments measured were 28.3 ± 3.3 Nm and 17.7 ± 3.1 Nm, for extension and flexion respectively. Seng and colleagues (2002) developed a device capable of measuring isometric neck strengths. Ten healthy male subjects performed maximal muscle contractions in four test directions. The mean peak moments measured were 45.29 ± 11.46 Nm and 23.34 ± 4.93 Nm, for extension and flexion respectively. Vassada and colleagues (2001) used a similar device to measure neck moments of 11 male subjects in different directions. The mean peak moments, recorded using load cells, were of 52 ± 11 Nm and 30 ± 5 Nm, for extension and flexion respectively. Strength data for the neck muscles show that the extensor muscles yield the greatest torque and that flexor muscles provide the least torque (Choi, 2003; Seng et al., 2002; Vasavada et al., 2001). Variations in the results may be due to the different techniques employed, the preparation of the subjects, and the position of the head and neck during the effort.

Few studies have examined the role of neck muscles activation in mTBI mechanism; nevertheless, its role in whiplash injury is fairly understood. Following the publication of articles exposing awareness of the impending accident as a possible factor of injury severity, it was demonstrated that pre-impact muscle contraction decreased head displacement by stiffening the head-neck complex, thus mitigating whiplash injuries (Blouin, Inglis, & Siegmund, 2006; Brolin, Halldin, & Leijonhufvud, 2005; Hendriks et al., 2005; Stemper, Yoganandan, Cusick, & Pintar, 2006; Stemper, Yoganandan, Rao, & Pintar, 2005; Sturzenegger, DiStefano, Radanov, & Schnidrig, 1994)

The influence of neck muscles on neck injury mechanism has been extensively studied in the automobile industry. It was determined that awareness of an impending impact had a role in the severity of the symptoms and total recovery time (Hendriks et al., 2005; Sturzenegger et al., 1994). Awareness is crucial because of the time required by the neck muscles to achieve full contraction. Figure 6 illustrates the inherent delay time of neck muscles for contraction, which lasts between 130 and 263 ms and can be decomposed in three segments. The first step (36-61 ms) is called reflex delay and represents the time lapse between the stimulus and the initiation of muscle activity. It is followed by an electromechanical delay (13-121 ms), which is the time taken to convert the electrical activity to a mechanical force. The last segment represents the time taken to achieve maximum contraction (± 81 ms) (Stemper et al., 2005).

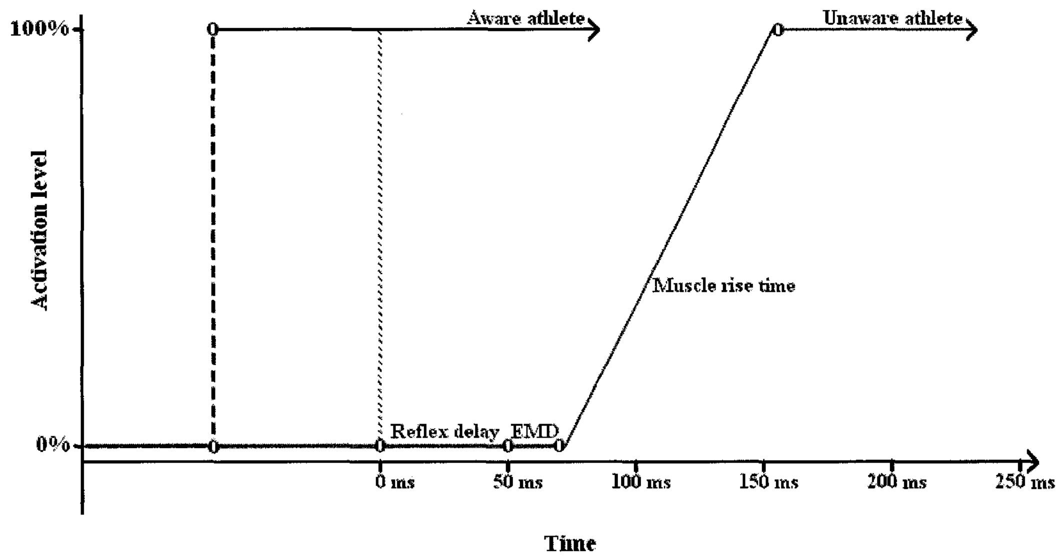

Figure 6: Timing of neck muscle contraction in aware and unaware athletes.

Based on the assumption that subjects aware of the impending impact will contract their muscles in anticipation of the collision, Stemper and colleagues (2006) hypothesized that "precontraction of neck muscles would reduce the level of soft tissue distortion during initial whiplash stages, which would correlate with a decreased likelihood of injury in aware occupants." Their results showed a 63% decrease in maximum head angle relative to T1 in the aware occupant. This indicates that, if given enough time to build up strength, the neck muscles are capable of mitigating the likelihood of an injury. Conversely, unaware subjects initiate neck muscle contraction following the impact. The late response is insufficient to achieve positive effects and might even aggravate the injury (Blouin et al., 2006; Stemper et al., 2005).

2.7 SUMMARY

In this review of literature, the complexity and risk of traumatic brain injuries was described. These injuries represent a severe risk to the career and wellbeing of all athletes and efforts need to be made to diminish their occurrence. The strategies mentioned above are well known and taught to young athletes; yet no attempt have been made to establish their efficiency. The goal of this thesis was to investigate impact deflection and neck compliance to determine if they truly have mitigating effects on the risk of mTBI.

CHAPTER 3

METHODOLOGY

The objective of this thesis was to determine the influences of impact deflection and neck compliance on peak linear and angular accelerations and the Gadd Severity Index (GSI). Since no similar studies were found, a unique methodology had to be developed using a Hybrid III headform and a pneumatic linear impactor. This device was chosen based on its ability to operate at higher velocities than conventional monorails as well as reflect the mechanics of impacts occurring during tackling in football. To mimic impact deflection, the headform was translated laterally by sliding the table on which it was installed. Neck compliance was altered using three Hybrid III necks. One was engineered to show a 30% increase in compliance and another to show a 30% decrease. Peak linear acceleration, peak rotational acceleration and GSI were measured using a 3-2-2-2 array of accelerometers in the Hybrid III head (Padgaonkar, Kreiger, & King, 1975).

3.1 Apparatus

3.1.1 Pneumatic linear impactor

The pneumatic linear impactor (Fig. 7) consisted of three major components: the support/piston frame, the impacting arm, and a Hybrid III headform mounted on a table. The frame supported the impacting arm, the compressed air canister, and the piston which was operated electronically by a control station. The impacting arm (mass 16.6 ± 0.1 kg) was propelled horizontally by compressed air. On the tip of the impacting arm was a cap consisting of a 0.677 ± 0.001 kg hemispherical nylon pad covering a 35.71 ± 0.01 mm thick vinyl nitrile 602 foam disc (Pellman et al., 2006).

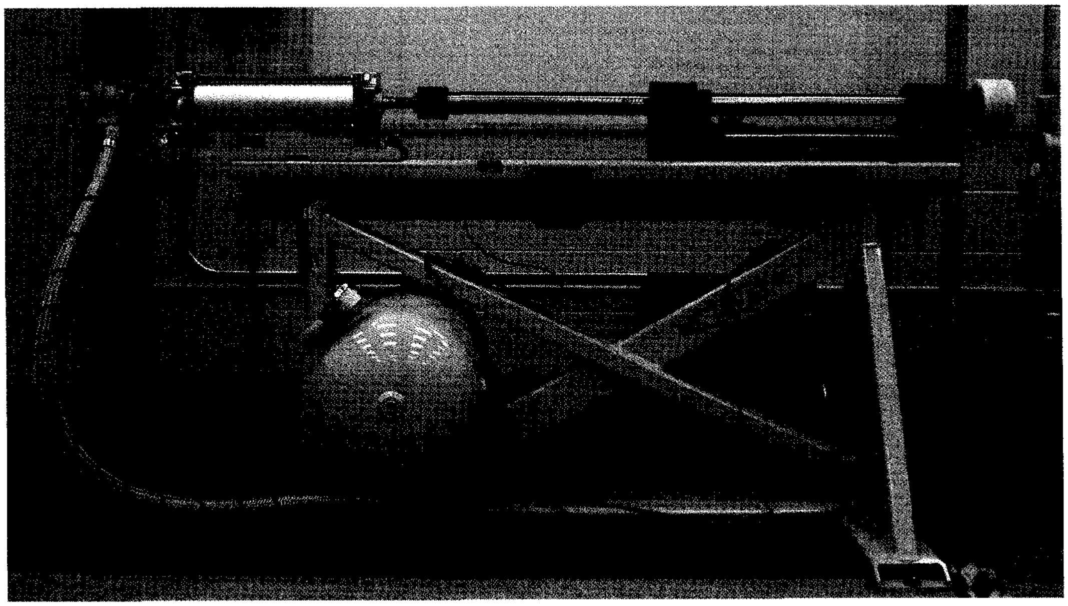

Figure 7: Linear impactor.

The mobile table was installed at the receiving end to support a Hybrid III head and neck (Fig. 8). The mass of the table was 12.782 ± 0.001 kg yet was installed on rails, allowing the dummy to slide backwards with little resistance. A spring loaded brake system provided a safe stop following a displacement of 0.54 ± 0.01 m.

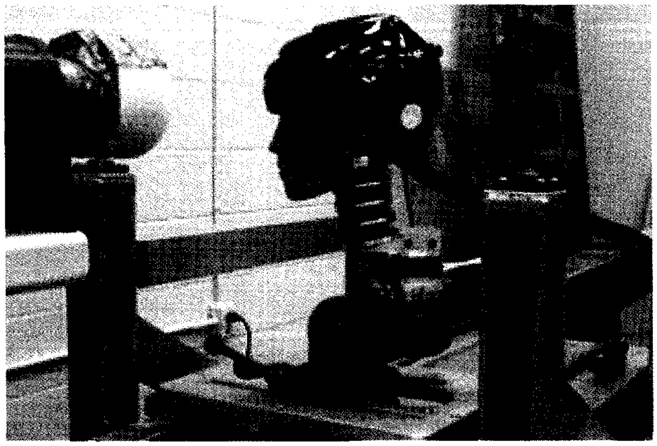

Figure 8: Mobile table supporting the Hybrid III headform.

The base supporting the Hybrid III head and attaching it to the moving table was built to allow for complete control over the location of impact. It could be adjusted in five degrees of freedom, including fore-aft (x), lateral (y), and up-down (z) translation, as well as fore-aft (y) and axial (z) rotation of the neck base. The adjustments were lockable and remained fixed throughout the testing.

3.1.2 Hybrid III head

A 50th percentile male Hybrid III head (FTSS, Plymouth MI, USA) (Fig. 19), with a mass of 4.54 ± 0.01kg, was used in this study. This type of headform is engineered to respond similarly to a human head under impact (Deng, 1989; Seemann et al., 1986).

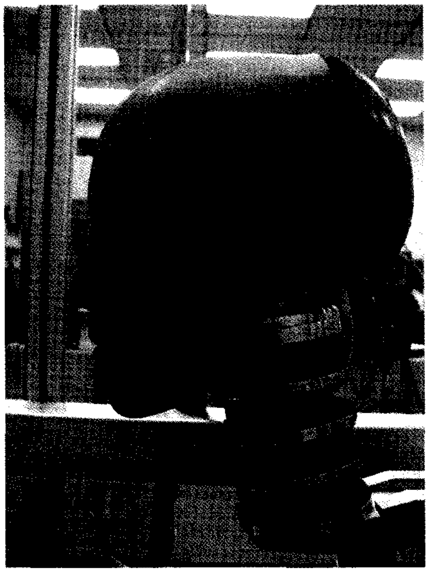

Figure 9: Hybrid III head form attached to a Hybrid III neck.

The sensors mounted inside the Hybrid III headform were 9 single-axis Endevco 7264C-2KTZ-2-300 accelerometers, measurement range 500 peak g (Fig. 10, left) and were calibrated by First Technology Safety System (Plymouth MI, USA). They were positioned in an orthogonal arrangement following the 3-2-2-2 array (Padgaonkar et al., 1975). Three sensors were mounted near the headform centre of gravity, two on the anterior surface of the skull, two on the lateral surface and two on the superior surface (Fig. 10, right). The processing of the nine signals allowed the determination of the complete three-dimensional motion of the head. The accelerations were collected at a frequency of 20 kHz.

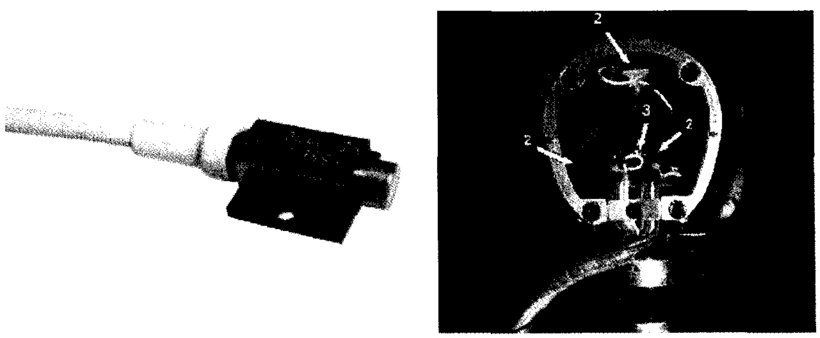

Figure 10: Endevco accelerometer (left); 3-2-2-2 cluster (right).

3.1.3 Hybrid III neck

The Hybrid III neck (Fig. 10), with a mass of 1.54 ± 0.05kg, was composed of four Butyl rubber discs interlocked between five aluminium plates to simulate human vertebrae. The discs were offset 0.5 cm towards the front of the neck and were slit to provide different response in flexion than in extension bending (Ashrafiuon et al, 1996).

To reproduce different neck compliances, three different Hybrid III necks were used. One neck remained unaltered and was arbitrarily chosen as the median stiffness neck, while the remaining two were specifically manufactured by Denton ATD to correspond to softer (-30% stiffness) and stiffer (+30% stiffness) necks. To attain proper compliance, modifications were made to the composition of the rubber discs.

All necks were tested following the specifications of the code of federal regulations: number 49, part 572 (FMVSS, Washington DC) using a neck pendulum (Appendix A). During extension testing for the softer neck, the back of the head impacted the lower neck bracket. To accommodate for this limitation, test velocities were lowered by approximately 1.1 m/s. Following two successful extension tests at this lower velocity with the soft neck, the stiff neck was also tested twice at the same velocity to have comparable data. All calibration results can be found in Appendix B.

3.2 PROCEDURE

The effect of impact deflection was evaluated by impacting a 50th percentile Hybrid III head with a linear impactor. The headform was hit nine times per impact location at the following velocities: 5 m/s and 7 m/s, which are comparable to impact velocities seen in football (Pellman, Viano, Tucker, Casson et al., 2003; Pellman et al., 2006).

The locations were chosen using the width of the headform as a reference (15.5 cm) and were the following: through the center of gravity; 3.875 cm lateral displacement; 7.75 cm lateral displacement; and 11.625 cm lateral displacement (Fig. 11). All impacts were located 30 ± 1 mm above the reference plane of the headform.

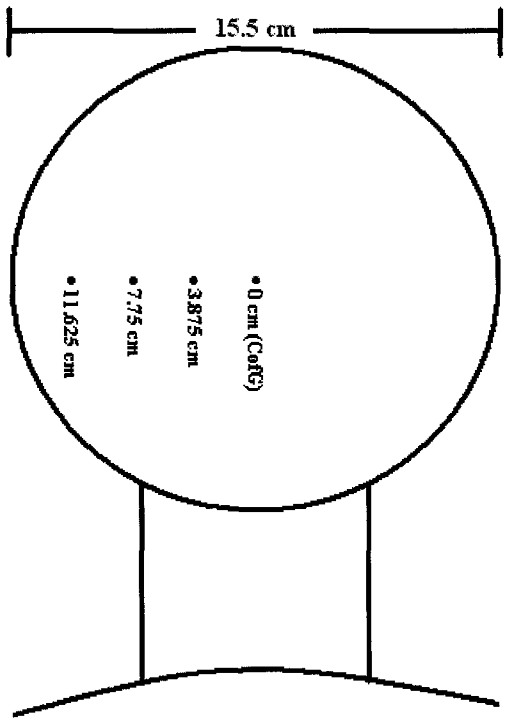

Figure 11: Impact locations on the headform to test the effects of impact deflection.

The effect of neck compliance was evaluated by impacting a 50th percentile Hybrid III head attached to three different Hybrid III necks with a linear impactor. The impacts were located 30 ± 1 mm above the intersection of the longitudinal plane and the reference plane of the headform (centre of gravity). Three sets of three impacts were performed for each neck at the following velocities: 5 m/s, 7 m/s and 9 m/s, which are comparable to impact velocities seen in football (Pellman, Viano, Tucker, Casson et al., 2003; Pellman et al., 2006).

To keep the results consistent, three impacts were done at 10 m/s prior to each day of testing. This allowed for the foam cap on the linear impactor to stabilize. Furthermore, the skin covering the rear of the head was removed. Impacts at high velocity caused it to displace, making readjustment time consuming. The removal of the cap showed to have no effect on previous pilot data.

Following each impact, peak linear and angular accelerations were recorded using a TDAS Pro Lab system (DTS, Seal Beach CA). A SAE J211 Class 1000 filter was used on the data obtained from the accelerometers. Each impact velocity was collected by a computer using National Instruments VI-Logger. All further data analysis was conducted by Bioproc 2 (developed by Dr. D.G.E. Robertson, University of Ottawa).

The impact velocities of the linear impactor were measured using a time gate which was validated using a High Speed Imaging PCI-512 Fastcam which recorded the impact at a frequency of 2000 Hz using Photron Motion Tools. The video of the impactor arm striker was then digitized prior to impact to produce distance traveled over time which was used to calculate impact velocity.

3.3 DEPENDANT VARIABLES

3.3.1 *Peak linear acceleration*

Linear acceleration was measured using the resultant of the nine accelerometers positioned in the headform.

3.3.2 *Peak angular acceleration*

Angular acceleration was measured using Padgaonkar's array which is also know as the 3-2-2 method, due to the nine uni-axial accelerometers positioned in an orthogonal arrangement (Padgaonkar et al., 1975). Rotational acceleration was computed using linear acceleration measures from each of the sensors while using the first principles of rigid body dynamics:

$$\alpha_x = (A_{z1} - A_{z0})/2\rho_{y1} - (A_{y3} - A_{y0})/2\rho_{z3} \quad (5)$$

$$\alpha_y = (A_{x3} - Ax_0)/2\rho_{z3} - (A_{z2} - A_{z0})/2\rho_{x2} \quad (6)$$

$$\alpha_z = (A_{y2} - A_{y0})/2\rho_{x2} - (A_{x1} - A_{x0})/2\rho_{y1} \quad (7)$$

Where α is the angular acceleration for component *i* (x,y,z), A_{ij} is the linear acceleration for component *i* (x, y, z) along orthogonal arm *j* (1, 2, 3), and ρ_{x2} is the length of orthogonal arm *j* acting along the component *i*.

Other techniques exist, such as the two-dimensional in-line method. For this system, five linear accelerometers are attached to an engineered surface inside the skull with their sensitive axes is aligned with the mid-sagittal plane of the headform. It is referred to as 2-D in-line because one set can only measure the rotational acceleration of one axis. As a result, three sets need to be installed in order to obtain tri-axial rotational acceleration measurements. Calculations are performed using a least-squares approximation of the rotational accelerations determined from the five accelerometers and corresponding moment arms about the CG (Newman et al., 2005).

Padgaonkar's array, the 2-D inline method as well as a direct rotational accelerometer were compared by impacting a Hybrid III headform equipped with all three independent systems (Newman et al., 2005). Results showed that although using a direct rotational accelerometer is simple, its inherent ringing makes it inadequate for these types of impacts. The other two methods showed similar peak rotational acceleration response in front impact differing only by 6%. Padgaonkar's method is more convenient because it only requires nine accelerometers while the in-line method requires 18. Furthermore, the computational algorithms for the 3-2-2-2 method are accurate, robust and based on physical principles. Lastly, the machined headform is commercially available (Newman et al., 2005).

3.3.3 GSI

The Gadd Severity Index equation was incorporated in a helmet standard by NOCSAE in the early 1970s. Its novelty was the incorporation of time duration within the equation. The equation is:

$$SI = \int_T a^{2.5}\, dt \tag{4}$$

Where *a* is the peak acceleration of the head form centre of gravity expressed in g. *t* is the duration of the impact in sec. The 2.5 weighing factor is based on the Wayne curve obtained by using animal impact data (Gadd, 1966).

This equation is important as it allows for the comparison of different waveforms. As seen in Figure 12, waveforms can vary, making it hard to compare injury potential. *Trace e*, for example, is a rectangular waveform with a peak of 54g while *trace f* is a triangular waveform peaking at 108g. A peak acceleration criterion would conclude that *trace f* is twice as injurious as *trace e*. On the other hand, a pulse-area criterion would conclude that both would have the same injury potential. Using GSI, a proper injury hazard can be calculated, concluding that *trace f* (861) is 1.61 times greater then *trace e* (535) (Gadd, 1966).

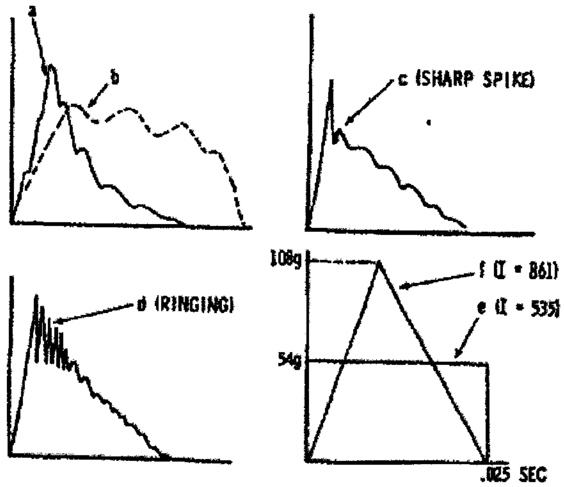

Figure 12: Six waveforms demonstrating the wide variation existing (Gadd, 1966).

3.4 RESEARCH DESIGN

The thesis was divided into two parts. The first part was a 4 x 2 fully crossed, balanced design which studied the effects of impact deflection on peak linear acceleration, peak angular acceleration and GSI (Table 2). Each condition was tested nine times.

The second part was a 3 x 3 fully crossed, balanced design which studied the effects of neck compliance on peak linear acceleration, peak angular acceleration and GSI (Table 3). Each condition was tested nine times.

3.4.1 Impact deflection

Table 2: Research design for impact deflection tests.

	B 1	B 2
A 1	A1 B1	A1 B2
A 2	A2 B1	A2 B2
A 3	A3 B1	A3 B2
A 4	A4 B1	A4 B2

Where:

A1 = 0 cm lateral displacement (Centre of gravity)

A2 = 3.875 lateral displacement

A3 = 7.75 cm lateral displacement

A4 = 11.625 cm lateral displacement

B1 = impact velocity of 5 m/s

B2 = impact velocity of 7 m/s

3.4.2 Neck compliance

Table 3: Research design for neck compliance tests.

	B 1	B 2	B 3
A 1	A1 B1	A1 B2	A1 B3
A 2	A2 B1	A2 B2	A2 B3
A 3	A3 B1	A3 B2	A3 B3

Where:

A1 = Soft neck

A2 = Median neck

A3 = Stiff neck

B1 = impact velocity of 5 m/s

B2 = impact velocity of 7 m/s

B3 = impact velocity of 9 m/s

3.5 STATISTICAL ANALYSIS

3.5.1 Impact deflection

In order to determine the effects of impact location, an analysis of covariance (ANCOVA) was performed on each dependant variable, using velocity as a covariate. This was done to identify the effect of lateral displacement across all velocities.

Further analysis was performed at each velocity (5 m/s and 7 m/s) using a simple one-way analysis of variance (ANOVA), followed by a post hoc test using Tukey's method. This was done to reveal significant differences between each impact location.

All statistical analysis will be performed using SPSS 11.5 software (SPSS Inc., Chicago IL, USA).

3.5.2 Neck compliance

In order to determine the effects of neck compliance, an analysis of covariance (ANCOVA) was performed on each dependant variable, using velocity as a covariate. This was done to identify the effect of neck compliance across all velocities.

Further analysis was performed at each velocity (5 m/s, 7 m/s and 9 m/s) using a simple one-way analysis of variance (ANOVA), followed by a post hoc test using Tukey's method. This was done to reveal significant differences between each neck compliances.

All statistical analysis will be performed using SPSS 11.5 software (SPSS Inc., Chicago IL, USA).

CHAPTER 4

RESULTS

4.1 IMPACT DEFLECTION

Nine impacts were completed at each impact location (0 cm, 3.88 cm, 7.75 cm and 11.63 cm) at both velocities (5 m/s and 7 m/s). The results indicated that impact deflection had a significant effect on peak linear acceleration ($F_{(3, 67)}$ = 176.32, p< .001), peak angular acceleration ($F_{(3, 67)}$ = 148.19, p< .001) and GSI ($F_{(3, 67)}$ = 91.83, p< .001). The results showed that an increase in lateral displacement caused a decrease in peak linear acceleration, peak angular acceleration and GSI. This signifies that impact deflection may successfully reduce injury risk. Further analysis was performed to determine the effect on each variable at the different velocities. The mean and standard deviation are shown in Tables 4 and 5.

Table 4: Peak linear acceleration and peak angular acceleration, mean ± 1 standard deviation, for front impacts at different lateral locations at 5 m/s.

Lateral displacement	Linear acceleration (g)	Angular acceleration (rad/s^2)	GSI
0 cm (centre of gravity)	79.1 ± 3.7	6925 ± 471	210 ± 8
3.88 cm	71.8 ± 5.5	6402 ± 680	187 ± 11
7.75 cm	44.9 ± 1.7	3441 ± 622	115 ± 5
11.63 cm	29.1 ± 1.9	2711 ± 392	32 ± 5

Table 5: Peak linear acceleration and peak angular acceleration, mean ± 1 standard deviation, for front impacts at different lateral locations at 7 m/s.

Lateral displacement	Linear acceleration (g)	Angular acceleration (rad/s^2)	GSI
0 cm (centre of gravity)	134.4 ± 9.4	11975 ± 788	541 ± 24
3.88 cm	118.1 ± 4.6	10010 ± 979	517 ± 16
7.75 cm	92.6 ± 4.0	7013 ± 680	372 ± 15
11.63 cm	38.9 ± 3.7	3660 ± 150	52 ± 9

Lateral displacement had a significant effect on peak linear acceleration at 5 m/s ($F_{(3, 32)}$ = 130.06, p< .001) and 7 m/s ($F_{(3, 32)}$ = 450.38, p< .001) (Fig. 13). Post hoc tests, using Tukey's method, indicated significant differences between all impact locations (p< .001).

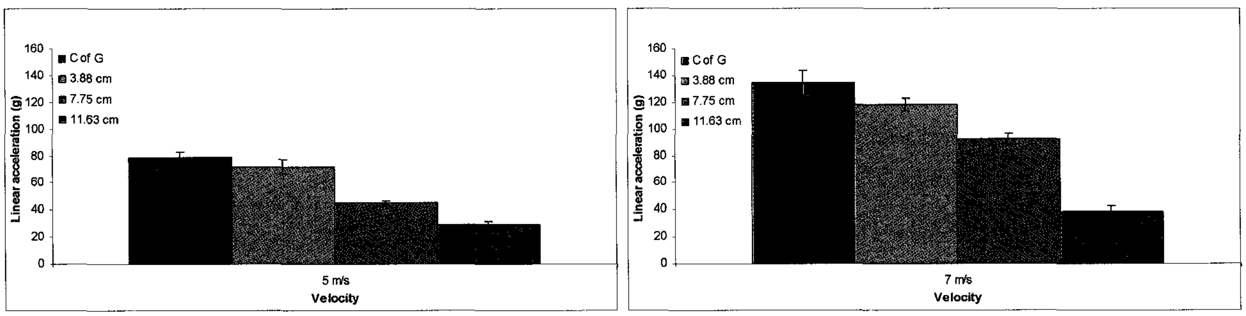

Figure 13: Comparison of peak linear accelerations (± 1 SD) between four head displacements at 5 m/s (left) and 7 m/s (right).

Lateral displacement also had a significant effect on peak angular acceleration at 5 m/s ($F_{(3, 32)}$ = 22.86, p< .001) and 7 m/s ($F_{(3, 32)}$ = 229.97, p< .001) (Fig. 14). Post hoc tests, using Tukey's method, indicated significant differences between all impacts locations (p< .001) except between impacts through the centre of gravity and impacts with a 3.88 cm displacement at 5 m/s.

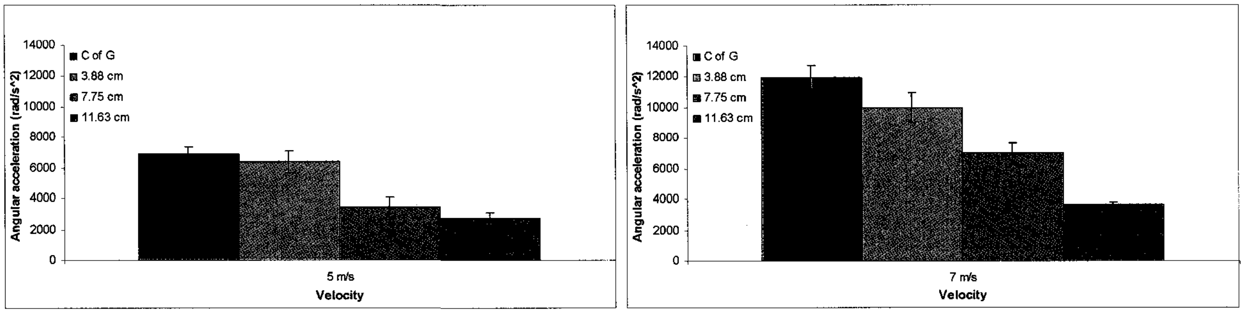

Figure 14: Comparison of peak angular accelerations (± 1 SD) between four head displacements at 5 m/s (left) and 7 m/s (right).

Likewise, lateral displacement had a significant effect on GSI at 5 m/s ($F_{(3, 32)} = 1024.32$, $p < .001$) and 7 m/s ($F_{(3, 32)} = 1570.75$, $p < .001$) (Fig. 15). Post hoc tests, using Tukey's method, indicated significant differences between all impacts locations ($p < .001$) except between impacts through the centre of gravity and impacts with a 3.88 cm displacement at 7 m/s ($p < .05$).

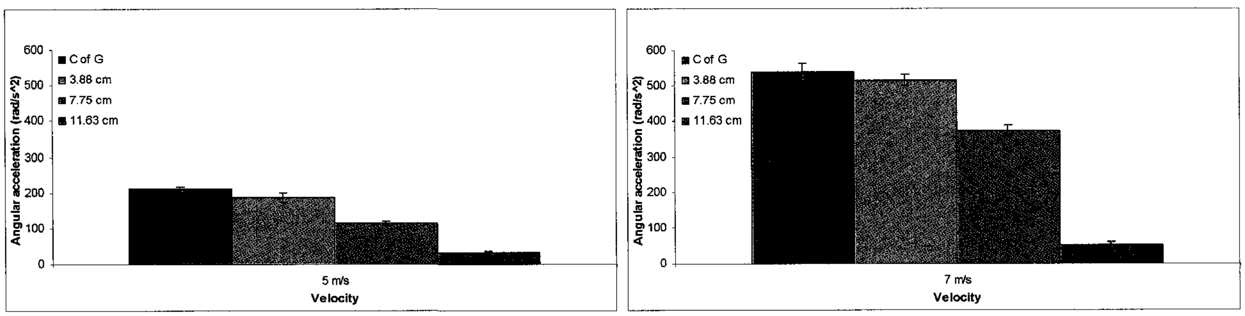

Figure 15: Comparison of GSI (± 1 SD) between four head displacements at 5 m/s (left) and 7 m/s (right).

4.2 NECK COMPLIANCE

Nine impacts were completed for each neck compliances (soft, median and stiff neck) at each velocity (5 m/s, 7 m/s and 9 m/s). The results revealed that neck compliance had a significant effect on linear acceleration ($F_{(2, 77)} = 9.20$, $p < .001$) and peak angular acceleration ($F_{(2, 77)} = 8.14$, $p < .001$); however, there were no significant differences for GSI ($F_{(2, 77)} = .26$, $p > .05$). The results showed that an increase in neck compliance caused a decrease in peak linear acceleration but an increase in peak angular acceleration. Further analysis was performed to determine the effect on each variable at the different velocities. The mean and standard deviation are shown in Tables 6 to 8.

Table 6: Peak linear acceleration, peak angular acceleration and GSI, mean ± 1 standard deviation, for front impacts using three neck compliances at 5 m/s.

Neck compliance	Linear acceleration (g)	Angular acceleration (rad/s^2)	GSI
Soft	78.4 ± 1.8	7443 ± 395	214 ± 14
Median	79.1 ± 3.7	6925 ± 471	210 ± 8
Stiff	82.0 ± 2.6	6114 ± 390	205 ± 7

Table 7: Peak linear acceleration, peak angular acceleration and GSI, mean ± 1 standard deviation, for front impacts using three neck compliances at 7 m/s.

Neck compliance	Linear acceleration (g)	Angular acceleration (rad/s^2)	GSI
Soft	137.3 ± 4.6 g	12530 ± 1173 rad/s^2	628 ± 31
Median	134.4 ± 9.4 g	11975 ± 788 rad/s^2	541 ± 24
Stiff	140.0 ± 3.6 g	11465 ± 1187 rad/s^2	585 ± 17

Table 8: Peak linear acceleration, peak angular acceleration and GSI, mean ± 1 standard deviation, for front impacts using three neck compliances at 9 m/s.

Neck compliance	Linear acceleration (g)	Angular acceleration (rad/s^2)	GSI
Soft	210.9 ± 4.4 g	15423 ± 2783 rad/s^2	2486 ± 565
Median	214.4 ± 4.3 g	15166 ± 2540 rad/s^2	2422 ± 491
Stiff	230.9 ± 5.1 g	12608 ± 1416 rad/s^2	2286 ± 267

Neck compliance had a significant effect on peak linear acceleration at 5 m/s ($F_{(2, 24)}$ = 4.16, p< .05) and 9 m/s ($F_{(2, 24)}$ = 4.04, p<0.05). The tests showed no significant effect at 7 m/s ($F_{(2, 24)}$ = 1.69, p> .05) (Fig. 16). At 5 m/s, post hoc tests, using Tukey's method, indicated significant differences only between the soft and the stiff neck (p< .05). At 9 m/s, post hoc tests, using Tukey's method, indicated significant differences between the stiff neck and the two other necks (p< .001).

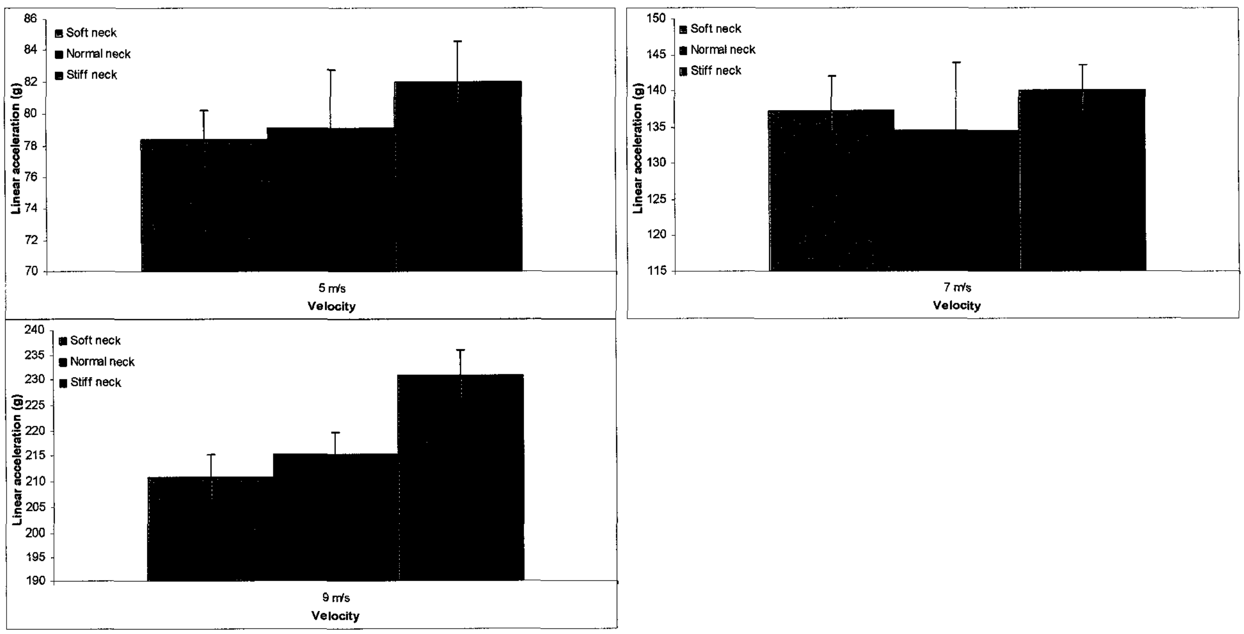

Figure 16: Comparison of peak linear accelerations (± 1 SD) between three neck compliances at 5 m/s (top left), 7 m/s (top right) and 9 m/s (bottom left).

Neck compliance also had a significant effect on peak angular acceleration at 5 m/s ($F_{(2, 24)} = 22.86$, $p < .001$) and 9 m/s ($F_{(2, 24)} = 45.82$, $p < .001$). The tests showed no significant effect at 7 m/s ($F_{(2, 24)} = 2.25$, $p > .05$) (Fig. 17). At 5 m/s, post hoc tests, using Tukey's method, indicated significant differences between the soft neck and the median neck ($p < .05$); between the soft neck and the stiff neck ($p < .001$); and between the median neck and the stiff neck ($p < .001$). At 9 m/s, post hoc tests, using Tukey's method, indicated significant differences only between the soft and the stiff neck ($p < .05$).

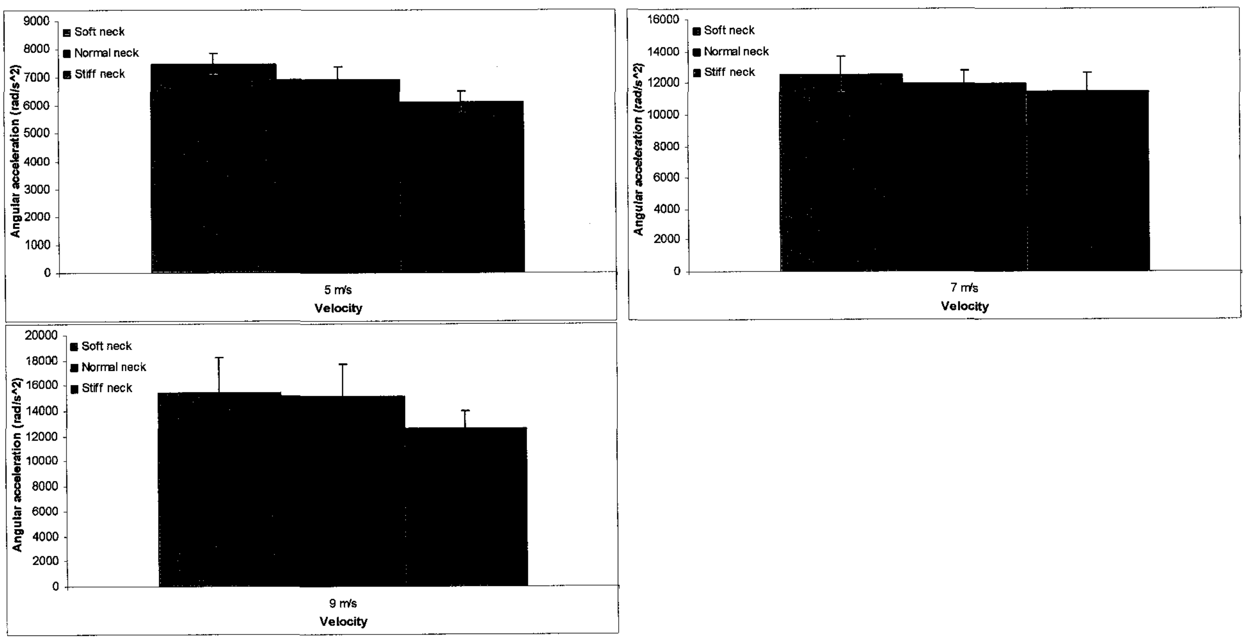

Figure 17: Comparison of peak angular accelerations (± 1 SD) between three neck compliances at 5 m/s (top left), 7 m/s and 9 m/s (bottom left).

CHAPTER 5

DISCUSSION

The purpose of this study was to examine the effects of neck compliance and head displacement on peak linear acceleration, peak angular acceleration and GSI. This study was the first to examine these conditions and compare the data in attempts to understand injury mechanism.

5.1 IMPACT DEFLECTION

As expected, peak linear acceleration, peak angular acceleration and GSI decreased when the impacts got further from the centre of gravity. This was expected because these types of collisions do not engage the headform fully, meaning that the total momentum is not transferred to the head.

Figure 18 shows a comparison of peak linear and angular acceleration using brain injury thresholds determined by Zhang and colleagues (2004). At that velocity, impacts through the centre of gravity and with a 3.875 cm displacement were above the 80% risk of injury. Impacts with a 7.75 cm and a 11.625 cm displacement were slightly above 50% and below 25%, respectively.

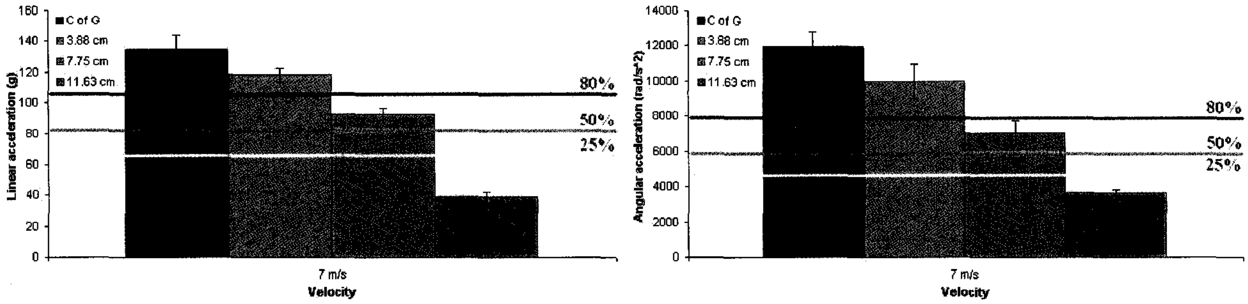

Figure 18: Peak linear acceleration and peak angular acceleration measured during front impacts through the centre of gravity and three displacements at 7 m/s. The lines represent a 25%, 50% and 80% probability of sustaining an mTBI.

Many studies have found that impacts to the side of the head, i.e. along the y-axis, are more prone to causing injuries than impacts to the forehead (Kleiven, 2003; Zhang et al., 2001; Zhang et al., 2004; Delaney, Puni & Rouah, 2006). Figure 19 demonstrates that although angular accelerations along the x-axis that although angular accelerations along the x-axis do increase for eccentric impacts, they remain well below injury risk (Zhang et al., 2004). By comparison, angular accelerations along the y-axis were higher reaching over 50% brain injury risk when impacted through the centre of gravity. This means that even a small movement way from the centre of gravity of the head can effectively reduce the risk of sustaining a head injury.

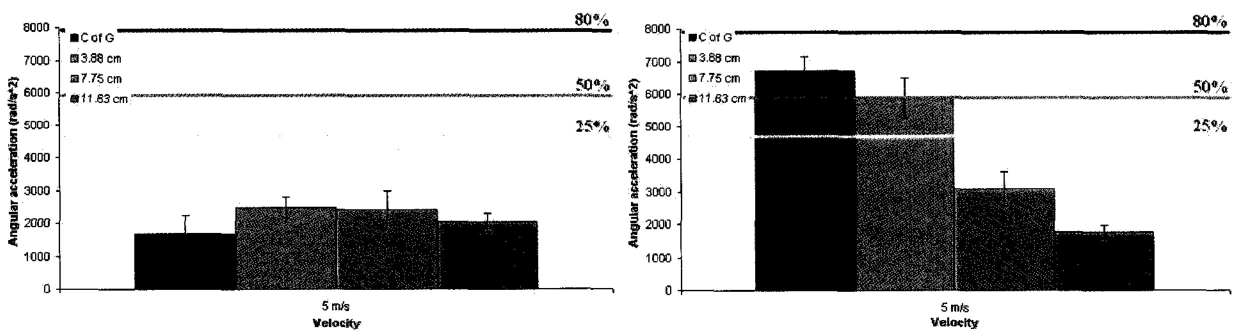

Figure 19: Peak angular acceleration along the x-axis (left) and the y-axis (right) measured during front impacts through the centre of gravity and three displacements at 5 m/s. The lines represent a 25% and 50% probability of sustaining an mTBI.

5.2 NECK COMPLIANCE

Contrary to what has been previously reported in the literature (Cantu & Mueller, 2003a; Johnston et al., 2001; Levy et al., 2004; Patel et al., 2005), a stronger, stiffer neck did not reduce peak linear acceleration of the head. It was expected that the higher effective mass generated by a stiffer neck would help absorb the impact, yet the results exposed a positive relationship between neck compliance and linear acceleration. On the contrary, results revealed a negative relationship between neck compliance and angular acceleration. The stiffer neck successfully resisted head rotation, yet it failed in reducing linear acceleration. This implies that forces acting on the head upon contact need to somehow be absorbed and that the compliance of the neck plays a role on how this is accomplished. More compliance will lead to a higher generation of angular acceleration, while less compliance will produce more linear acceleration. Considering this, both forms of acceleration need to be considered in order to account for injury risk associated with neck compliance.

Studies by Deng and Goldsmith (1987) and Kabo & Goldsmith (1983) did not intend to study neck compliance; nevertheless, nevertheless, the data they collected supports the results reported in this study. In their study, they compared linear accelerations produced by impacts to the front and to the back of the head, which corresponds to higher and lower neck compliance. Their conclusion, which was that impacts to the rear of the head produced higher linear acceleration, is in accordance with the data in this thesis. Svensson & Lövsund (1992) studied the effect of compliance by comparing the Hybrid III neck to BioRid necks with different neck stiffness. Their results demonstrated that an increase in neck stiffness resulted in a reduction in angular acceleration.

Figure 20 shows a comparison of peak linear and angular acceleration as well as thresholds published by Zhang and his colleagues (2004). It shows that peak linear accelerations measured when using the stiffer neck were associated with a probability of 50% of sustaining an mTBI. This would suggest that players with tense neck muscles prior to an impact would be at a higher risk of sustaining a head injury then those who have not contracted their muscles. Conversely, peak angular accelerations measured when using the softer neck were near an 80% probability of sustaining an mTBI. Even though a compliant neck produces lower linear accelerations, it can produce high angular accelerations that may put a player at a higher risk of sustaining a head injury. The positive relationship between neck compliance and GSI supports the previous statement.

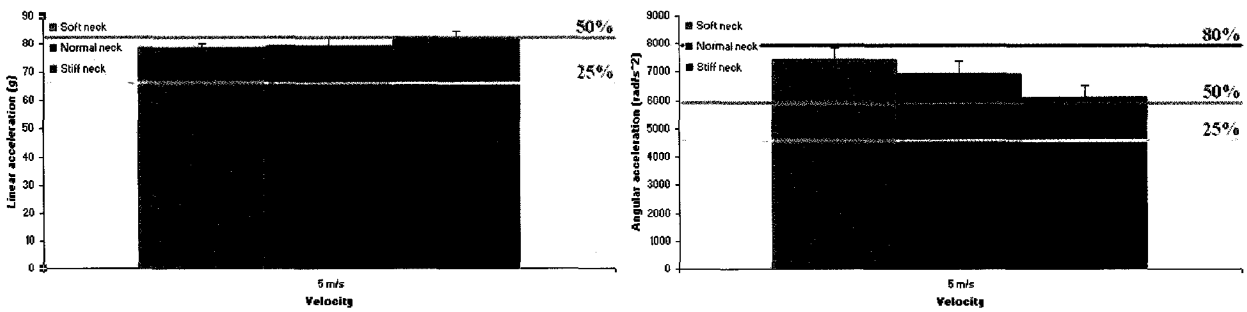

Figure 20: Peak linear acceleration and peak angular acceleration measured during a direct impact to the front of a Hybrid III head attached to different necks at 5 m/s. The lines represent a 25%, 50% and 80% probability of sustaining an mTBI.

5.3 PREVENTION

The data presented in this thesis provides a better understanding of the relationship between risk of brain injury and strategies for injury prevention. Training methods (strength and training), administrative intervention (rules and regulations) and protective equipment design (helmet) all benefit from a better understanding of the mechanism of injury. These results support training neck muscles to improve strength in order to reduce the risk of mTBI (Bailes & Cantu, 2001; Barthe et al., 2001). Impact deflection or avoidance should be considered as a strategy to prevent brain injuries resulting from head impacts. Skill training contributes to better ball tracking, helps avoid contact and increases the chance of impact deflection, thus reduces the risk of mTBI (McIntosh & McCrory, 2005).

Rules are one of the most common methods to prevent injury (McIntosh & McCrory, 2005). The results underline the importance of establishing and enforcing rules to protect players in vulnerable positions, where the athletes are unable to protect themselves. Such rules have already been put in motion for other dangerous actions such as spearing and have successfully reduced injury rates (Boden et al., 2006; Cantu & Mueller, 2003a, 2003b; Torg et al., 2002).

While the data collected in this study was not done using helmets, the results presented are important to their design. Helmet manufacturers need to consider the importance of avoidance as a mean to decrease head injuries in sport. Larger helmets are certainly helpful in managing impacts; however, they considerably reduce the ability of the athletes to avoid the impact force. It is clear that, once the relationship is better understood, a compromise between size and protection needs be established.

In addition, the opposite response of angular and linear head accelerations to the effect of increased neck compliance during impact was an unexpected finding. Linear acceleration increased with a decrease in neck compliance while angular acceleration decreased. This provides important insight into the effect of the neck on the dynamic response of the head during impact. Helmet engineers seldom test the efficacy of their designs using a neck and head apparatus and generally assume that a decrease in linear acceleration is associated with a decrease in angular accelerations. This has been shown to be untrue and new testing methods should be devised.

CHAPTER 6

CONCLUSION

The objective of this paper was to determine the influence of impact deflection and neck compliance on peak linear acceleration, peak rotational acceleration and GSI during a front impact to a Hybrid III head. As expected, an increase in lateral displacement caused a reduction in peak linear and angular accelerations and GSI. Furthermore, an increase in neck stiffness caused a decrease in peak angular acceleration; however, an increase in neck stiffness resulted in an increase in peak linear acceleration, while having no significant effect on GSI. It is expected that this information will be used to help injury prevention through training, rules and equipment design.

6.1 RESEARCH HYPOTHESIS

6.1.1 Impact deflection

1. An increase in impact deflection will result in a decrease in peak linear acceleration.
    i. Accepted
2. An increase in impact deflection will result in a decrease in peak angular acceleration.
    i. Accepted
3. An increase in impact deflection will result in a decrease in GSI.
    i. Accepted

6.1.2 Neck compliance

4. An increase in neck stiffness will result in a decrease in peak linear acceleration.

    i. Rejected

    5. An increase in neck stiffness will result in a decrease in peak rotational acceleration.

    i. Accepted

    6. An increase in neck stiffness will result in a decrease in GSI.

    i. Rejected

6.2 SUMMARY

This thesis will hopefully contribute to a reduction in head injury risk by providing a better understanding of the effects of impact deflection and neck compliance. It is expected that this information can be used to help injury prevention through training, rules and equipment design.

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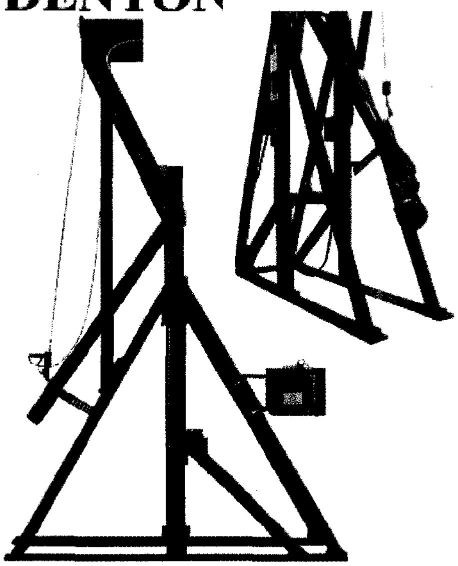
APPENDIX A

Denton ATD, Inc. Neck Pendulum Test Stand

Model Number: TF-200-0000

DENTON



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

Soft Hybrid III (1820) Neck Calibration Results

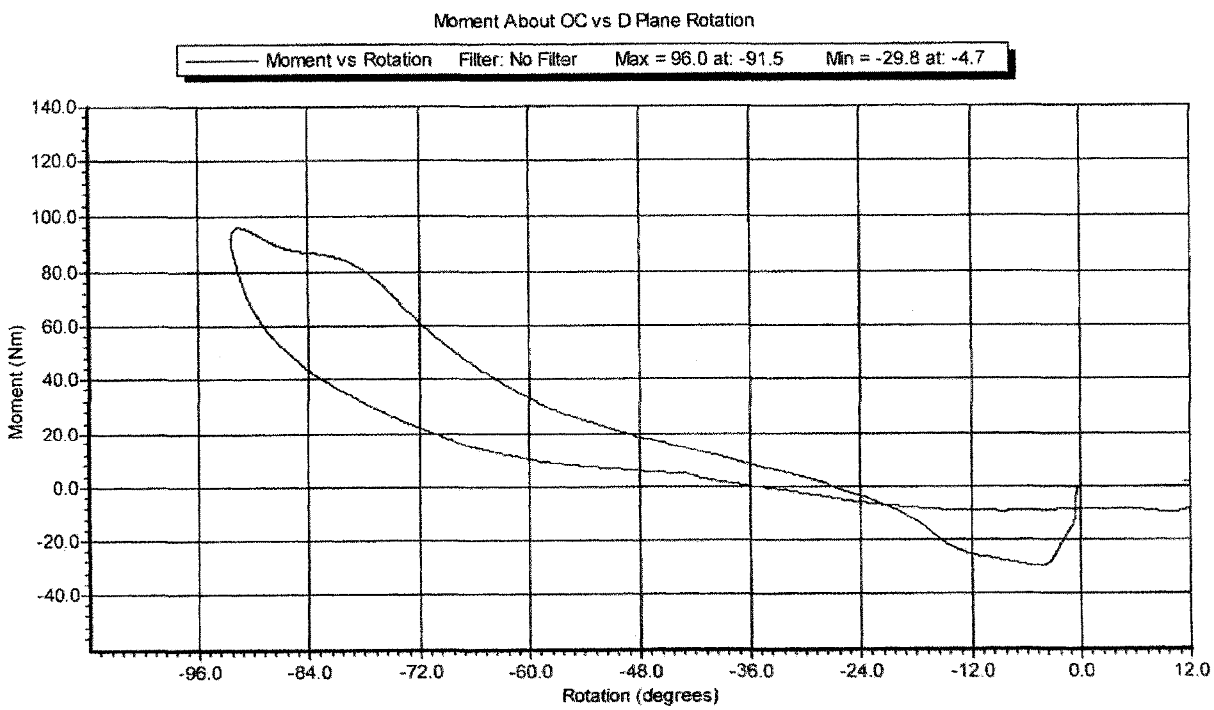
Test Name: **Neck Flexion**
Dummy type: **Hybrid III 50th**
Test ID: **AA1820-2**
Test Number: **2**

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.**



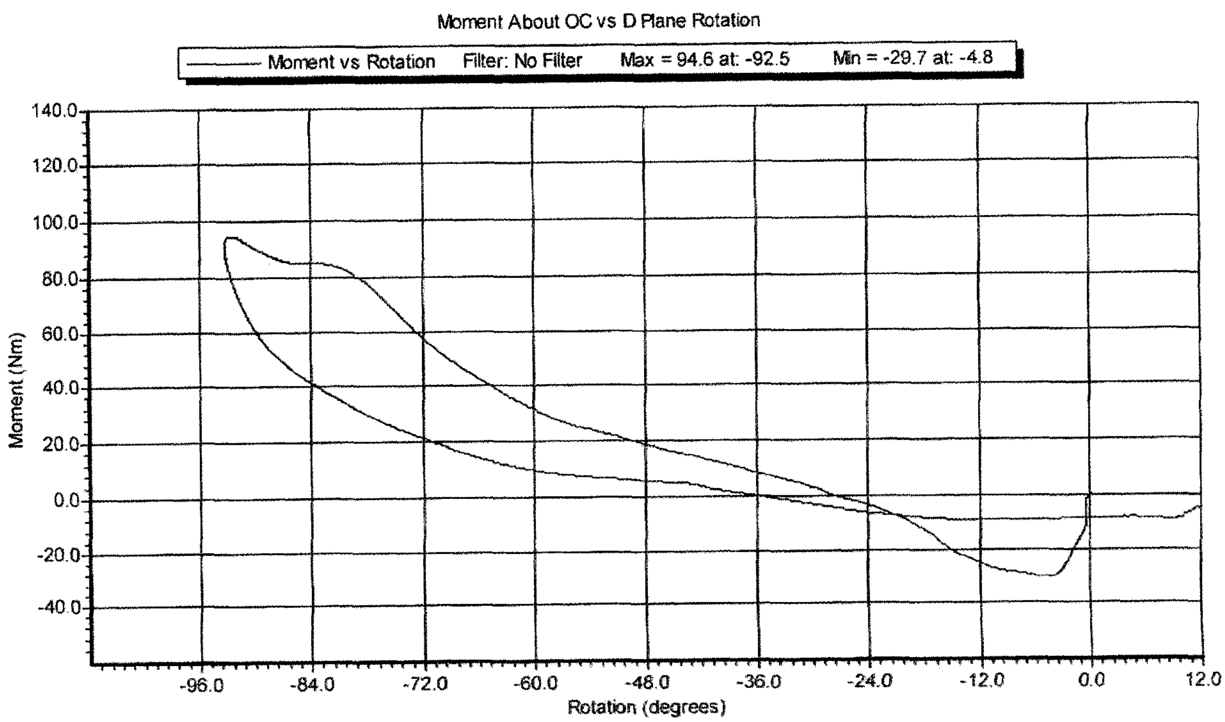
Test Name: **Neck Flexion**
Dummy type: **Hybrid III 50th**
Test ID: **AA1820-3**
Test Number: **3**

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	F
Humidity			10 – 70	**56** %RH	F
Velocity			6.89 – 7.13	**7.00** m/s	F
Pendulum Deceleration at	10	ms	22.5 – 27.5	**24.6** g	F
Pendulum Deceleration at	20	ms	17.6 – 22.6	**21.0** g	F
Pendulum Deceleration at	30	ms	12.5 – 18.5	**17.6** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 29.0	**17.6** g	F
Decel Time to	5	g	34.0 – 42.0	**37.4** ms	F
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	F
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.**



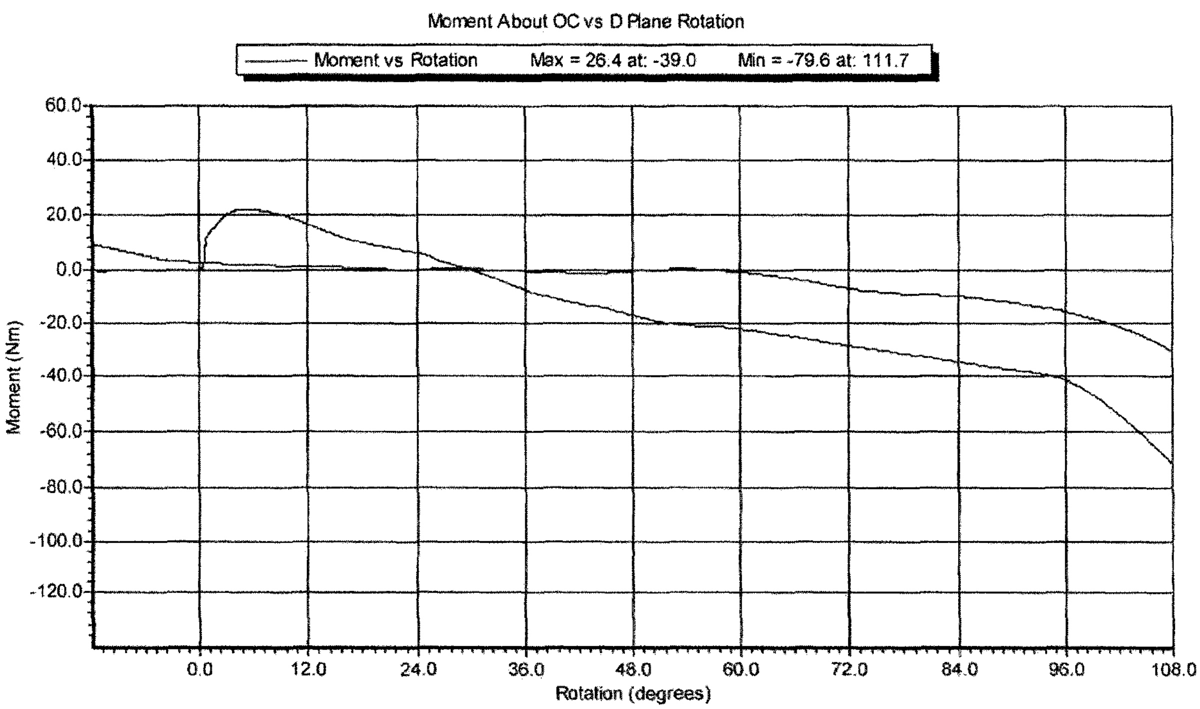
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1820-1**
Test Number: **1**

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.**



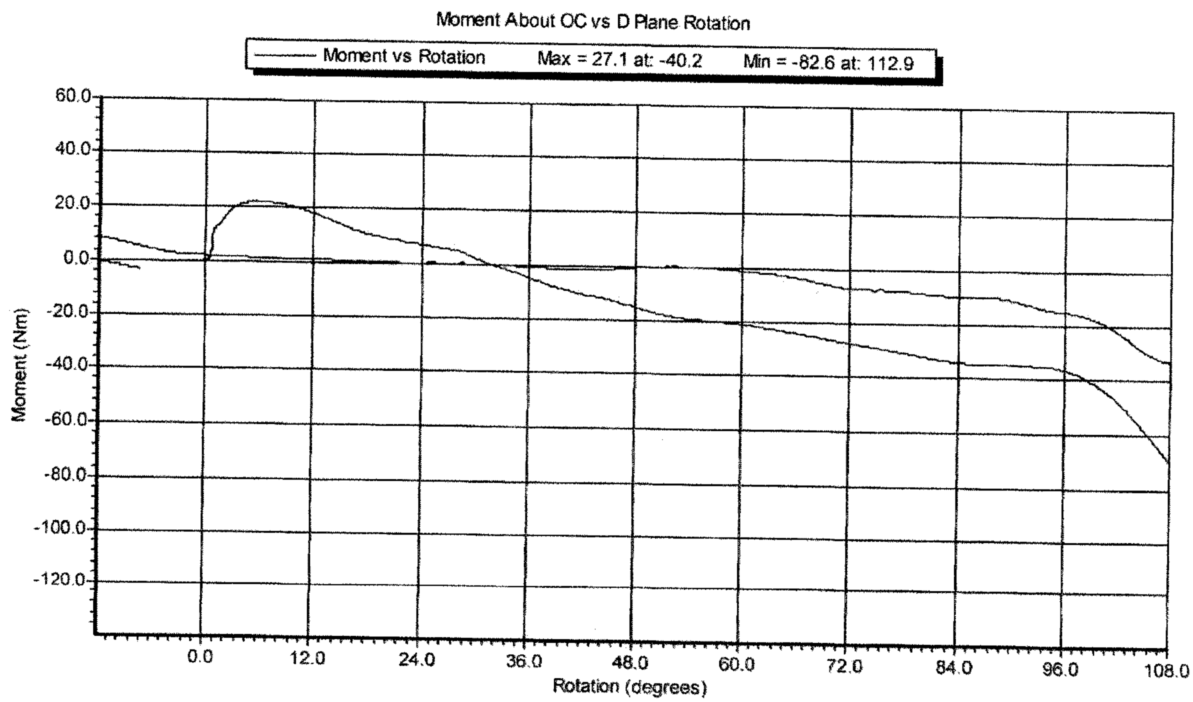
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1820-2**
Test Number: 2

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	F
Humidity			10 – 70	**58** %RH	F
Velocity			5.94 – 6.19	**5.99** m/s	F
Pendulum Deceleration at	10	ms	17.2 – 21.2	**19.7** g	F
Pendulum Deceleration at	20	ms	14.0 – 19.0	**18.4** g	F
Pendulum Deceleration at	30	ms	11.0 – 16.0	**15.2** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 22.0	**15.2** g	F
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	F
Rotation Decay to Zero			147.0 – 174.0	**166.1** ms	F
Moment About Occipital Condyle			-80.0 – -52.9	**-82.6** Nm	F
Time at Max Moment			65.0 – 79.0	**73.1** ms	F
Moment Decay to Zero			120.0 – 148.0	**130.9** ms	F

Technician: **GS**
Company: **Denton ATD, Inc.**



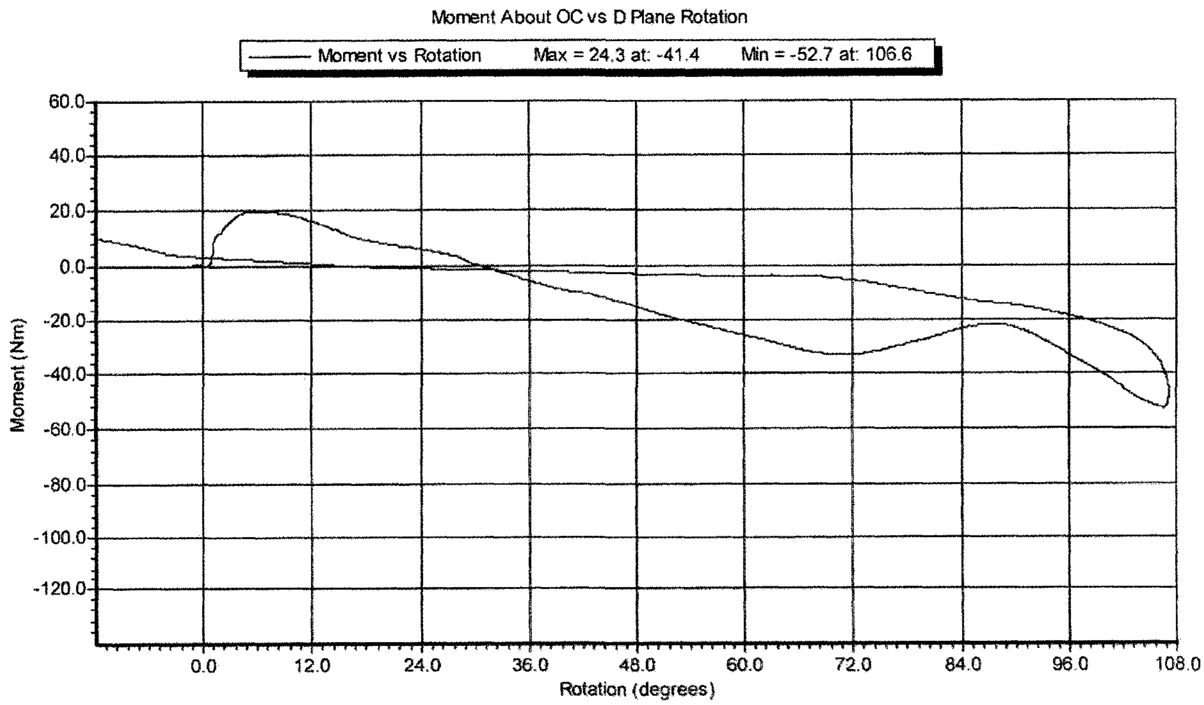
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1820-4**
Test Number: **4**

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.**



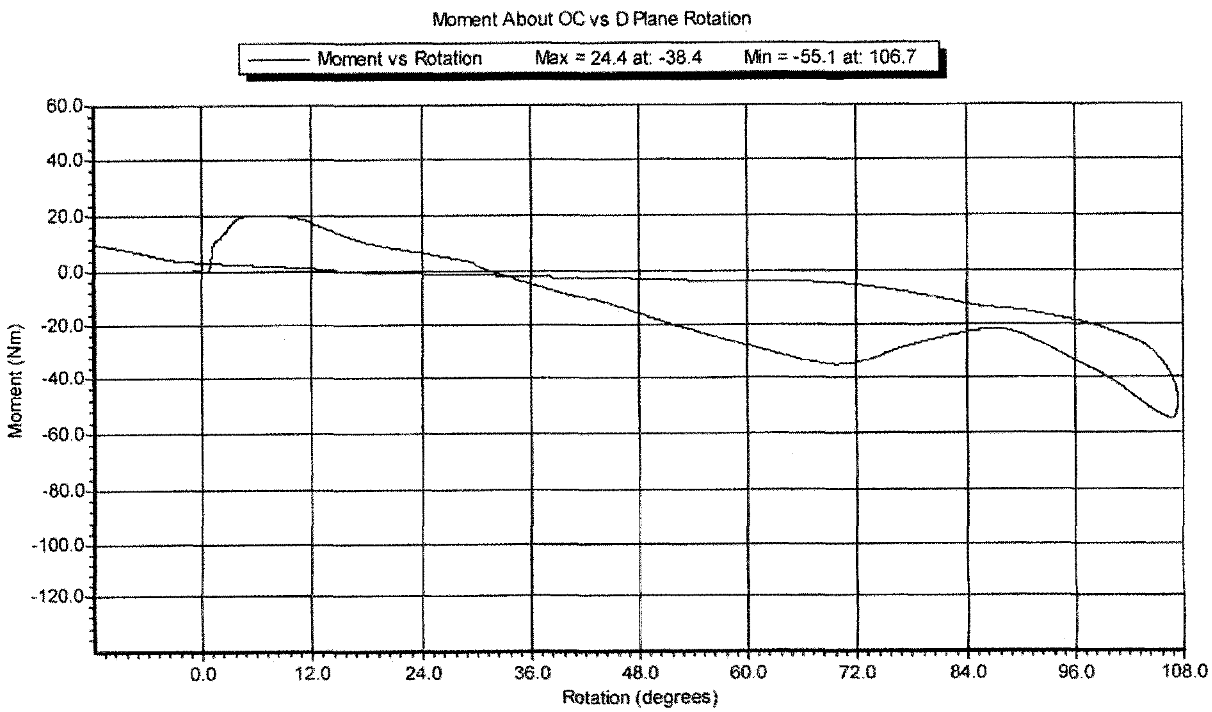
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1820-5**
Test Number: **5**

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	F
Humidity			10 – 70	**50** %RH	F
Velocity			5.94 – 6.19	**4.91** m/s	F
Pendulum Deceleration at	10	ms	17.2 – 21.2	**19.7** g	F
Pendulum Deceleration at	20	ms	14.0 – 19.0	**17.0** g	F
Pendulum Deceleration at	30	ms	11.0 – 16.0	**10.7** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 22.0	**10.7** g	F
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	F
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:

Manufacturer	Model	Serial Number	Calibration Date
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**

Stiff Hybrid III (1821) Neck Calibration Results

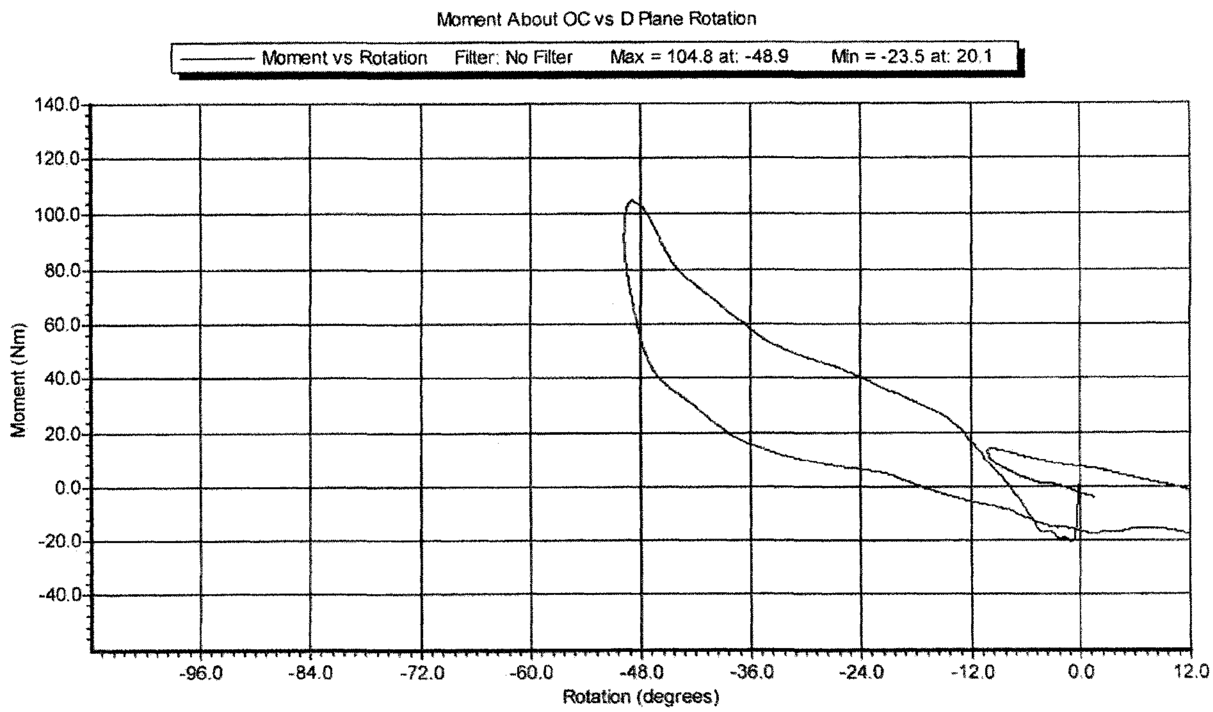
Test Name: **Neck Flexion**
Dummy type: **Hybrid III 50th**
Test ID: **AA1821-1**
Test Number: **1**

Test Date: **8/28/2007**
Test time: **9:51:53 AM**

Comments: **Hard rubber.**

Test Parameters			Test Specifications	Test results (avg)	
Temperature			20.6 – 22.2	**21.3** deg C	F
Humidity			10 – 70	**53** %RH	F
Velocity			6.89 – 7.13	**6.97** m/s	F
Pendulum Deceleration at	10	ms	22.5 – 27.5	**23.0** g	F
Pendulum Deceleration at	20	ms	17.6 – 22.6	**21.9** g	F
Pendulum Deceleration at	30	ms	12.5 – 18.5	**16.1** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 29.0	**16.1** g	F
Decel Time to	5	g	34.0 – 42.0	**39.8** ms	F
D Plane Rotation			-78.0 – -64.0	**-49.8** degrees	F
Time at Max Rotation			57.0 – 64.0	**51.3** ms	F
Rotation Decay to Zero			113.0 – 128.0	**98.6** ms	F
Moment About Occipital Condyle			88.1 – 108.4	**104.8** Nm	F
Time at Max Moment			47.0 – 58.0	**47.3** ms	F
Moment Decay to Zero			97.0 – 107.0	**84.9** ms	F

Technician: **GS**
Company: **Denton ATD, Inc.**



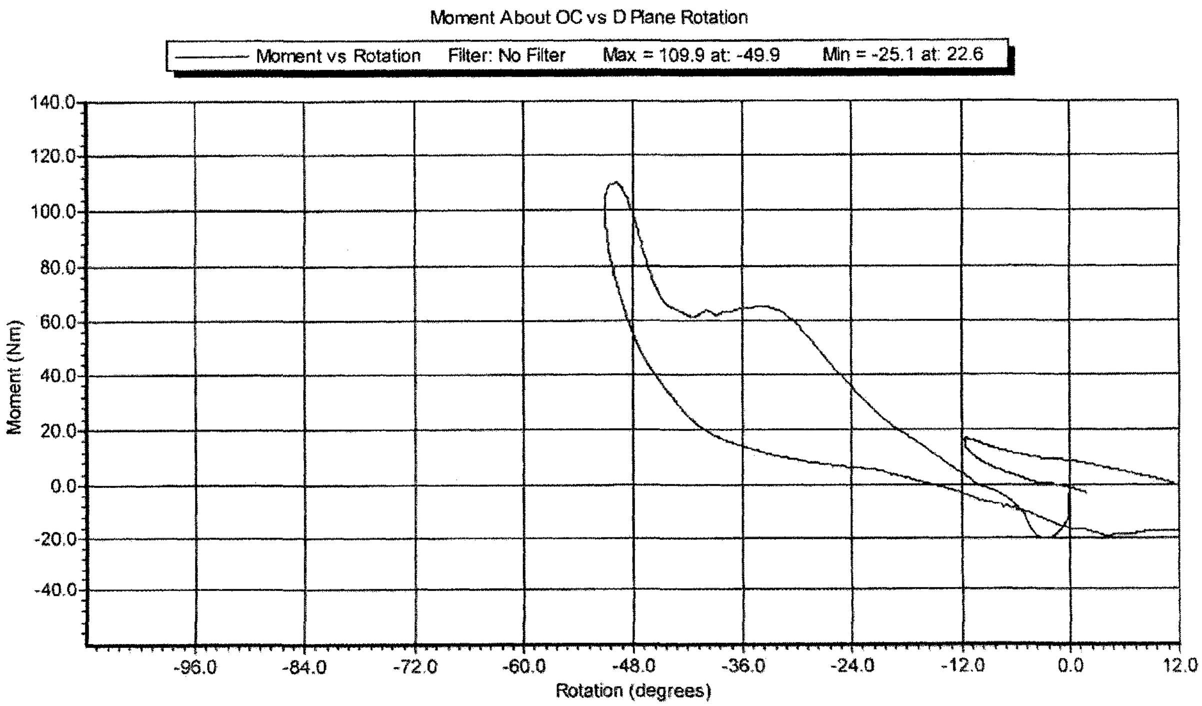
Test Name: **Neck Flexion**
Dummy type: **Hybrid III 50th**
Test ID: **AA1821-4**
Test Number: **4**

Test Date: **8/28/2007**
Test time: **12:58:26 PM**

Comments: **Hard rubber.**

Test Parameters			Test Specifications	Test results (avg)
Temperature			20.6 – 22.2	**21.1** deg C F
Humidity			10 – 70	**54** %RH F
Velocity			6.89 – 7.13	**6.97** m/s F
Pendulum Deceleration at	10	ms	22.5 – 27.5	**23.0** g F
Pendulum Deceleration at	20	ms	17.6 – 22.6	**20.2** g F
Pendulum Deceleration at	30	ms	12.5 – 18.5	**18.3** g F
Max Pendulum Deceleration after	30	ms	0.0 – 29.0	**18.3** g F
Decel Time to	5	g	34.0 – 42.0	**36.9** ms F
D Plane Rotation			-78.0 – -64.0	**-51.1** degrees F
Time at Max Rotation			57.0 – 64.0	**49.0** ms F
Rotation Decay to Zero			113.0 – 128.0	**97.1** ms F
Moment About Occipital Condyle			88.1 – 108.4	**109.9** Nm F
Time at Max Moment			47.0 – 58.0	**45.7** ms F
Moment Decay to Zero			97.0 – 107.0	**86.1** ms F

Technician: **GS**
Company: **Denton ATD, Inc.**



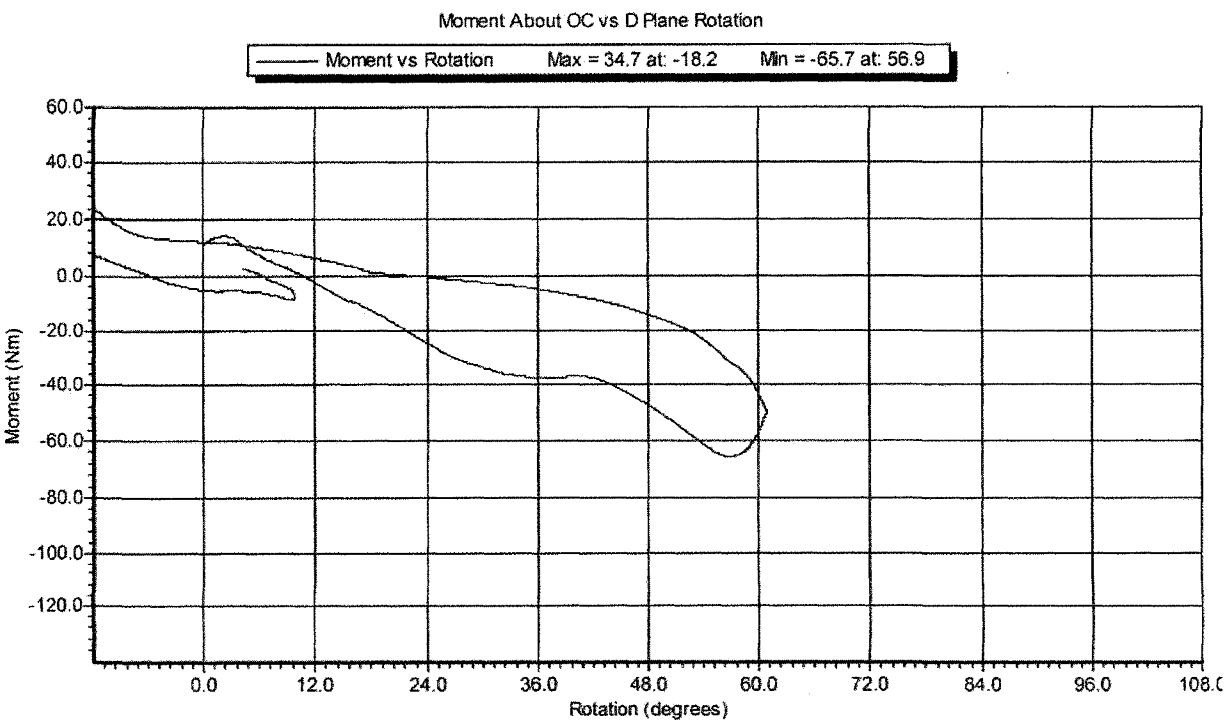
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1821-1**
Test Number: **1**

Test Date: **8/28/2007**
Test time: **2:16:26 PM**

Comments: **Hard rubber.**

Test Parameters			Test Specifications	Test results (avg)	
Temperature			20.6 – 22.2	**21.7** deg C	F
Humidity			10 – 70	**58** %RH	F
Velocity			5.94 – 6.19	**5.99** m/s	F
Pendulum Deceleration at	10	ms	17.2 – 21.2	**18.7** g	F
Pendulum Deceleration at	20	ms	14.0 – 19.0	**17.7** g	F
Pendulum Deceleration at	30	ms	11.0 – 16.0	**15.5** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 22.0	**15.5** g	F
Decel Time to	5	g	38.0 – 46.0	**40.6** ms	F
D Plane Rotation			81.0 – 106.0	**60.9** degrees	F
Time at Max Rotation			72.0 – 82.0	**65.5** ms	F
Rotation Decay to Zero			147.0 – 174.0	**131.7** ms	F
Moment About Occipital Condyle			-80.0 – -52.9	**-65.7** Nm	F
Time at Max Moment			65.0 – 79.0	**53.8** ms	F
Moment Decay to Zero			120.0 – 148.0	**112.4** ms	F

Technician: **GS**
Company: **Denton ATD, Inc.**



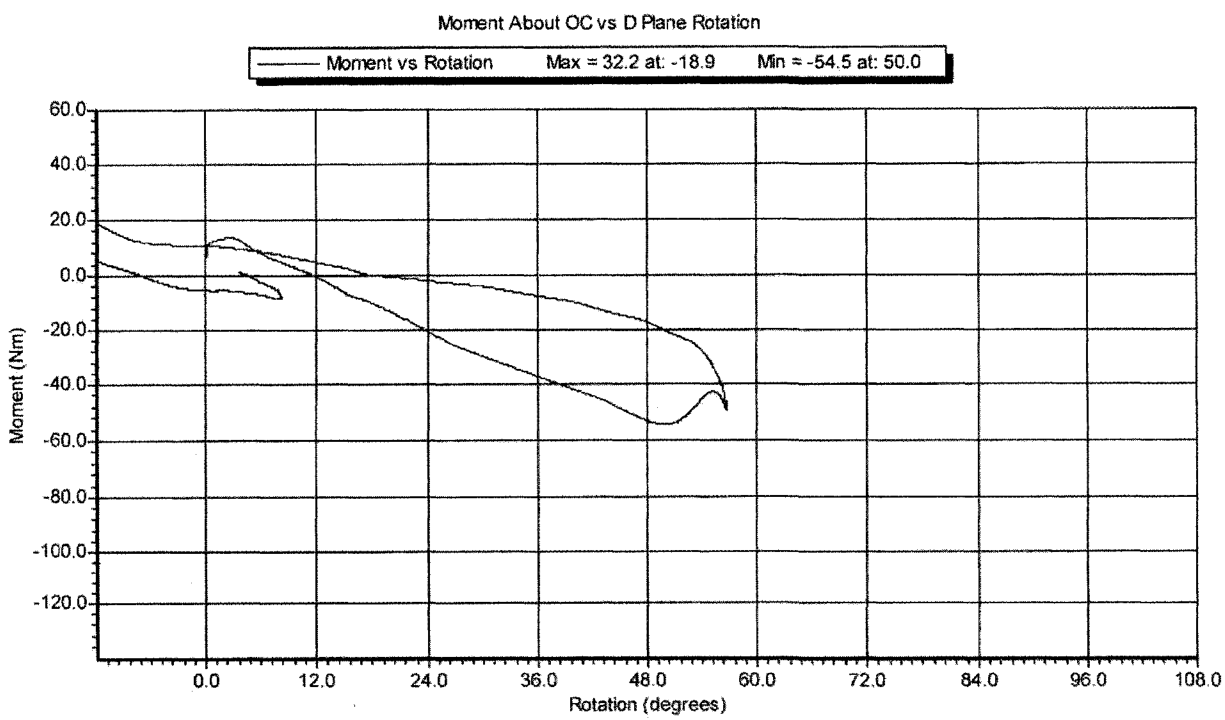
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1821-3**
Test Number: **3**

Test Date: **8/28/2007**
Test time: **4:16:50 PM**

Comments: **Hard rubber. Modified velocity.**

Test Parameters			Test Specifications	Test results (avg)	
Temperature			20.6 – 22.2	**21.3** deg C	F
Humidity			10 – 70	**51** %RH	F
Velocity			5.94 – 6.19	**4.93** m/s	F
Pendulum Deceleration at	10	ms	17.2 – 21.2	**18.7** g	F
Pendulum Deceleration at	20	ms	14.0 – 19.0	**16.5** g	F
Pendulum Deceleration at	30	ms	11.0 – 16.0	**14.6** g	F
Max Pendulum Deceleration after	30	ms	0.0 – 22.0	**14.6** g	F
Decel Time to	5	g	38.0 – 46.0	**34.3** ms	F
D Plane Rotation			81.0 – 106.0	**56.6** degrees	F
Time at Max Rotation			72.0 – 82.0	**62.7** ms	F
Rotation Decay to Zero			147.0 – 174.0	**129.3** ms	F
Moment About Occipital Condyle			-80.0 – -52.9	**-54.5** Nm	F
Time at Max Moment			65.0 – 79.0	**47.1** ms	F
Moment Decay to Zero			120.0 – 148.0	**113.4** ms	F

Technician: **GS**
Company: **Denton ATD, Inc.**



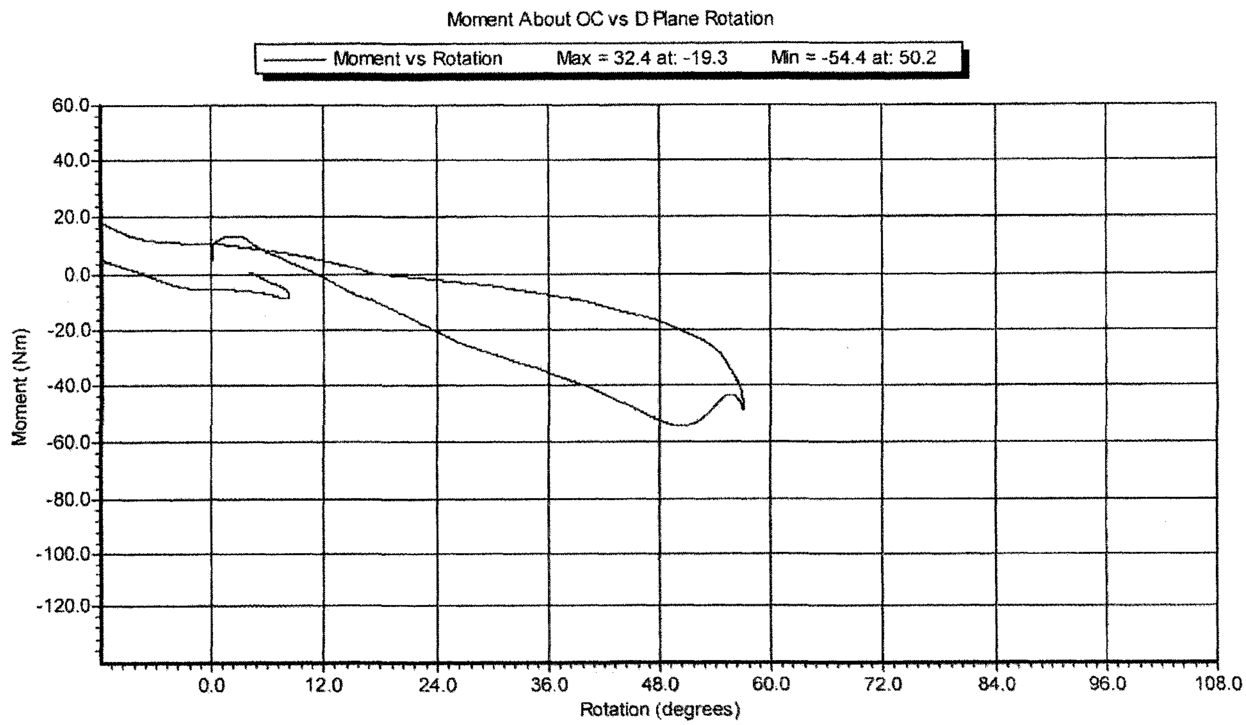
Test Name: **Neck Extension**
Dummy type: **Hybrid III 50th**
Test ID: **AC1821-4**
Test Number: **4**

Test Date: **8/28/2007**
Test time: **4:47:09 PM**

Comments: **Hard rubber. Modified velocity.**

Test Parameters			Test Specifications	Test results (avg)	
Temperature			20.6 – 22.2	**21.3** deg C	P
Humidity			10 – 70	**51** %RH	P
Velocity			5.94 – 6.19	**4.93** m/s	F
Pendulum Deceleration at	10	ms	17.2 – 21.2	**18.0** g	P
Pendulum Deceleration at	20	ms	14.0 – 19.0	**16.2** g	P
Pendulum Deceleration at	30	ms	11.0 – 16.0	**14.7** g	P
Max Pendulum Deceleration after	30	ms	0.0 – 22.0	**14.7** g	P
Decel Time to	5	g	38.0 – 46.0	**34.6** ms	F
D Plane Rotation			81.0 – 106.0	**57.1** degrees	F
Time at Max Rotation			72.0 – 82.0	**64.3** ms	F
Rotation Decay to Zero			147.0 – 174.0	**129.8** ms	F
Moment About Occipital Condyle			-80.0 – -52.9	**-54.4** Nm	P
Time at Max Moment			65.0 – 79.0	**47.4** ms	F
Moment Decay to Zero			120.0 – 148.0	**114.0** ms	F

Technician: **GS**
Company: **Denton ATD, Inc.**



Reference equipment:

Manufacturer	Model	Serial Number	Calibration Date
Endevco	**7231-750T**	**C16958**	**9/5/2006**
Denton ATD	**78051-342**	**7921-0325**	**1/12/2007**
Denton ATD	**78051-342**	**7921-0326**	**1/12/2007**
Denton	**1716A**	**344**	**12/12/2007**
Denton	**1716A**	**344**	**12/12/2007**

* for test AC1821-1 :

Manufacturer	Model	Serial Number	Calibration Date
Endevco	**7231-750T**	**C16958**	**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**

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**

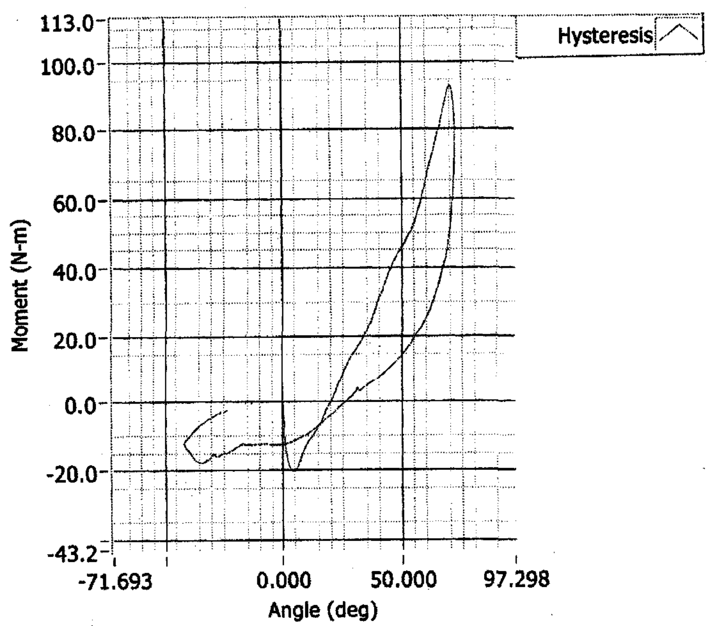
Test Date: **5/15/2007**
Test time: **10:51 AM**

Comments:

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**

Resultant Data - Hybrid III 50th Neck Neck Flexion



Test Name: **Neck Flexion**

Dummy type: **Hybrid III 50th**
Test ID: **144812**
Test Number: **4682-2**

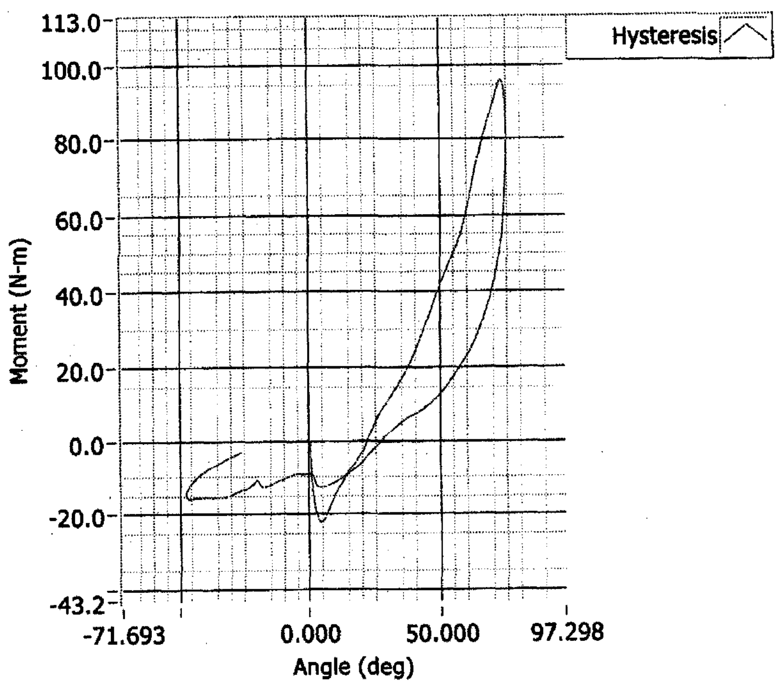
Test Date: **5/15/2007**
Test time: **11:32 AM**

Comments:

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**

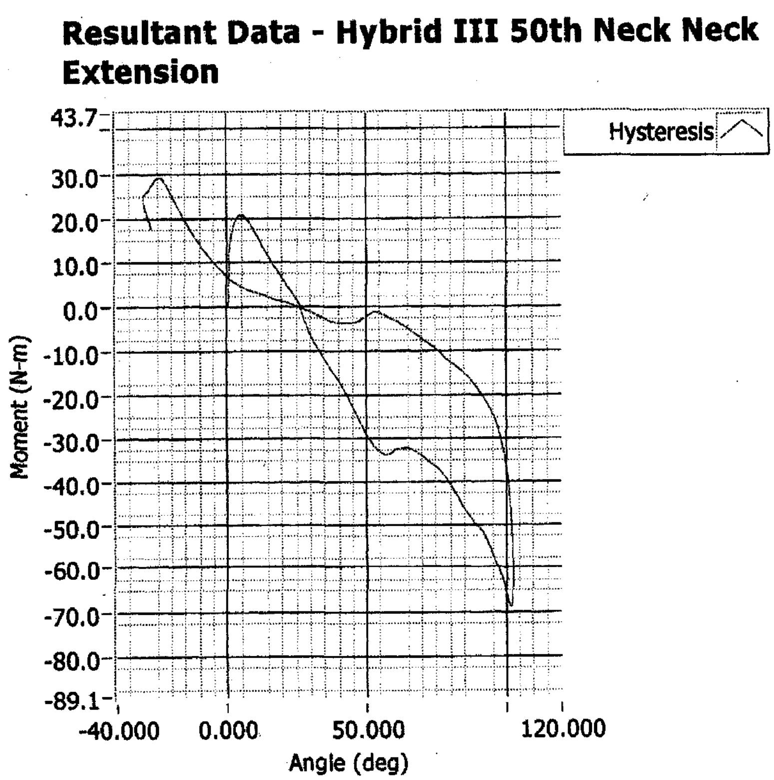
Dummy type: **Hybrid III 50th**
Test ID: **144813**
Test Number: **4682-1**

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

Comments:

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**



Test Name: **Neck Extension**

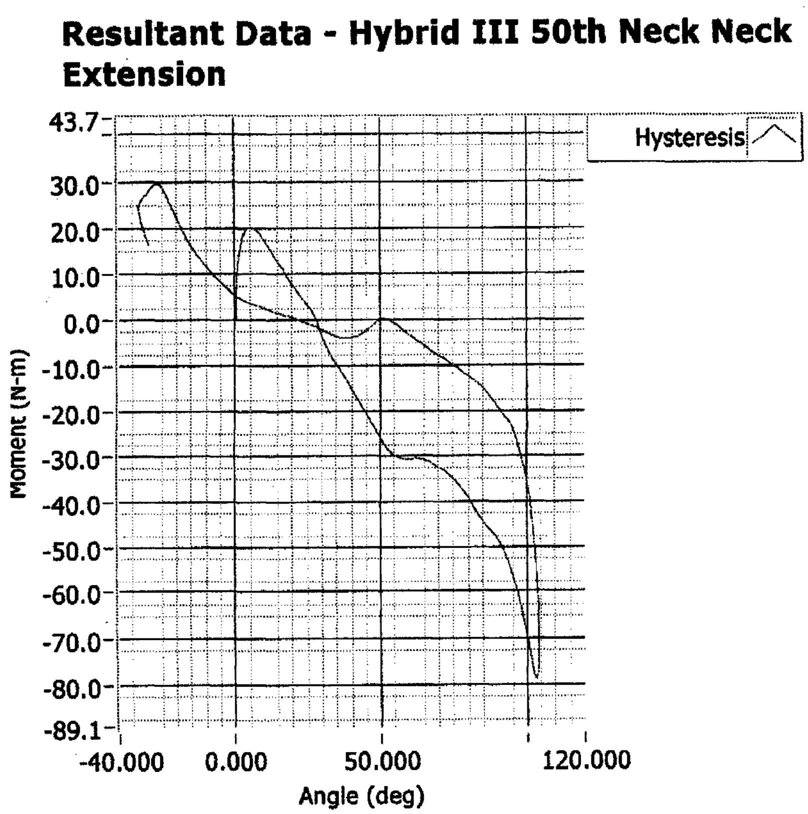
Dummy type: **Hybrid III 50th**
Test ID: **144831**
Test Number: **4682-2**

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

Comments:

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**



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