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Efficacy of Hard Hats in Attenuating Head Accelerations: A Combined Experimental and Computational Investigation

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Efficacy of Hard Hats in Attenuating Head Accelerations: A Combined Experimental and Computational Investigation By

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ABSTRACT

Struck-by accidents are a leading cause of traumatic brain injuries in the construction industry. While hard hats are the conventional means of industrial head protection, the current test standard to evaluate hard hat performance does not assess their ability to mitigate head accelerations from such impacts. To address this gap in knowledge, three investigations were pursued as part of this thesis. First, a variety of commercially available hard-hat designs – differentiated by shell design, number of suspension points, and suspension tightening system – were tested for their ability to attenuate accelerations during vertical impacts to the head. All hard-hats appreciably reduced head acceleration to the unprotected condition. However, neither the addition of extra suspension points nor variations in suspension tightening mechanism appreciably influenced performance. Second, the same hard hat designs were tested for their ability to attenuate head accelerations when subjected to impacts in two different head orientations – upright and forward-flexed by 30°. Impacts to the forward-flexed head resulted in the largest measured angular accelerations, and hard-hats were least effective at mitigating angular accelerations in this head position. Additionally, no correlations were observed between hard hat performance in an upright head orientation versus forward-flexed orientation. Results from this study provide insight into why impacts to a forward-flexed head are prevalent in epidemiological data, and also suggest that current hard-hat designs may not be optimized for impacts to a forward-flexed head. Lastly, a validated finite element model of a hard-hat was developed that accounted for more geometric detail than other models previously seen in literature. This validation process highlighted the importance of specific design features present in a hard-hat (such as headband attachments) and their influence in construction worker's safety against head injuries. Taken together, the work here represents a significant advance towards improving occupational safety in the construction sector.

Introduction

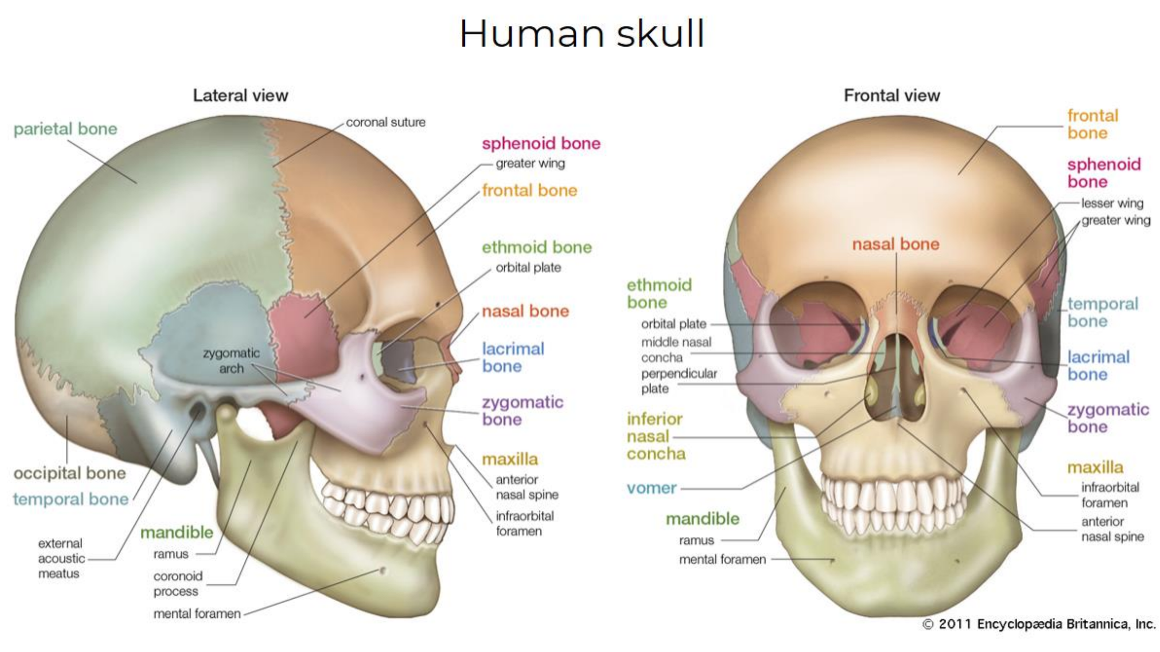
1.1 INFLUENCE OF TRAUMATIC BRAIN INJURY IN CONSTRUCTION

Traumatic brain injuries (TBI) cause a substantial number of deaths and lead to life-long disability for many individuals in the manual labor workforce[1-7]. Moreover, a recent study found that the greatest number of serious work-related injuries involving TBI were in the construction industry[7]. Workers in the construction industry face numerous hazards associated with the dynamic and changing nature of their work environment. Included among these hazards are falling objects or collapsing materials – categorized as “struck-by” incidents when a worker is impacted/injured by them. Struck-by accidents are the leading cause of non-fatal injury in the construction industry[8] and represent the second highest cause of traumatic brain injuries (TBIs) and concussions. Data from the Ontario Workplace Safety Insurance Board records indicate a specific prevalence of mild TBI of 49 per 10,000 in the construction industry[9]; with an estimated 10.3 million construction workers in the US in 2016[10], as many as 50,000 brain injuries are estimated to be occurring annually to US construction workers.

Many studies have found that TBI is one of the costliest injuries in terms of lost-time worker’s compensation claims[1-7]. In total, TBIs are linked to approximately $76 billion in annual costs in the United States alone[11]. In fact, a single severe TBI sustained by a 20-year-old may have lifetime medical costs in excess of $1.2 million[12]. The non-monetary consequences of severe TBI are also devastating. TBI can affect all aspects of an individual’s life, including interpersonal relationships, the ability to function at work, doing household tasks, driving, or participating in other daily activities[13]. Therefore, the burden of TBIs in construction worker’s life motivated this project and the information gathered should inform hard hat manufacturers and construction companies on equipment design influences on performance with the final goal of improving worker's safety.

1.2 HEAD ANATOMY

The human head is composed of the skull, scalp, meninges, brain, cranial nerves, sense organs, and parts of the digestive systems[14]. Each of these structures have crucial roles in the human body functions. However, the head parts which will be discussed in this article are the components that protect the brain against head accelerations, which is the main focus of this study. Therefore, it is crucial to provide background information on the head components first, which improves the understanding of this possible injury outcomes. The most important protective part of the head is the skull. It is a structure made of bones fused together, with different configurations regarding thickness and curvature[14-16]. Figure 1 shows the basic anatomy of the head bones and structure in the lateral and frontal views.



Lateral and anterior views of a human skull.

Encyclopædia Britannica, Inc.

Figure 1 - Human Skull Anatomy Frontal and Lateral View[17]

The skull is the primary defense mechanism against head injuries. However, there are other components seeking to prevent damage to the brain. For instance, the meninges are three layers with the goal of protecting and supporting the brain and the spinal cord, as well as providing a structure for the veins, arteries, as well as other body components[15,16]. The meninges layers are the dura mater, the arachnoid mater, and the pia mater, as shown in Figure 2.

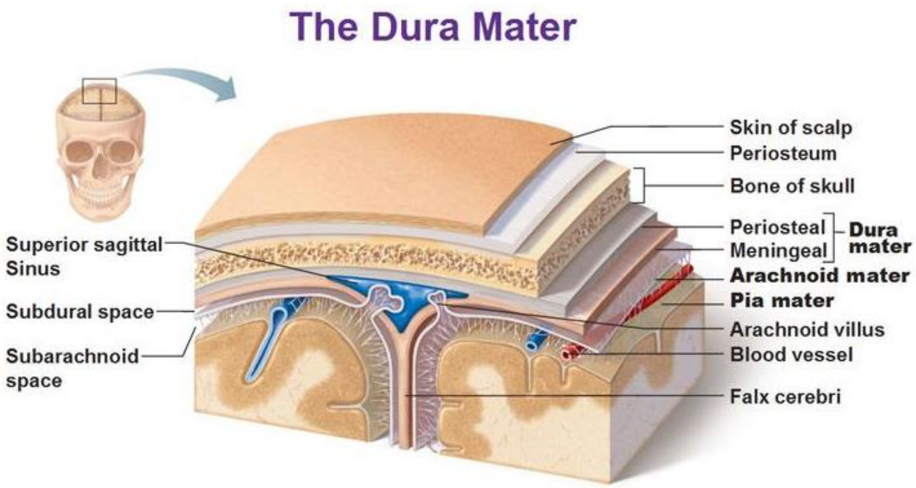


Figure 2 – Detailed Schematics of The Meninges[18]

The top section is the dura mater. It is the outer most layer of the meninges, and it is the toughest and thicker of all the three layers. The layer located below of the dura mater is called the arachnoid mater. In between these two layers there is a space which is called the subdural space, which is composed of a structure that resembles a web. The third and last layer of the meninges is the pia mater, also it is the closest layer to the brain. The arachnoid and pia maters are divided by the subarachnoid space, which is the region that contains the cerebral spine fluid (CSF). This fluid flows around the head providing support, protection, and nutritive functions to the brain and spinal cord[14]. In addition, there are bridging veins which run through all the meninges and provide vital support to the brain. These veins are an area of interest since they can tear and cause head injuries when someone suffers an impact or change in acceleration[16]. Thus, these body sections are the defense against brain and head injuries. However, the damage or failure of this components can also lead to medical conditions such as traumatic brain injuries.

1.3 PATHOLOGY OF TRAUMATIC BRAIN INJURY

Brain injuries can potentially involve damage to brain tissue, which can be challenging to examine with the current medical tests currently available. Thus, it is crucial to discuss the primary used method to identify the severity of an injury which is the Abbreviated Injury Scale or AIS. As it can be seen on Figure 3, codes were created to define the degree of injury[14,16].

INJURY DESCRIPTION	AIS-1998 SCORE	AIS-2005 SCORE
*Cerebellum: contusion, single or multiple: tiny; <1cm diameter	3	2
Cerebellum: hematoma (hemorrhage): epidural or extradural: NFS	4	3
*Cerebellum: hematoma (hemorrhage): epidural or extradural: tiny; <0.6cm thick	4	2
Cerebellum: hematoma (hemorrhage): intracerebellar including petechial and subcortical: NFS	4	3
*Cerebellum: hematoma (hemorrhage: intracerebellar including petechial and subcortical: tiny; <0.6cm diameter [includes radiographic "shearing" lesions]	4	2
Cerebellum: hematoma (hemorrhage): subdural: NFS	4	3
*Cerebellum: hematoma (hemorrhage): subdural: tiny; <0.6cm thick	4	2
Cerebellum: laceration [not from penetrating injury]: NFS	4	3
Cerebellum: laceration [not from penetrating injury]: ≤2cm length or depth	4	3
Cerebellum: subarachnoid hemorrhage	3	2
Cerebellum: subpial hemorrhage	3	2
*Cerebrum: contusion: single: tiny; <1cm diameter	3	2
*Cerebrum: contusion: multiple, on same side: tiny; each <1cm diameter	3	2
*Cerebrum: contusion: multiple, at least one on each side: tiny: each <1cm diameter	3	2
Cerebrum: diffuse axonal injury: NFS	5	4
Cerebrum: diffuse axonal injury: confined to white matter or basal ganglia	5	4
Cerebrum: hematoma (hemorrhage): NFS or "extra axial"	4	3
Cerebrum: hematoma (hemorrhage):epidural or extradural: NFS	4	3
*Cerebrum: hematoma (hemorrhage):epidural or extradural: tiny; <0.6cm thick	4	2
Cerebrum: hematoma (hemorrhage): intracerebral: NFS	4	3
* Cerebrum: hematoma (hemorrhage): intracerebral: tiny; single or multiple <1cm diameter	4	2
* Cerebrum: hematoma (hemorrhage): intracerebral: petechial hemorrhage(s) [includes radiographic "shearing" lesions]	4	2
* Cerebrum: hematoma (hemorrhage): intracerebral: petechial hemorrhage(s) [includes radiographic "shearing" lesions]; not associated with coma >6 hours	4	2
Cerebrum: hematoma (hemorrhage): intracerebral: small; ≤30cc or ≤15cc if age ≤ 10; 1-4cm diameter or ≤1cm if ≤age 10; subcortical hemorrhage: not associated with coma >6 hours	4	3
Cerebrum: hematoma (hemorrhage): subdural: NFS	4	3
Cerebrum: hematoma (hemorrhage): subdural: tiny; <0.6cm thick [includes tentorial (subdural) blood one or both sides]	4	3
Cerebrum: hematoma (hemorrhage): subdural: small; moderate; ; ≤50cc or ≤25cc if age ≤ 10; 0.6-1cm thick: bilateral [both sides 0.6-1.0cm thick]	5	4
Cerebrum: laceration [not from penetrating injury]: NFS	4	3
Cerebrum: laceration [not from penetrating injury]: ≤2cm length or depth	4	3
Cerebrum: intraventricular hemorrhage: NFS	4	2
Cerebrum: intraventricular hemorrhage: not associated with coma >6 hours	4	2
**Cerebrum: ischemic brain damage directly related to head trauma: associated with coma >6 hours	3	5
Cerebrum: subarachnoid hemorrhage: NFS	3	2
Cerebrum: subarachnoid hemorrhage: not associated with coma >6 hours	3	2
Cerebrum: subpial hemorrhage: NFS	3	2
Cerebrum: subpial hemorrhage: not associated with coma >6 hours	3	2
Cerebral Concussion: NFS	2	1

Figure 3 – The AIS 2005 Rating Relating the Ranking System to Head Injuries[19]

Within each section of the AIS, there are various types of injuries. However, only a few injuries will be discussed in this study. Generally, the most common injuries are hematomas/hemorrhage and diffuse axonal injury. TBI is the general description of the other three injuries mentioned, each one represents a more specific injury and they can range from mild to severe. For instance, the scope of TBIs in terms of AIS codes range from 1 to 4, which represents all the injuries mentioned earlier. The symptoms most commonly associated with TBIs are confusion, loss of consciousness and changes in personality. However, depending on the severity of these injuries, it might not be possible to confirm them through usual medical imaging procedures. Therefore, there are specific tests and criteria to diagnosing a mild TBIs. There were conventions to discuss the best approach and it was decided in 2001 to simply define a concussion as simple or complex based loss of consciousness, and few more medical criteria.

As mentioned earlier, TBIs can be further described as specific injuries. Furthermore, TBIs are divided in two categories: focal and diffuse. The concept of focal injuries is that the change in acceleration during an impact to the head can cause damage directly to the region of impact, these are also called coup contusions. The diffuse injuries on the other hand happens when the lesion is located opposite to the region of impact. Focal brain injuries have been more frequently related to changes in translational acceleration while, diffuse brain injuries are more often caused by changes in rotational acceleration[16]. For instance, Subdural Hematoma (SDH) is a type focal brain injury. This injury can be caused by several minor impacts or rapid changes in accelerations to the head, where one of the possible injury mechanics can be the rupture of the bridging veins. The blood from the bridging veins fills the subdural region with blood, increasing the brain intracranial pressure and lack of blood supply to the required regions. SDH's mortality rate has been above 30% in most researches[16]. Further analyzing the specific injuries, DAI is one of the highest scores in the AIS scale in the TBI spectrum. It is a diffuse type of brain injury related to damage to the white matter of the brain. DAIs are caused by the "disruption to the axons in the cerebral hemispheres and the subcortical white matter"[16]. In addition, it is frequently related to intracranial pressure problems, which further enhances the DAI severity.

1.4 INJURY CRITERIA

The injuries described in the previous section were only identified after impact and its causing factors were not fully explained. Therefore, researchers dating back to as early as 1953 were working towards developing injury criteria to provide quantitively metrics which could then be related to injury outcome. More specifically, these biomedical scientists sought to compare physical parameters, such as forces, accelerations, and stresses and correlate these magnitudes to brain injuries[16]. However, the creation of an injury criteria is a complex process. There are a few methods to achieve injury metrics. They could be studies with cadavers, or animals, or reconstruction of accidents with test dummies and each of these procedures have specific conditions in each the information is valid for. In addition to these limitations, it is important to note the biological heterogeneity of humans. Thus, most metrics are used reference, they are not precise thresholds. Thus, it is necessary to use this injury metrics to allow researchers to quantify impacts and analyze methods to improve the safety of head equipment.

The first criteria developed was the Wayne State Tolerance Curve (WSTC), the researchers collected data that suggested injury thresholds based on linear acceleration values along with impulse durations[16]. It was a crucial step towards creating a correlation between injury outcomes and physical metrics. Furthermore, the severity index (SI) was developed based on the WSTC, which complemented the WTSC with a weighing factor to the acceleration pulse entirely to fix some of its limitations[20]. The development of these the threshold standards were crucial for the researchers in the past and it provided fundamental information used in the current most accepted injury metric, Head Injury Criterion (HIC). It was initially published by the National Highway Traffic Safety Administration (NHTSA) and it proposes an empirical interpretation of the acceleration pulse to categorize risk of injury. The acceleration pulse was filtered and for the largest magnitude window 15 milliseconds or 36ms, depending on the HIC metric, was integrate and weighted according to Equation 1[16].There was extensive research on developing risk injury thresholds and a few values were stablished. For instance, a 50th percentile male that suffered head acceleration changes, which resulted in a HIC_{15} = 700, the likelihood of sustaining a fracture, AIS≥2, is around 31%. The same injury was linked to 48% risk on a HIC_{36} = 1000[16].

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

There are many other injury criteria and different ramifications of study for each one. For instance, this study only evaluates HIC, Peak Linear Acceleration (PLA), and Peak Angular Acceleration (PAA). PLA and PLA are commonly used because they are not interpretation of results, they are physical measurements gathered from tests. Thus, many studies have sought to identify limits for them according to previously established accidents and injury outcome. Ommaya et al. identified that a 50% probability of sustaining a cerebral concussion was linked to linear accelerations magnitude of 200g and rotational accelerations magnitudes of 1800 $\frac{rad}{s^2}$ for 20 milliseconds[16]. There are many other available risk injury limits established, however these were created based on specific accidents and population, which leads to the discrepancy shown in limits. It is impossible to define a single threshold for every person, but it is important that the researches developed those to provide a baseline and understanding of the physical effect causing these injuries. Furthermore, injury criteria are fundamental in development of new safety equipment such as hard hats. It can be to evaluate performance and to rank equipment, guiding designer to create better and improved safety gear.

1.5 ANTROPOMORPHIC TEST DEVICE HEADFORM – HYBRID III

In order to obtain the injury metrics previously discussed and compare it possible injury outcome and correlate it to the possible injuries also discussed earlier, it is necessary to conduct crash testing recreating likely construction accidents. It is complex and difficult to study cadavers and animal for example. Thus, the tests conducted in this project used the Hybrid III 50^{th} percentile male headform to collect the impacts acceleration pulses. This anthropomorphic test device (ATD) is currently the most common tool used in crash testing to understand impacts and improve safety. It is widely used in the automotive industry, and it is in fact incorporated in federal standards as well as a requirement in the National Highway Traffic Safety Administration (NHTSA)[21]. The model was created based on the 50^{th} male human characteristics, the intended goal of this device was to reproduce a human body with accurate dimensions and masses. Even though the materials used are not biofidelic, there are many researchers who sought to develop and test this ATD and showed this is the one the most reliable equipment commercially available. Thus, Hybrid III dummies enable scientists to study impacts without using animals or cadavers for instance.

1.6 CONSTRUCTION HEAD PROTECTION

Hardhats are the conventional means of head protection used in the construction industry, and thereby represent the primary protective mechanism against TBIs from struck-by events. Therefore, a succinct introduction to its components and method of attenuating head acceleration will be discussed. Generally, hard hats consist of three major elements: an extruded polymeric exterior shell, a fabric webbing strap and with an extruded polymeric headband. The shell's purpose is to prevent objects from penetrating the head and to disperse impact loads to the suspension system, which then attenuates and transfer the loads across the head. Plastic headbands are incorporated with the suspension systems, the headband serves to support and stabilize the hard hat on the head through the tightening mechanism that is the back portion of the headband. Moreover, the headband also contains a polymer attachment which connects the shell to the headband and serves as an anchor point for the straps. There are two classes of certification for hard hats – ANSI/ISEA Type 1 hard hats meet vertical impact and penetration requirements; while type 2 hard hats meet both vertical and lateral impact and penetration requirements. Type 1 hard hats are the most common on construction sites, and therefore are the hard hat models used in this study. Across manufacturers, subtle design differences exist in each of these features (e.g. shell material, shell geometry, straps material, number of suspension points, headband material, headband tightening mechanism, headband attachment). This project will evaluate how such design variations relate to impact attenuation. The amount of commercially available hard hats with different design features as well as the lack of information on how they impacted head accelerations attenuation is one of the crucial reasons for this study.

1.7 THESIS MOTIVATION

Hard hats are the conventional means of head protection used in the construction industry, and thereby represent the primary protective mechanism against TBIs from struck-by events. However, despite the adverse effects have in the lives of construction workers, the test standard used to evaluate hardhat performance (ANSI/ISEA Z89.1 2014) does not assess the ability of hard hats type I to mitigate head accelerations from such impacts. Rather, ANSI/ISEA Z89.1 2014 primarily evaluates the ability of hard hats to reduce neck loads, to prevent objects from penetrating through the shell and impact energy attenuation. However, studies have shown that TBIs have higher incidence rates the neck injuries[22]. Furthermore, neck loads are more frequently linked to fractures rather than brain injuries. Since there is no clear relationship between neck loads and risk of brain injury, workers may be purchasing/using helmets that offer sub-optimal protection against TBI.

The failure of quantifying the role of hardhats in preventing TBI is a critical knowledge gap in the field of construction safety. Therefore, it is necessary to evaluate the ability of existing hard-hat designs to protect against brain injuries through experimental tests. Moreover, it is fundamental to evaluate a variety of hardhat models (selected to represent a spectrum of commercially available designs: four- vs. six-point suspensions, different headband designs, tightening mechanisms and various shell designs). First, this project evaluated a group of six hard hats total which represented most of the model variation mentioned in the previous sections. The data collected is intended to quantify the difference in distinct hard hats through a parametric evaluation as well as determine the influence of the hard hat features in performance. The evaluation of relative performance is important to this field because the current certification tests do not evaluate changes in head acceleration, but also the tests are pass/fail. Thus, consumers are not informed which PPE provides further protection against brain injuries and chapter 2 identifies these questions. Furthermore, the Bureau of Labor Statistics (BLS) collected information and discovered that most workers that sustained head injuries were partially looking down[22]. Taking in consideration, that biomedical researchers imply the significant influence of PAA in TBIs, and the fact that there is no current standard testing for hard hats with this configuration or studies following these guidelines. Literature urgently needs data on the performance on hard with a head orientation that better represents accidents. Chapter 3 conveys information that addressed the different head postures scenarios. Finally, the last section of this study was the validation of a set of impacts conducted on chapter 2, through finite element simulations. The computerized model enables designers to understand the effect each design feature has on head acceleration attenuation and can be used to optimize designs to improve workers safety. Additionally, since the experimental tests are conducted with an ATD, brain stresses and strains are not evaluated. Thus, the validated hard hat model could be analyzed with a human head model and recreate impacts to more comprehensively understand the brain response to these impacts. The three upcoming chapter address fundamental questions that exist in the construction safety field and they provide insightful information on hard hats currently commercially available and it highlights a few questions that were previously not in the scope of researchers in this field.

Chapter 2 - The influence of Hard Hat Design Features on Head Acceleration Attenuation

2.1 INTRODUCTION

Work-related traumatic brain injuries (TBIs) are a leading cause of disability and death in the United States[23]. TBIs are one of the costliest injuries in terms of lost-time worker's compensation claims, linked to approximately $76 billion in annual costs in the United States alone[1-7,11]. Additionally, in 2017 the Bureau of Labor Statistics reported that the second highest rate of head injuries leading to days away from work (DAWF) was from the construction industry[24]. These head injuries are the most frequent outcome experienced in the construction industry as the result of impacts to the head, specifically from struck by incidents[22].

Hard hats are the standard form of personal protective equipment (PPE) used for head protection in the construction industry. Thus, hard hats represent the primary protective mechanism against TBIs, which are typically associated with abrupt changes in head acceleration[16,25-27] . Nevertheless, the current US test standard used to certify hard hats in terms of impact protection (ANSI/ISEA Z89.1-2014) does not assess the ability of hard hats to mitigate head accelerations from such impacts[28]. Rather, ANSI Z89.1 primarily assesses the ability of hard hats to reduce forces applied to the neck, and the effectiveness of shell protection against object penetration[28].

The main design components in hard hats which function to attenuate head acceleration are the suspension system and hard hat shell, and these features have generally been identified as essential to the effectiveness of head PPE attenuating head acceleration[29]. For example, in the 1990's Hulme, et al. tested a small sample of hard hats – each with different shell material, webbing cradle, and presence of foam liner[30] – and concluded that foam liners offered superior protection against skull fracture as compared to webbing cradles. Even so, this study failed to use hard hats that only had single design variations (thus isolating the role of individual design features); furthermore, current hard hat designs predominantly use webbing rather than foam liners, so the applicability of such findings to modern designs is unclear. More recently, Suderman et al. compared a single industrial hard hat to other types of helmets (e.g., a snow sport), depicting the different behavior of energy absorbing foam and plastic straps[31]. While the helmets were demonstrated to perform differently, little insight into the role of hard hat design features can be garnered.

Modern commercially available hard hats are available with many options of shell material and shape, suspension systems, and headband tightening mechanisms (e.g., the design of the strap attachment systems and the number of suspension points can be drastically different when sourcing alternative brands). However, to date, no information is available to indicate how such design variations affect impact attenuation in hard hats, particularly within the context of concussion/TBI. Based on data from sports testing in helmets, it is reasonable to assume that shell material, cushioning systems and the level of coupling achieved between the helmet and head all play an important role in determining helmet performance[32-35]. Therefore, a more comprehensive study comparing different suspension systems, including a different number of suspension system attachments, attachment design, and strap design with a consistent shell design could provide valuable insight into the role of these components.

Within this context, the main objective of this study was to examine the ability of varying hard hat designs, with particular focus on the influence of suspension systems and the number of attachments, in head acceleration attenuation. The performance was quantified using experimental testing with a Hybrid III 50th percentile head/neck from with outcomes including common injury criteria for brain injuries, including peak linear acceleration (PLA), peak angular acceleration (PAA) and Head Injury Criterion (HIC). Results from this study will provide an understanding of how certain design features influence the protective capacity of hard hats, which represents fundamental knowledge that can be applied in construction safety programs and can be further used as a foundation in the development of future designs.

2.2 MATERIALS AND METHODS

The general approach to this study involved conducting experimental impact tests with a variety of hard hat designs. The test fixture used for experimental drop tests consisted of a nine-foot-tall extruded aluminum frame, with a vertical linear rails system, as well as a horizontal system attached to the frame as seen in (Figure 4).

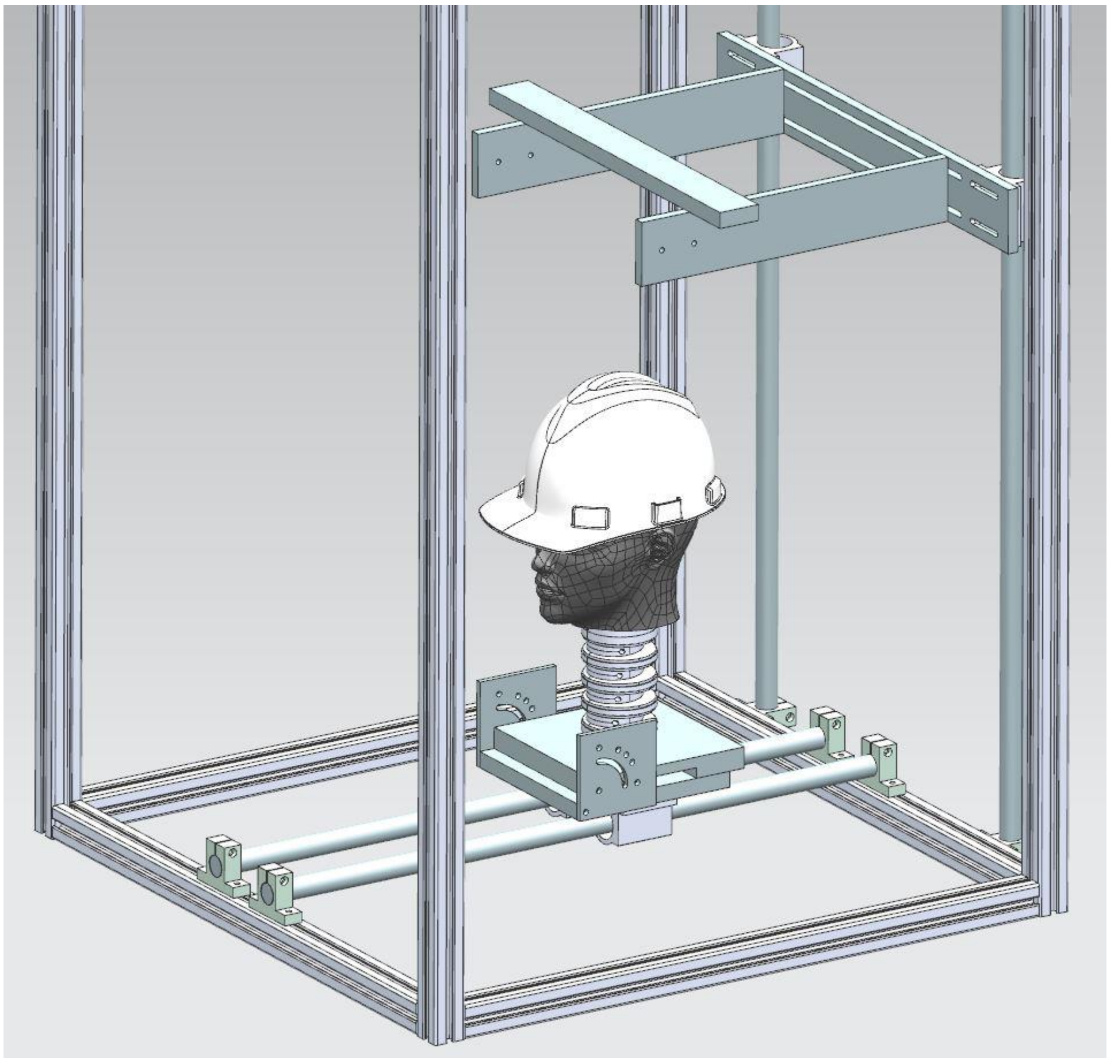


Figure 4 – Rendering of drop rig test fixture

The head and neck forms are a Hybrid III 50th percentile male anthropomorphic test dummy (ATD). The drop rails extend below the height of the head, thereby simulating a free-fall impact but providing consistent delivery of the impactor. Aluminum plates were bolted to linear bearings on the vertical rail to create a support where the object to be dropped could be rested and carefully positioned to impact the desired location, the center of gravity of the headform. The ATD was rigidly attached to a set of aluminum plates that were connected to the horizontal linear rail system, enabling the headform to be precisely positioned for each impact and to translate in an anterior-posterior direction after impact. Such a system is commonly used in impact testing[28,36].

2.2.1 Testing Parameters

Tests were conducted with three simulants for common construction materials: steel, wood, and lead shot. Steel and wood were selected since they are the first- and second-most common source of injury for head injuries in the BLS study of 1980[22] . Also, they fit the criteria of different stiffness properties, thereby providing insight into impact attenuation behavior of the hard hats in response to varying impactor stiffness. Lead shot was chosen as a third impactor to simulate loosely connected construction materials (e.g. a bag of bolts, nuts, nails, screws or small metal items in general), another common construction material. Each impactor material was generated at two different weights, 1.8 kilograms and 3.6 kilograms. The 3.6 kgs weight was chosen based on ANSI Z89.1.2014[28] (which specifies the impactor to be 3.6kgs) and 1.8kgs was chosen to investigate how hard hats would behave on lower energy impacts. All impact tests were conducted with a free-fall drop from a height of 1.83 meters This height was chosen based on average impact height reported in BLS 1980[22].

2.2.2 Equipment Description

Six hard hats design types were tested, selected from two different commercially available brands with three types of suspension system (Figure 5). The different designs represent the spectrum of commonly used strap styles and suspension systems in commercially available hard hats. In addition, the same tests were also conducted for the headform without head protection, which is the worst-case scenario and to serve as a baseline. Each hard hat was tested three times for each combination of impactor (steel, wood, and lead shot) and weight (1.8 kgs. and 3.6 kgs.) resulting in N = 126 total tests.

Figure 5 - MSA (top) & ERB (bottom) hard hats – 4-point pin-lock (left), 4-point ratchet (middle), and 6-ratchet (right) designs were tested from each manufacturer[37,38].

2.2.3 Data Analysis

Data was collected via a SLICE MICRO data acquisition system (DTS, Seal Beach, CA), which was set to sample acceleration data at 20 kHz with 4 kHz anti-alias filtering for three linear accelerometers and three angular rate sensors (DTS 6DX PRO 2K-18K, Seal Beach, CA). Data was post-processed through a custom MATLAB program (The MathWorks, Inc., Natick, MA) in accordance with SAE J211 (Instrumentation for Impact Tests) using a channel frequency class of 1000. Angular accelerations and other injury metrics were calculated in accordance with SAE

J1727 (Calculation Guidelines for Impact Testing). The injury criteria used in this analysis and the results from the post processed data are expressed in terms of peak linear acceleration (PLA), peak angular acceleration (PAA), impact duration (Δt), and Head Injury Criterion (HIC).

Analysis of variance (ANOVA) was used to quantify statistical significances for these impacts. Specifically, three four-way ANOVAs were conducted in SSPS (IBM, Armonk, NY) with independent factors including hard hat design, impactor material, impactor weight, and the dependent factor for each ANOVA was an injury criterion. Basic inspection of the hard hat models revealed that the design implementation of features varied across brands (e.g. MSA helmets have different suspension attachments to the shell from the ERB, also they have double webbing straps while ERB's use single, etc.). Therefore, hard hat design was implemented as a factor with six levels (one level corresponding to each model) rather than attempting to lump dissimilar brands by factors such as "4-point" and "6-point" suspension system. Thus, three ANOVAs were generated, with the addition of Tukey post hoc tests for the factors and their cross products, to investigate statistical differences across the variables. In addition, six two-way ANOVAs, and six two tailed, two sample unequal variance (heteroscedastic), t-tests were conducted to examine differences across specific impacts and specific comparison between certain hard hat models, as well as damaged and undamaged equipment.

2.3 RESULTS

All hard hat designs had superior performance as compared to the unprotected condition. For instance, the hard hats reduced PLA by 78-95%, PAA by 7-54%, and HIC by 79-99%, across all the tests conducted. Based on the substantial difference from any hard hats versus the unprotected condition, the four-way ANOVA was performed with the unprotected condition excluded.

The tests of between-subjects' effects for each variable: PLA, PAA, and HIC were conducted and listed on (Table 1). Significant differences existed across all three injury metrics based on impactor material (steel vs. wood vs. lead) and impactor weight (1.8 vs. 3.6 kg), with the largest values of each metric generally corresponding to the stiffer impactor materials and larger masses. The ANOVA indicated that variations in hard hat design did not influence PAA or HIC measures, although hard hat type (and interaction effects with weight and impactor) were significant in measures of PLA.

Table 1 - Tests of Between-Subjects Effects for each Injury Metric

Significance (p)	PLA	PAA	HIC
Impactor	**>0.001**	**>0.001**	**>0.001**
Weight	**>0.001**	**>0.001**	**>0.001**
Hard hat	**0.008**	0.141	0.417
Impactor * Weight	**>0.001**	**>0.001**	**>0.001**
Impactor * Hard hat	**0.032**	0.178	0.459
Weight * Hard hat	**0.028**	0.659	0.200
Impactor * Weight * Hard hat	**0.006**	0.091	0.314

Based on the outcome from the four-way overall ANOVA (i.e., that PLA was the only outcome variable with significant statistical differences), post hoc analysis on this outcome variable was warranted. Tukey HSD post hoc tests displayed two statistically significant differences: The ERB 6 ratchet and the MSA 4 ratchet performed significantly better in PLA reduction when compared to the ERB 4 ratchet. However, the individual ANOVAs for each test condition (e.g. 4 pounds steel bar) showed several statistical differences with no clear evidence of a best performer across all test conditions. In order to supplement more information on this subject, the average PLA, PAA, and HIC were displayed in (Table 2-4) with their standard deviation. In addition, the post hoc Tukey HSD statistical differences detected for the two-way ANOVAs are shown in the table through the superscripts, to highlight the differences identified for each impact condition.

Table 2 - Steel Impacts Average Results for each Injury Criteria & Two - Way ANOVA Statistical Differences

Weight	Hard hat	PLA (g)	PAA (rad/s^2)	HIC$_{15}$
4	MSA 4 Pin	23.60±4.04^{a}	894.44±23.93^{a}	6.88±0.10^{a}
	MSA 4 Ratchet	20.11±1.27	843.47±81.08^{b}	7.13±0.74^{b}
	MSA 6 Ratchet	18.32±3.10^{b}	929.71±11.22^{c}	5.42±0.95
	ERB 4 Pin	25.40±2.21$^{b, c}$	899.61±101.92^{d}	5.23±0.41
	ERB 4 Ratchet	23.02±1.91^{d}	1201.08±3.00$^{a, b, c, d, e}$	5.93±1.23
	ERB 6 Ratchet	15.99±1.24$^{a, c, d}$	838.29±18.10^{e}	4.60±0.86$^{a, b}$
8	MSA 4 Pin	27.91±7.41	1323.80±287.65	14.50±4.81
	MSA 4 Ratchet	29.57±1.70	1278.56±62.62	15.12±5.01
	MSA 6 Ratchet	32.85±3.27	1694.40±160.83	20.22±5.11
	ERB 4 Pin	30.27±0.44	1389.39±228.03	18.81±1.73
	ERB 4 Ratchet	33.67±2.16	1370.53±75.24	18.15±1.90
	ERB 6 Ratchet	28.92±1.13	1346.52±84.15	17.63±0.24

Note: Superscript characters (a, b, c, d, e) indicate significant differences within a given test condition ($p<0.05$)

Table 3 - Wood Impacts Average Results for each Injury Criteria & Two - Way ANOVA Statistical Differences

Weight	Hard hat	PLA (g)	PAA (rad/s^2)	HIC^{15}
4	MSA 4 Pin	20.86±1.60$^{a, b}$	739.51±64.40	6.10±0.74
	MSA 4 Ratchet	17.66±1.75$^{a, c}$	762.51±60.96	5.45±1.11
	MSA 6 Ratchet	22.00±1.57$^{c, d, e}$	915.03±174.25	7.57±0.56$^{a, b}$
	ERB 4 Pin	18.99±0.25^{d}	854.88±57.39	4.82±0.19^{a}
	ERB 4 Ratchet	19.97±0.65	778.37±80.89	5.84±1.89
	ERB 6 Ratchet	17.40±0.34$^{b, e}$	913.07±70.56	4.97±0.04^{b}
8	MSA 4 Pin	32.84±2.96	1313.76±23.99^{a}	19.92±1.58
	MSA 4 Ratchet	28.69±1.37$^{a, b}$	1395.24±186.58	15.90±1.18
	MSA 6 Ratchet	30.25±1.30^{c}	1313.88±214.75^{a}	20.65±2.91
	ERB 4 Pin	34.08±3.82	1508.92±72.12	17.77±0.74
	ERB 4 Ratchet	38.64±3.37$^{a, c}$	1333.56±114.47	19.06±0.65
	ERB 6 Ratchet	36.03±0.41^{b}	1365.31±109.61	19.54±0.52

Note: Superscript characters (a, b, c, d, e) indicate Significant differences within a given test condition ($p<0.05$)

Table 4 - Lead Impacts Average Results for each Injury Criteria & Two - Way ANOVA Statistical Differences

Weight	Hard hat	PLA (g)	PAA (rad/s^2)	HIC^{15}
4	MSA 4 Pin	11.16±0.16[a]	622.26±79.74	1.81±0.12[a]
	MSA 4 Ratchet	9.66±1.36	637.99±104.60	1.67±0.37
	MSA 6 Ratchet	7.992±0.19[a, b, c]	567.47±73.79	1.14±0.16[a, b, c]
	ERB 4 Pin	10.57±0.87[b]	719.12±110.41	1.93±0.21[b]
	ERB 4 Ratchet	11.07±0.85	612.93±56.75	1.72±0.30
	ERB 6 Ratchet	11.03±0.38[c]	656.49±30.99	2.03±0.12[c]
8	MSA 4 Pin	12.67±0.71	625.27±111.82	2.80±0.41[a]
	MSA 4 Ratchet	15.12±0.98	605.01±123.86	4.03±0.56[a, b]
	MSA 6 Ratchet	13.23±0.28	686.82±129.97	3.13±0.10
	ERB 4 Pin	12.76±1.02	630.68±63.13	2.91±0.33[b]
	ERB 4 Ratchet	13.22±0.54	696.93±115.03	3.01±0.36
	ERB 6 Ratchet	13.99±0.81	611.05±55.38	3.34±0.37

Note: Superscript characters (a, b, c, d, e) indicate Significant differences within a given test condition ($p<0.05$)

2.4 DISCUSSION

The focus of this study was to quantify the performance of different hard hat designs in response to vertical impacts from construction materials. While all hard hats offered significant protection compared to the unprotected condition, no clear trend was observed for any particular design feature to consistently offer better protection. For example, consistent improvements were not observed when comparing 4-point to 6-point strap systems or pin-lock to ratcheting tightening mechanisms. The lack of statistical differences between the simplest hard hat styles in this study (e.g. 4 suspension connection points and pin connection tightening) and the most complex designs (6 suspension connection points and ratchet tightening mechanism) indicates that such design features do not consistently improve performance. Therefore, the premise that hard hat performance would improve with increased design complexity was not confirmed by our results. The only notable exception to this conclusion was that the ERB 6 Ratchet performed significantly better than the ERB 4 Ratchet when comparing PLA results. This finding may reflect that the specific implementation of the 6-point design by ERB (and given all other details of their design, such as webbing material and shell design are held constant) is superior, but this effect should be treated with caution given that it was not observed across all injury metrics or other brands.

A thorough visual inspection was performed for each hard hat after testing. This inspection was performed because previous studies have found that damage to structures of protective headgear can serve to reduce head accelerations[39,40], offering a potential explanation for variations in outcome metrics observed in our study. Damage occurred in many cases, particularly with stiffer impactor materials such as steel. In almost all hard hats designs the location of damage was at the polymer connection points between the headband and the shell, as shown in (Figure 6 and 7), which would either plastically deform or fail altogether. To investigate this effect, t-tests were conducted to compare differences in acceleration metrics between undamaged and damaged hard hats. In four-pound impacts, hard hats that were damaged performed in average 23% better (in terms of in PLA) than undamaged hard hats for steel ($p = 0.007$) and 14% for wood, ($p = 0.003$). No differences were detected for the lead-shot impactor, which may be attributable to the compliance of that impactor, which would likely dissipate energy (through deformation of the shot) and prolong contact duration regardless of the presence of failure. Interestingly, no performance improvement was observed for any of the eight-pound impact conditions. We hypothesize the effect may not have been observable in the 8 lbs. impacts due to the significantly higher energy involved, which is above the threshold these attachments damage can dissipate. This provides motivation to further analyze these attachment designs and their energy dissipations principles.

Figure 6 - Undamaged ERB plastic attachments (left), damaged ERB plastic attachments (right)

Figure 7 - Undamaged MSA plastic attachments (left), damaged MSA plastic attachments (right)

A notable limitation to this study's findings is that the impacts simulated in experiment might differ from real life situations. Specifically, all experimental impacts were all conducted using pristine hard hats with the headform in a perfectly upright configuration and aligned through the center of gravity of the headform. Even though this is not the most common configuration of struck-by incidents reported from epidemiological studies, it does occur frequently[22,29] and it also matches the configuration used in ANSI Z89.1-2014[28]. We therefore caution that, while our results hold for vertical impacts with upright (non-flexed) head postures, future studies should evaluate the performance of hard hats to a variety of impact orientations, including the most common situations involving forward flexion of the head. Additionally, the hard hats tested in this study were all pristine. However, polymers can degrade over time and with environmental exposure. The performance of hard hats subjected to multiple impacts has also been studied by Wu et al, where it was identified that performance worsened upon repetitive impacts[41]. However, it is possible that design features such as extra webbing attachment points may make hard hats more resilient to such impact events, and differences between different designs could potentially be detectable in such a scenario.

An unexpected aspect of our results was that the injury metrics of PAA and HIC did not have any dependence on hard hat styles or design features, while PLA did have some dependence. Since HIC scores incorporate information about both the magnitude of acceleration and the duration over which they exist, this may indicate that helmets that reduce PLA achieve this behavior through concomitant increases in contact duration. Differences in PAA were likely not observed since the headform was vertically positioned and the impactor was lined up with the center of gravity of the ATD. However, researchers suggest that head injuries have been more frequently linked to PAA[42]. Thus, the perspective of impact location and the influence of PAA in traumatic brain injuries indicates the need for a clearer understanding regarding the protection provided by hard hats in terms of PAA, which yet again reinforces the need for additional types of impact testing to be conducted, including eccentric contact and with angled head postures.

Chapter 3 - Experimental Evaluation of Hard Hat Performance During Impacts to a Forward-Flexed Head

3.1 INTRODUCTION

Traumatic brain injuries (TBIs) are one of the most frequent injuries suffered by construction workers[6-8,22,30,43-47]. For example, Colantino et al. reported an incidence rate of mild TBIs of 49 cases for every 10000 full time employees[9]. While hard hats are the conventional means of industrial head protection, it is not clear that they are optimized to prevent TBI or that they are optimized for all impact configurations.

Towards this end, epidemiological data from the BLS indicates that looking partially down is the most common head position during injurious impacts. However, hard hats are currently certified by evaluating their performance on a headform in a neutral head position, which represents a person standing upright with no neck flexion[28]. As such, it is unclear that results from upright certification tests provide relevant data regarding a hardhat's performance when subjected to an impact with the head in a non-upright (e.g., forward flexed) position. From a biomechanical perspective, an additional concern about the forward-flexed head position is that it could increase the magnitude of rotational acceleration, and in fact, mild TBIs have been more frequently linked with changes in rotational acceleration rather than linear[16,42]. However, to our knowledge, only one study has tested hardhat performance for impacts to a non-upright head (which examined transversely applied loads[48]) and no data exists for a forward-flexed head posture. Consequently, understanding the performance of the personal protective equipment (PPE) used in construction with the headform in the forward-flexed orientation is crucial for understanding injury risk as well as the level of protection afforded by hardhats from these types of impacts.

Therefore, it is necessary to better understand the protective performance of hard hats with different head orientations. Within this context, we specifically identify two objectives. First, this study seeks to evaluate protective capabilities of hard hats for forward-flexed versus upright head postures (assessed by common biomechanical injury metrics, including peak linear acceleration (PLA), peak angular acceleration (PAA), and head injury criterion (HIC)). Second, this study seeks to evaluate whether the impact performance of hard hats tested in an upright head posture is predictive of performance in forward flexed impacts. Such data could be beneficial for understanding occupational injury risk and for beginning the process of optimizing hard hats for a variety of the most common impact configurations.

3.2 MATERIALS AND METHODS

The experimental test fixture used consisted of an extruded aluminum frame, with two linear rail systems, as seen in Figure 8. A set of aluminum plates were connected to the vertical rails via low-friction pillow block bearings to bear the impactor and allow the object to be positioned in line with the center of gravity of the headform. The anthropomorphic test device (ATD) was mounted at the bottom of the frame, where it was rigidly attached to another set of aluminum plates. This group of plates had the ability to be fixed at any angle ranging from 0° to 90°. These plates were attached to a horizontal linear rail system via low-friction pillow blocks, enabling the translation of the ATD in the anterior-posterior direction after the impacts. Similar testing format and structure has used and reported elsewhere in literature[28,36].

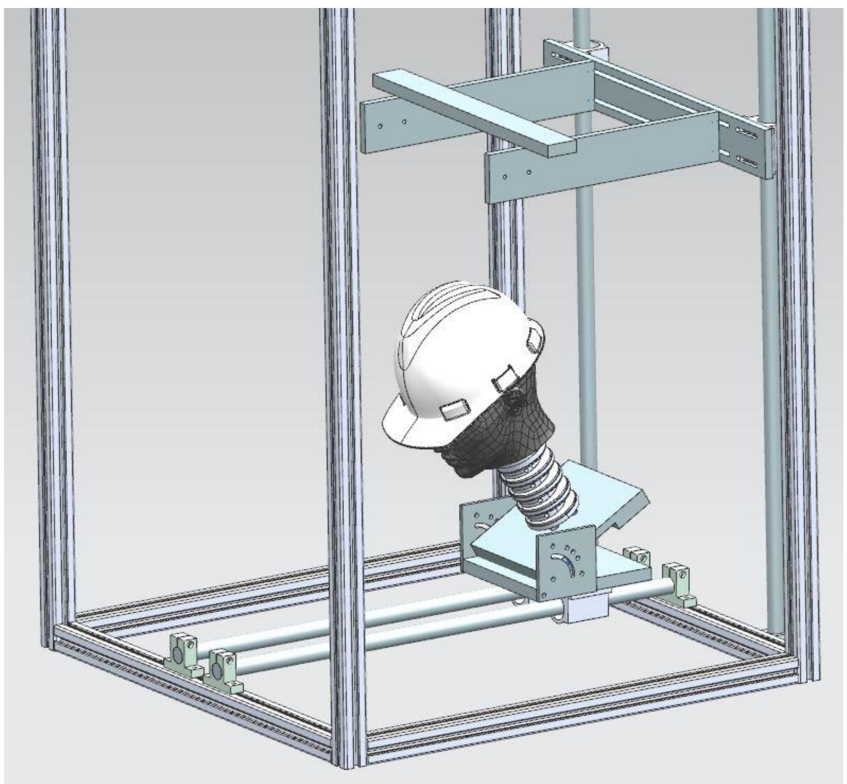


Figure 8 - Rendering of the drop rig fixture used in experiments, showing the head in a forward-flexed position prior to impact.

3.2.1 Testing Parameters

Three impactor types were used to conduct the tests: steel, wood, lead shot. Steel and wood objects are the first and second most common source of injury for head injuries, respectively[22]. The third type of impactor chosen was lead shot, as it can simulate collections of loose metallic items, such as nails, screws, or nuts. All impactors were fabricated to have a weight of 3.6 kg, which was chosen since it matches the standard used in current hard hat testing by ANSI Z89.1 2014[28]. The impactors were positioned 1.83 m above the ATD headform, consistent with the range of the most frequent impact heights according to BLS[22].

The base plate of the test fixture can be locked in a variety of angled positions to simulate forward flexion of the head. Two head position configurations were selected for this study, neutral (also called upright) and forward flexed. The angle of flexion used for the test configuration was 30°, chosen based on data collected for the work tasks and associated body postures in the construction industry[22,49].

3.2.2 Equipment Description

The hard hats selected for this experiment were six models from two different brands with three distinct suspension systems, as seen in Figure 9. The different designs represent the most common commercially available construction head PPE in rtelation to styles of strap and suspension systems. Each hard hat was tested three times for each combination of impactor (steel, wood, and lead shot) and head position (0° and 30°). Additionally, tests without helmets were also conducted for each test condition, resulting in a total of N=126 tests.

Figure 9 - MSA (top) & ERB (bottom) hard hats: 4-point pin-lock (left), 4-point ratchet (middle), and 6-ratchet (right) designs were tested from each manufacturer[37,38]

3.2.3 Data Analysis

Data was collected via a data acquisition system (Slice Micro, DTS, Seal Beach, CA) mounted at the CG of the headform, which was set to sample acceleration data at 20 kHz with 4 kHz anti-alias filtering for three linear accelerometers and three angular rate sensors (DTS 6DX PRO 2K-18K, Seal Beach, CA). The data was then post-processed through a custom MATLAB program (The MathWorks, Inc., Natick, MA) in accordance with SAE J211 (Instrumentation for Impact Tests) using a channel frequency class of 1000. Angular accelerations and other related injury metrics were calculated in accordance with SAE J1727 (Calculation Guidelines for Impact Testing). The injury criteria used in this analysis and the results from the post-processed data are expressed in terms of peak linear acceleration (PLA), peak angular acceleration (PAA), and Head Injury Criterion (HIC).

Analysis of variance (ANOVA) was performed using SSPS (IBM, Armonk, NY) to assess factors influencing magnitudes of measured injury criteria, thereby providing insight to the first objective of this study – to evaluate the basic biomechanical response for upright versus forward flexed impacts. Two sets of this statistical tool were generated. First, three four-way ANOVAs were conducted, where the independent factors were: hard hat design, impactor material, head orientation, and the dependent factor for each ANOVA was an injury metric (e.g. PLA, PAA, or HIC). The second set were three three-way ANOVAs, where orientation was not a factor, therefore exclusively comparing the angled impacts among themselves. Tukey HSD post hoc tests were performed on any ANOVA where overall significance was detected in order to investigate statistical differences across each factor and their interaction.

To investigate whether the results form upright/neutral testing were predictive of forward-flexed results, a linear Pearson Correlation analysis was performed. Related, nine plots scatter plots and a table, separated by impact condition and injury metric (e.g. 8 pounds steel bar – PLA 0° X PLA 30°), were created to depict the relationship between injury outcome metrics measured in vertical versus forward flexed impacts.

3.3 RESULTS

For impacts to an unprotected headform, the upright head position tended to result in larger values for PLA and HIC (reaching significant statistical differences, $p < 0.001$, for PLA and for HIC for any impactor) whereas the forward flexed position produced significantly greater values of PAA ($p < 0.001$). These trends did not remain the same when the hard hats were added (Table 5). Specifically, most linear outcome metrics (PLA and HIC) were significantly higher when the head was forward-flexed ($p < 0.05$). Angular metrics consistently remained higher for the forward flexed position even with the use of the hard hats ($p < 0.05$) – the highest values measured in upright head testing were less than the lowest values measured during forward-flexed testing. The hard hats were not effective at reducing PAA across all test conditions with the forward-flexed head posture. For example, for the lead shot impactor, the best performing hard hat reduced PAA by only 25% and the worst performing hard hat was equivalent to the unprotected condition ($p > 0.90$).

Table 5 - Testing outcomes for upright versus forward-flexed head positions, including unprotected and range of hard hat outcomes (range was selected as the performer with the smallest mean and largest mean, with ± two standard deviations reported; the selection

		Upright		Forward-Flexed	
		Unprotected	Hardhat Range	Unprotected	Hardhat Range
Steel	PLA (g's)	331.0 ± 3.4	27.9 ± 7.4 to 33.7 ± 2.2	222.6 ± 11.1	31.7 ± 1.5 to 41.55 ± 4.79
	HIC	720.8 ± 19.2	14.5 ± 4.8 to 20.22 ± 5.11	290.6 ± 6.6	16.2 ± 1.2 to 30.3 ± 2.0
	PAA (rad/s^2)	2751 ± 836	1279 ± 63 to 1694 ± 161	4225 ± 153	2012 ± 98 to 3117 ± 728
Wood	PLA (g's)	247.1 ± 6.9	28.7 ± 1.4 to 38.6 ± 3.4	221.0 ± 6.2	28.7 ± 0.9 to 39.6 ± 0.7
	HIC	472.1 ± 11.5	15.9 ± 1.2 to 20.7 ± 2.9	417.6 ± 16.6	16.2 ± 1.2 to 22.6 ± 0.9
	PAA (rad/s^2)	2608 ± 576	1313 ± 24 to 1509 ± 72	8529 ± 760	1608 ± 106 to 2393 ± 105
Lead Shot	PLA (g's)	74.5 ± 11.3	12.7 ± 0.7 to 15.1 ± 1.0	67.3 ± 14.6	13.0 ± 1.5 to 16.5 ± 0.3
	HIC	25.4 ± 8.2	2.8 ± 0.4 to 4.0 ± 0.6	17.5 ± 8.9	3.1 ± 0.2 to 5.0 ± 0.7
	PAA (rad/s^2)	885 ± 148	605 ± 124 to 697 ± 115	1431 ± 152	1071 ± 104 to 1445 ± 150

After adjusting for all factors, the ANOVA indicated that PLA, PAA, and HIC were significantly higher for impacts in a forward-flexed posture as compared to upright (based on the independent factor, orientation, where the comparisons for all injury criteria was significantly different ($p<0.001$)). This was further confirmed via orientation pairwise comparisons for all injury metrics were ($p<0.001$). Tables 6, 7 and 8 display all the test results obtained in average for each hard hat separated by impactor type.

Table 6 - Steel Impactor Test Results

PPE	Head Orientation	PLA (g)	PAA (rad/s^2)	HIC
MSA 4 Pin	F	41.4 ± 3.11	3000.64 ± 846.08	30.31 ± 2.00
	U	27.91 ± 7.41	1323.80 ± 287.65	14.50 ± 4.81
MSA 4 Ratchet	F	41.55 ± 4.79	3117.43 ± 728.27	26.91 ± 2.78
	U	29.57 ± 1.70	1278.56 ± 62.62	15.12 ± 5.01
MSA 6 Ratchet	F	38.60 ± 1.90	2656.50 ± 217.97	24.76 ± 1.19
	U	32.85 ± 3.27	1694.40 ± 160.83	20.22 ± 5.11
ERB 4 Pin	F	33.89 ± 2.74	2158.42 ± 186.12	16.24 ± 1.18
	U	30.27 ± 0.44	1389.39 ± 228.03	18.81 ± 1.73
ERB 4 Ratchet	F	31.68 ± 1.47	2011.58 ± 98.28	17.91 ± 0.39
	U	33.67 ± 2.16	1370.53 ± 75.24	18.15 ± 1.90
ERB 6 Ratchet	F	32.77 ± 2.35	2094.96 ± 154.66	19.65 ± 0.89
	U	28.92 ± 1.13	1346.52 ± 84.15	17.63 ± 0.24
Unprotected	F	237.78 ± 21.11	18748.13 ± 13579.62	334.09 ± 38.45
	U	331.02 ± 3.41	2750.80 ± 835.58	720.84 ± 19.15

Note: F – Forward Flexed, U – Upright

Table 7 - Wood Impactor Test Results

PPE	Head Orientation	PLA (g)	PAA (rad/s^2)	HIC
MSA 4 Pin	F	39.56 ± 0.66	2393.28 ± 104.62	22.63 ± 0.85
	U	32.84 ± 2.96	1313.76 ± 23.99	19.92 ± 1.58
MSA 4 Ratchet	F	32.89 ± 3.07	1608.13 ± 106.33	19.32 ± 0.64
	U	28.69 ± 1.37	1395.24 ± 186.58	15.90 ± 1.18
MSA 6 Ratchet	F	28.70 ± 0.86	1831.02 ± 105.98	16.99 ± 1.35
	U	30.25 ± 1.30	1313.88 ± 214.75	20.65 ± 2.91
ERB 4 Pin	F	32.18 ± 0.79	1839.90 ± 198.80	13.87 ± 1.38
	U	34.08 ± 3.82	1508.92 ± 72.12	17.77 ± 0.74
ERB 4 Ratchet	F	32.91 ± 0.42	2191.96 ± 453.61	15.89 ± 0.67
	U	38.64 ± 3.37	1333.56 ± 111.47	19.06 ± 0.65
ERB 6 Ratchet	F	29.33 ± 0.19	2384.12 ± 169.64	15.55 ± 1.08
	U	36.03 ± 0.41	1365.31 ± 109.61	19.54 ± 0.52
Unprotected	F	231.19 ± 10.05	22840.92 ± 20841.38	461.38 ± 58.17
	U	247.14 ± 6.91	2608.22 ± 575.71	472.09 ± 11.53

Note: F – Forward Flexed, U – Upright

Table 8 - Lead Shot Impactor Test Results

PPE	Head Orientation	PLA (g)	PAA (rad/s^2)	HIC
MSA 4 Pin	F	16.53 ± 0.32	1444.95 ± 149.65	5.01 ± 0.72
	U	12.67 ± 0.71	625.27 ± 111.82	2.80 ± 0.41
MSA 4 Ratchet	F	14.81 ± 1.02	1315.19 ± 25.47	4.59 ± 0.53
	U	15.12 ± 0.98	605.01 ± 123.86	4.03 ± 0.56
MSA 6 Ratchet	F	14.64 ± 0.78	1346.89 ± 68.10	4.69 ± 0.34
	U	13.23 ± 0.28	686.62 ± 129.97	3.13 ± 0.10
ERB 4 Pin	F	15.47 ± 0.09	1191.01 ± 143.53	4.34 ± 0.14
	U	12.76 ± 1.02	630.68 ± 63.13	2.91 ± 0.33
ERB 4 Ratchet	F	13.03 ± 1.54	1357.11 ± 97.24	3.14 ± 0.23
	U	13.22 ± 0.54	696.93 ± 115.03	3.01 ± 0.36
ERB 6 Ratchet	F	14.20 ± 0.70	1070.81 ± 104.1	3.91 ± 0.27
	U	13.99 ± 0.81	611.05 ± 55.38	3.34 ± 0.37
Unprotected	F	67.31 ± 14.64	1430.82 ± 152.4	17.50 ± 8.93
	U	74.49 ± 11.34	884.70 ± 147.92	25.36 ± 8.24

Note: F – Forward Flexed, U – Upright

To investigate whether a correlation existed between injury metrics measured for upright vs. forward-flexed impacts, nine separate linear regressions and correlation analyses were performed. Correlation coefficients are provided in Table 9. None of the correlations were statistically significant ($p>0.16$). Moreover, many of the relationships with the highest correlation coefficients had negative correlations, indicating that hard hats that performed well in upright tests were often poor performers in forward-flexed tests (and vice versa). A sample scatter plot depicting such a correlation is provided in Figure 10, which shows that PLA measured from upright impacts has a negative correlation with PLA measured for the same hard hat in forward flexed impacts.

Table 9 - Correlation Coefficients for the Different Orientation Test Results. Note that none of the relationships are statistically significant.

TEST CONDITION	INJURY METRIC	CORRELATION COEFFICIENT
STEEL	PLA	-0.425
	PAA	-0.075
	HIC	-0.644
WOOD	PLA	-0.003
	PAA	-0.444
	HIC	0.053
LEAD SHOT	PLA	-0.308
	PAA	0.402
	HIC	0.059

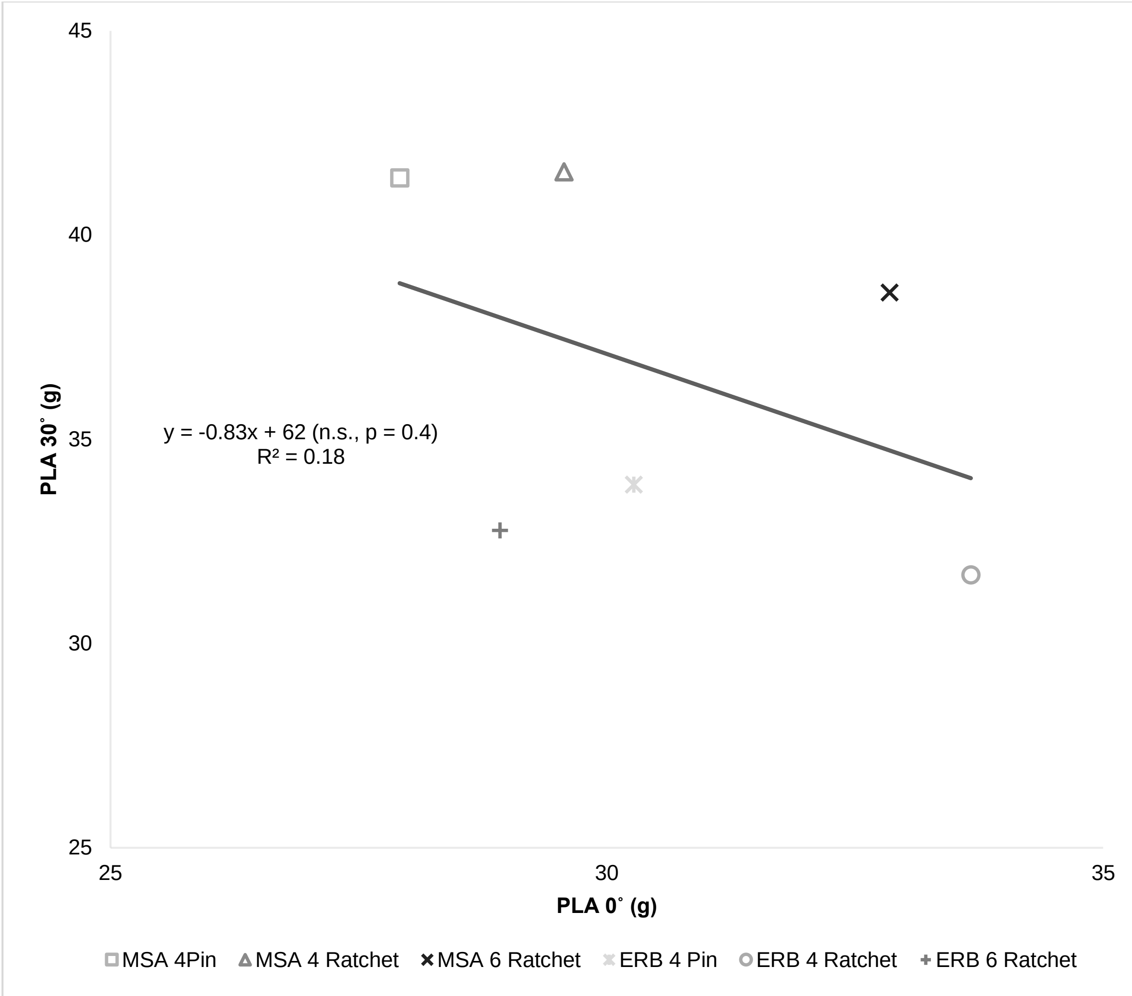


Figure 10 - Correlation between PLA measurements for forward-flexed head posture versus upright head posture for a steel impactor

Additionally, it was noted that different helmet models performed differently with the headform positioned forward flexed. The p values for all injury metrics for the cross product, Orientation X hard hat, were less than 0.002. Tukey HSD post hoc tests on the three-way ANOVAs were performed to provide more insight on the differences between hard hat models. Table 10 displays the averages for the angled impacts, and the standard deviation for all the tests.

Table 10 - Hard Hat Models Average Results for each Injury Criteria for the Forward Flexed Orientation with Significant Statistical Differences.

Hard hat	PLA (g)	PAA (rad/s^2)	HIC^{15}
MSA 4 Pin	32.5 ± 12.2$^{a, b, c, d}$	2280 ± 861^{a}	19.3 ± 11.3$^{a, b, c, d, e}$
MSA 4 Ratchet	29.8 ± 12.3$^{e, f}$	2014 ± 951	16.9 ± 10.0$^{d, f, g, h}$
MSA 6 Ratchet	27.3 ± 10.5^{d}	1945 ± 594	15.5 ± 8.8$^{e, i, j, k}$
ERB 4 Pin	27.2 ± 9.3^{a}	1730 ± 468^{a}	11.4 ± 5.5$^{a, f, i}$
ERB 4 Ratchet	25.9 ± 9.7$^{b, e}$	1854 ± 479	12.3 ± 7.0$^{b, g, j}$
ERB 6 Ratchet	25.4 ± 8.7$^{c, f}$	1850 ± 617	13.0 ± 7.1$^{c, h, k}$

Note: Superscript characters (a, b, c, d, e, f, g, h, I, j, k) indicate significant differences within a given test condition ($p<0.05$)

3.4 DISCUSSION

Fundamental epidemiological data supports the fact that the most common head orientation at the time of head injuries is looking partially down[22]. However, certification standards have not been developed according to the forward-flexed head position, and it is unclear if hard-hat manufacturers design their products for such a situation. The data set from this study is completely novel (we are aware of no studies that have ever tested hard hats on a headform with neck flexion) and its findings highlight the importance of head orientation in hard-hat performance. Comparisons between the two head orientations showed that the forward-flexed posture produced higher values for all injury metrics, with a modest increase for linear metrics and the greatest increase being seen for peak angular acceleration (7.8% for PLA, 10.9% for HIC, and 41.5% for PAA). Although hard hats attenuated accelerations reasonably well for a forward-flexed head position (compared to the unprotected values), they were less effective at doing so than when the headform was upright and were least effective at attenuating angular accelerations. Moreover, for forward-flexed postures subjected to lead and steel impacts, the measured values of PAA still approached some tolerance thresholds[16] even when equipped with a hard hat. Several biomechanical studies have suggested that rotational accelerations are more predictive of TBI than linear[16,25,42]. The results from this study provide insight into why impacts to a forward-flexed head are prevalent in epidemiological data, but also suggest that current hard-hat designs may not be optimized for attenuating angular accelerations during impacts to a forward-flexed head.

A fundamental finding from this study is that the performance of individual hard hat models from upright testing conditions was not correlated to the performance of the same models in a forward-flexed condition. Moreover, many of the correlations that were developed had negative relationships (although not significant), indicating that some hard hats that performed well in upright testing often performed poorly in forward-flexed conditions (or vice versa). This suggests that the selection of a hard hat based on results from upright test conditions may in fact result in reduced protection during the most common and injurious accident scenarios involving flexed postures. Future studies should seek to identify mechanistic sources of these negative correlations (i.e., design features that cause helmets to perform well in either vertical or forward-flexed conditions) such that new generations of hard hats can be developed that respond well to impacts under a variety of head postures.

While it was not a specific goal of this study to compare brands or design features relative to protective performance, the inconsistency in performance relative to head orientation and the difference in relative performance when comparing only the forward flexed results for the two brands of hard hats, suggests a design factor is influencing head acceleration attenuation. For example, all ERB models performed significantly better than all the MSA models in terms of HIC ($p<0.05$). Also, the ERB models generally reduced PLA and PAA more effectively than the MSA models (although the differences were not statistically significant). Even though models were selected from each manufacturer based on their similarity in features (a 4-point pin-lock and ratcheting model as well as a 6-point ratcheting model were used from each), this difference suggests that some other design feature on the ERB models is causing this improvement.

Analyzing the high-speed video footage of impacts, there was evidence that the ERB models displaced more and tended to decouple from the headform, which provides a potential mechanistic explanation for the greater attenuation of accelerations experienced during testing. This physical phenomenon can be observed in the still frames shown in Figure 11, where the change in position of the ERB suspension system is evident (the brim has tilted to the level of the nose, and the backstrap of the headband has fallen beneath the occiput) while the MSA model has remained in approximately the same position as before impact. As seen in Figure 2, the headband, ratchet contact materials, and the connecting pin from the headband to the back section of the tightening system of the ERB models are substantially different as compared the MSA models. Qualitatively, the ERB headband and tightening mechanism materials seemed to be made from materials with less friction than the MSA models, potentially enabling the hard hat to move independently of (i.e., decoupled from) the headform. Thus, as previously mentioned in literature, the decoupling effect is hypothesized to attenuate head acceleration[40,50,51].

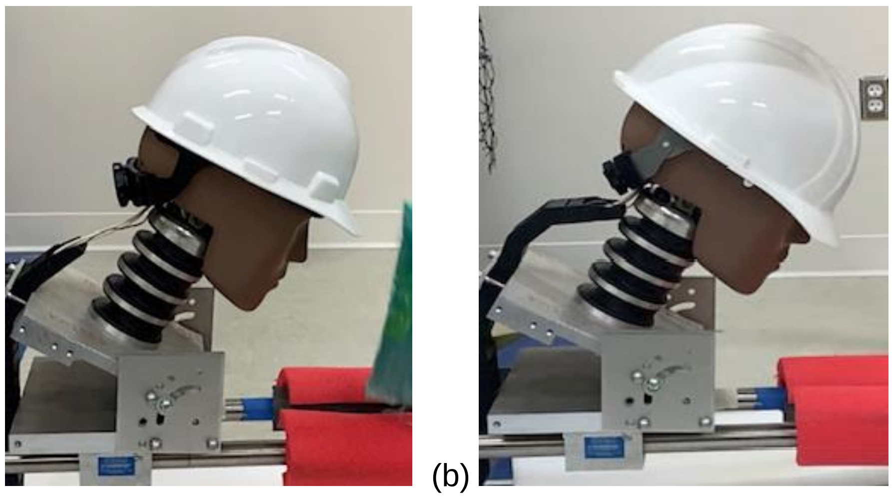

(a) (b)

Figure 11 - Still frames taken from high-speed video showing (a) an MSA hard hat and (b) an ERB hard hat immediately post impact.

A limitation to this study is that only six hard hat models were tested. While this is relatively large in comparison to many prior studies that have experimentally tested hard hats, there are nevertheless numerous models of hard hats commercially available. It is not possible to test all hard hat designs currently on the market. The hard hats that were selected were chosen since they are amongst the best-selling models and their features are generally representative of most commercial models. The general features that were identified as being potentially important performance in this study can also be seen or improved by a majority of the other models. However, in that we observed substantial performance differences across brands that were potentially attributable to relatively subtle design differences, it is possible that other hard hat models/brands that were not evaluated as part of this study could exhibit different performance behavior/trends than were observed here.

Taken together, this study provides novel data regarding the performance of hard hats in a different head orientation from the most commonly tested upright posture, and the findings provide compelling evidence that construction PPE should be further analyzed for a variety of head postures to improve worker safety. Notably, hard hats did not attenuate accelerations as effectively when the head was in a forward-flexed posture. Moreover, there was no correlation between hard hat performance in upright versus forward-flexed postures, indicating that models that were most effective in upright tests generally did not perform well in forward-flexed tests (and vice versa). Moving forward, design features should be optimized to help design hard hats that effectively attenuate accelerations for a variety of head postures.

Chapter 4 - Validation of Hard Hat Finite Element Models Using Experimental Data

4.1 INTRODUCTION

Traumatic brain injuries (TBIs) have immense consequences from a medical, financial, and emotional perspective[1-6, 52] . According to a Workers' Compensation Board of British Columbia (WCB) report from 2018, the construction sector has the highest serious injury rates among all sectors: 0.82 per 100 person-years of employment. Within the same injury group, struck-by (the term for being hit by a falling or moving object) was the highest incidence type among all claims, and concussion was the third most frequent injury type[53]. In many industries, but mainly in the construction sector, hard hats are the principal method of protecting workers against head injuries.

The most common literature regarding impact analysis of construction head PPE involves experimental testing evaluation of the equipment's' performance. For instance, Suderman et al. gathered information on the performance of a single hard hat design when impacted by different objects at different height[31] . Wu et al. investigated the performance and degradation of hard hats subjected to repeated impacts[41]. However, these types of experimental evaluations are subject to certain drawbacks, such as the inability to provide insight into the stresses/strains experienced in the brain and the inability to parametrically vary hard hat features.

Finite element (FE) models overcome these limitations by providing the ability to easily modify hard hat geometry and material properties as well as the ability to simulate impacts to human head models (including brain tissues and other relevant structures) to better understand injury risk. However, a fundamental question remains unanswered regarding the level of detail required in these FE models such that they can be validated against experiment. For example, in 2013, Long et al. developed an FE hard hat model that only incorporated the shell and the nylon webbing straps to study the head-brain response to vertical struck-by events[54]. In 2017, Wu et al. published results from a model that included a polymer headband in addition to the straps[55]; however, the headband was rigidly attached to the straps and shell, which is not the case in any known commercial models (see Figure 12 for an illustration). While both Long et al.'s and Wu et al.'s models provided valuable biomechanical insights, they cannot be directly compared to one another due to differences in impact conditions and head model, and neither study was directly validated against experiment. As such, it is not known whether the simplifying geometric assumptions used in those models are appropriate, or whether additional details of the hardhat suspension must be modeled.

Figure 12 - Example of a hard hat headband and nylon suspension system. Note that the headband connects to the shell via the four hinged polymer structures, which have been flipped out to horizontal position for visibility[56]

Towards this end, the purpose of this study is to develop a series of FE models of hard hats that parametrically incorporate suspension system features (such as the headband, webbing system, etc.) with the goal of identifying the simplest geometries that can be validated against experimental data. Specifically, we seek to evaluate the effect of varying geometry and material properties in FE models of hard hats on macro-scale predictions of peak linear acceleration (PLA), and to assess which (if any) models can predict outcomes of experimental tests. Such a validated FE model would be valuable for understanding injury patterns and risk, as well as enabling the development and optimization of future hard hat models.

4.2 MATERIALS AND METHODS

The general study design involved validation of various versions of FE hard hat models against matching experimental conditions, with the objective of identifying the simplest FE model capable of replicating experimental outcomes. In order to minimize confounding factors, both experiment and FE simulations used a 50th percentile Hybrid III head/neck form. The Hybrid III FE model was obtained directly from LS DYNA[57,58] and has been validated elsewhere for its ability to match experimental impact conditions[59].

The test fixture used for experimental drop tests consisted of a nine-foot-tall extruded aluminum frame with vertical linear rails, allowing a carriage attached via low-friction pillow block bearings to precisely drop impactors onto the headform. The headform was mounted near the base of the frame as seen in (Figure 13). Tests were conducted with three materials exhibiting a range of mechanical properties: steel, wood, and lead shot. Steel and wood were selected since they are common sources of head injuries[22], and lead shot was chosen to simulate loosely connected construction materials. Each impactor material was controlled to have the same mass of 1.8 kilograms. All tests were conducted with a vertical free-fall drop from a height of 1.83 meters, with the center of the impactor aligned with the center-of-gravity of the headform. The headform was initially tested in an unprotected condition (i.e. without a hard hat) n=3 times for each of the three impactor materials, which was done so that the properties of the impactors could be calibrated in the FE model for a corresponding unprotected condition. Subsequently, the Hybrid III was equipped with a hard hat (MSA 4 Pin V Gard, MSA Safety Inc., Pittsburgh, PA, USA) and n=3 impacts were performed for each impactor material. A pristine hard hat was used for each test.

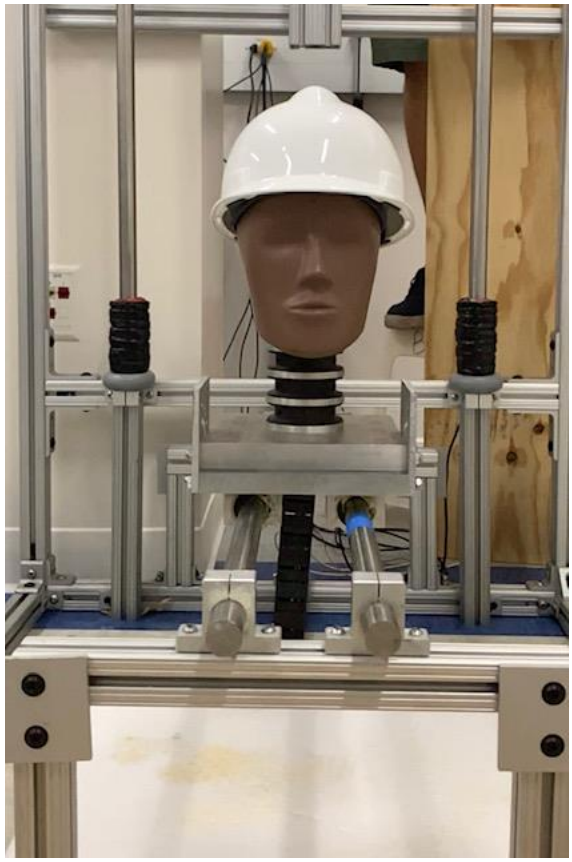


Figure 13 - Experimental Test Fixture

Experimental data was collected via the SLICE MICRO data acquisition system (DTS, Seal Beach, CA). It was set to sample acceleration data at 20 kHz with 4 kHz anti-alias filtering for three linear accelerometers as well as three angular rate sensors (DTS 6DX PRO 2K-18K, Seal Beach, CA). Acceleration pulses were post-processed through a custom MATLAB program (The MathWorks, Inc., Natick, MA) in accordance with SAE J211 (Instrumentation for Impact Tests) using a CFC of 1000. Accelerations and other injury metrics were calculated according to SAE J1727 (Calculation Guidelines for Impact Testing).

Subsequently, a series of FE models were created according to the experimental test conditions. Within the FE modeling environment, three major components were identified that had to be replicated, including the anthropomorphic test device (ATD), impactor materials, and hard hats.

The ATD model used was the 50th percentile male Hybrid III model, created and validated by LS–DYNA (Livermore Software Technology Corporation, Livermore, CA, USA)[57,58]. In that, the ATD model has been validated for its ability to simulate impacts, no changes were made to the geometry or material properties.

As described above, the three impactors used in experiments were a steel bar, a piece of lumber, and a bag of lead shot – each with a mass of 1.8 kg. These were modeled with solid sections in LS-DYNA according to the physical dimensions of the experimental impactors. Approximate material properties were obtained from published sources[60-62] but were allowed to be changed (within physical reason) via a calibration process such that the predicted PLA for the unprotected FE models approximately matched experiment. Rigid material types were used for the steel and wood impactors since no significant deformation of those materials occurred during testing, however elastic properties were assigned that controlled the contact relationship between these impactors and the helmet's shell. The lead shot impactor was modeled as being deformable based on observations from experiment. The final dimensions and properties are provided in Table 11.

Table 11 - Impactor physical dimensions and material properties used in simulations

Impactor	Material Type	Dimensions (mm)	Mass Density (Kg/mm^{3})	Young's Modulus (E) GPA	Poisson's Ratio (ν)
Steel	Rigid	241.3 x 50.8 x 15.875	9.25 X 10^{-6}	210	0.3
Wood	Rigid	311.15 x 233.63 x 38.1	6.56 X 10^{-7}	112.83	0.162
Lead Shot	Elastic	148 x 94.5 x 50.8	2.53 X 10^{-6}	0.05	0.42

The MSA 4 Pin V Gard hard hat was scanned by the Microfocus X-Ray CT System, inspeXio SMX-225CT FPD Plus, from Shimadzu (Shimadzu, Kyoto, Japan). The 3-D geometry was then imported to Geomagic (3D Systems, Rock Hill, SC, USA), a reverse engineering software, which was used to smooth the geometry as well as remove any undesired imaging artifacts and to improve the geometric accuracy. After this process, the model was imported as a STEP file into LS DYNA and meshed. The hard hat was formulated as a shell with rigid material properties, and mechanical properties for high-density polyethylene (HDPE) were assigned: density = 1.27 X 10^{-6} kg/mm^3, elastic modulus = 1.45 GPa, Poisson's ratio = 0.3)[61].

Three variations of the hard hat FE model were created by altering the components of the suspension system within the shell. The first model included only the nylon suspension straps (termed the “straps-only model”). The straps were modeled as two-dimensional shell elements anchored at the center of the attachment slots (the physical location where the straps are attached via a press-fit mechanism in the actual hard hat). The strap material was identified as nylon fabric by the manufacturer[63]. This material has been modeled and validated elsewhere in literature[39], and therefore those published mechanical properties were selected: density = 1.1 X 10^{-6}, elastic modulus = 3.0 GPa, Poisson’s ratio = 0.42.

The headband system is the component which positions, secures and tightens the hard hat to the head. The headband was included in two subsequent FE models, which differed in the way that the headband was connected to the shell. In both models, the headband geometry was simply created as a shell that followed the contour and geometry of the Hybrid III headform. The headband was determined to be an extruded HDPE via Fourier transform infrared spectroscopy Attenuated Total Reflection (FTIR ATR),(Shimadzu, Kyoto, Japan) and corresponding elastic material properties were assigned: density = 1.27 X 10^{-6} kg/mm^3, elastic modulus = 1.45GPa, Poisson’s ratio = 0.3[61]. In the first model, the headband was rigidly attached to the shell at each of the four attachment slots (termed the “rigid headband model”). This was inspired by the model developed by Wu et al. It captures the stabilizing and positioning benefits of the headband interacting with the head, with the potential consequence of enabling a non-physical amount of load to be carried by the headband.

In reality, the headband is connected to the shell using a highly compliant polymeric system. This system consists of tabs that are inserted into the shell via press-fit mechanism, then connected to the headband via v-shaped hinge/tab mechanism (Figure 14). This mechanism allows the headband to move both vertically and laterally within the shell with relatively little force. To capture this behavior, 1-D springs were used to attach the headband to the hard hat shell in the last model (termed the "compliant headband model"). This allowed for the same benefits of the headband as in the previous model but limited the load that could be carried by the headband. The only mechanical property needed for this model was the spring constant, for which a value of 106 N/mm was selected. This value was selected via a qualitative assessment of the headband deformation behavior and further refined through a small number of iterative analyses with the steel impactor (but was not specifically calibrated to match any experimental condition).

Figure 14 - The V-shaped Hinge/Tab Headband Attachment

Figure 15 shows the general geometry that was used for the simulations. With the steel bar being depicted as the impactor. The impactor was aligned with the CG of the headform and was prescribed an initial velocity of 6.0 m/s (selected as the average of the experimental impact velocities from a height of 1.83m). The computational analysis involved 12 simulations, three were the unprotected tests (used to calibrate impactor properties) and the other nine simulations included the three different hard hat models analyzed for each impactor.

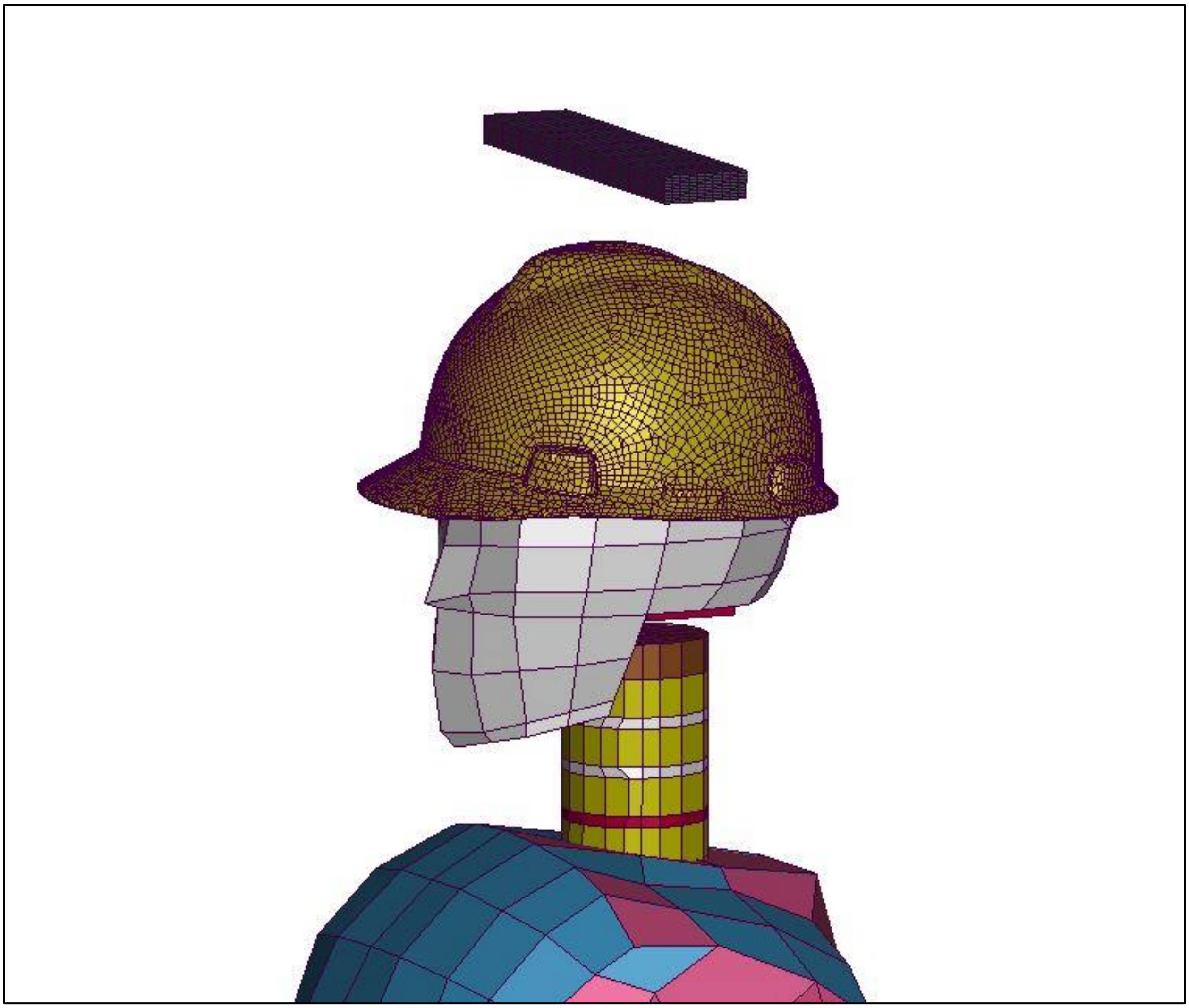


Figure 15 - Finite Element Simulations Test Configuration

4.3 RESULTS

The impactor mechanical properties were calibrated and/or validated using simulations with the unprotected headform. The results obtained from the experimental and computational tests are shown in Figure 16. Experimental measures of PLA varied between about 50 g's for the lead shot to over 300 g's for the steel impactor. In all three conditions, the computational predictions were within 4% of the experimental values – achieved while keeping impactor mechanical properties at (or as near as possible to) the prospectively selected values (Table 1). The steel properties were E=210 GPa and Poisson's ratio=0.3. The range of allowable wood properties that were considered were based on a study from the Federal Highway Administration (FHWA); the specific wood conditions that met the experimental test was the saturated yellow pine properties (consistent with the purchased impactor of pressure-treated yellow pine)[62]. The lead shot was the most complex of the impactors to simulate and we are not aware of any standardized mechanical properties for such an object. Acceptable behavior was found using a 3-D parallel-piped created with the same dimensions as the lead shot bag. An effective density was applied in order to achieve the design mass, and the other properties were defined on a trial-and-error base until the computational simulation resulted in approximately the same accelerations as the experimental tests (E=50 MPa, ν=0.42).

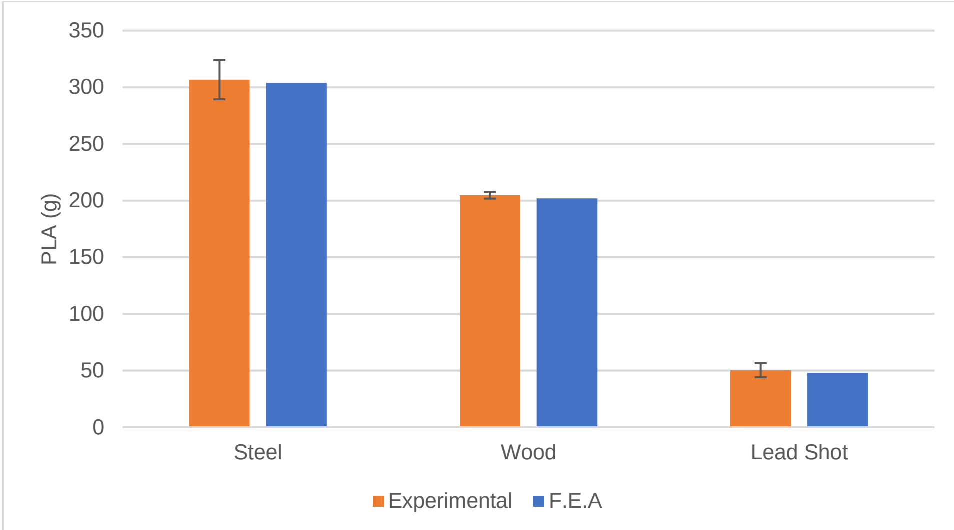


Figure 16 - PLA for unprotected tests, used in calibration/validation of impactor material properties

Using the above-obtained material properties for the impactor materials, simulations were conducted for each of the three hard hat models for each impactor. The results are summarized and displayed in the form of bar charts in Figure 17. Additionally, error percentages are provided in Table 12. For the steel impactor, both the compliant- and rigid-headband materials predicted PLA from experiment within 5% error, whereas the straps-only model under-predicted by 22%. For the wood impactor, the compliant headband model under-predicted by a little more than 5%, while the straps-only model over-predicted by 2% and the rigid-headband model over-predicted by 40%. None of the models agreed perfectly with the lead-shot experimental results, but the compliant headband model was the best performer with 29% error (approximately a 3g over-prediction), while the rigid-headband and straps-only models over-predicted by 54% and 152%, respectively.

Table 12 - Percentage Difference in PLA from computational tests compared to the average

IMPACTOR	COMPLIANT HEADBAND MODEL	RIGID HEADBAND CONNECTION	STRAPS ONLY
STEEL	-3.39%	-1.27%	-22.46%
WOOD	-5.56%	40.46%	2.11%
LEAD SHOT	29.03%	151.79%	63.98%

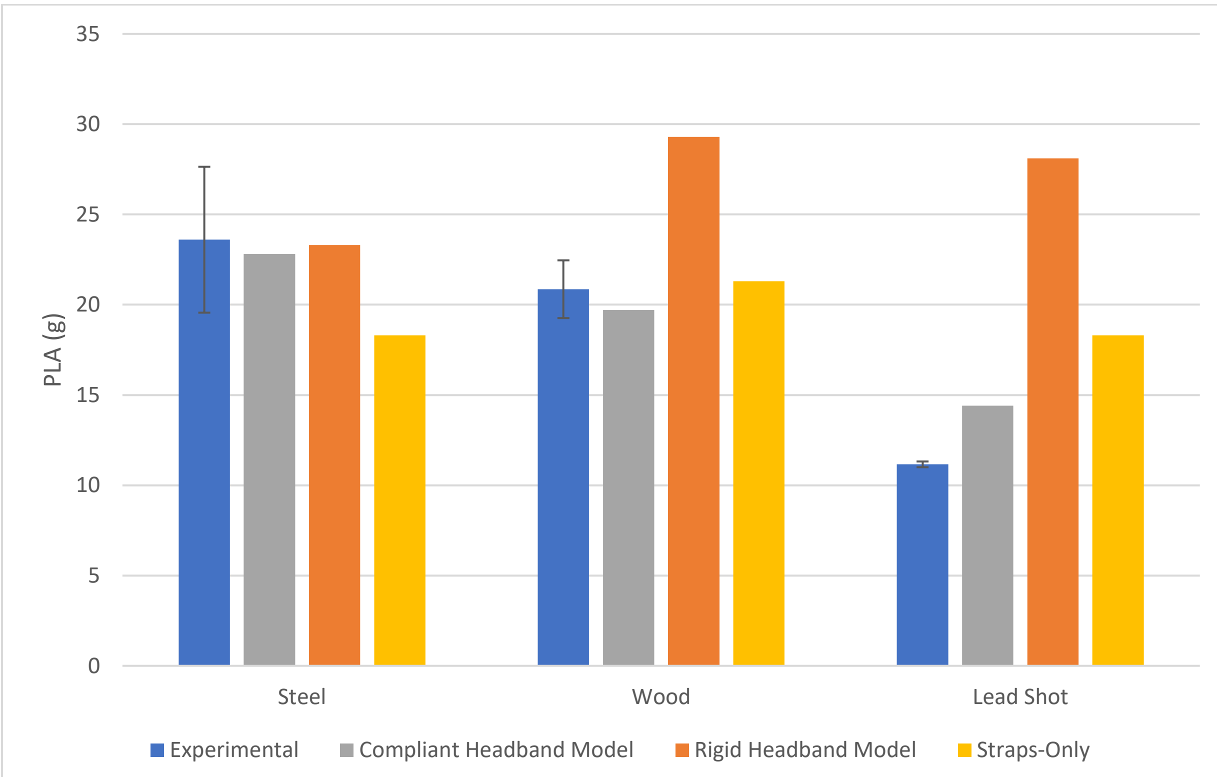


Figure 17 - Summary of the PLA Results Obtained by Each Hard Hat Model in Comparison to Experimental Testing

4.4 DISCUSSION

Three different computational models of a hard hat were analyzed for their ability to predict experimental measures of PLA in this study. The three models primarily varied regarding the level of detail incorporated in modelling the suspension system (including both the webbing straps, headband, and headband attachment to hard hat shell). While the performance of each of the models varied with impactor type and no model was able to perfectly predict all impact conditions, the most complex model (the compliant headband model) was generally the best performer – it accurately predicted hard hat performance in the wood and steel impacts, and also was the best predictor for the lead shot impacts. The rigid headband model generally predicted higher accelerations than were experimentally measured, which is likely a result of the headband being capable of carrying a non-physical load, thereby resulting in stiffer response from the model. The simplest model, which only including the suspension webbing material, accurately predicted the wood impact, but not the other conditions. We also observed that the lack of connections as well as stabilizing mechanisms made the straps-only model poorly behaved from a computational perspective. Taken together, our data suggests that models with greater levels of geometric/physical fidelity perform better at predicting experiment.

We are aware of two published studies involving the analysis of similar impacts to those studied in this present project – Long et al. and Wu et al. developed similar models to two of the models studied here (similar to the straps-only and rigid-headband model, respectively)[54,55]. While neither of these geometric models were able to consistently predict experimental results across all impactor types in this present study, we note that the both model types performed reasonably well for the types of impactor materials simulated by Long and Wu (steel and wood, and steel, respectively). Therefore, the general trends and findings presented in those studies are likely to be reasonably accurate. Nonetheless, findings from the present study suggest that more accurate geometric/physical modeling of hard-hat suspension components can improve results.

Despite the overall success that our models had in predicting experimental measurements of PLA, several limitations should be noted. First, none of the models developed here explicitly modeled the exact geometry of the headband system or the flexible polymeric attachments that connect the headband to the shell. The springs used here mechanistically simulate the behavior of flexible attachments in a linear region. However, for impactors with larger masses (~3.6 kg), the polymer attachments have been observed to experience large deformations (even to the point of failure in many instances), and this failure has been hypothesized to play a significant role in impact attenuation performance[64]. It is therefore likely that more complex geometric and material modeling of the headband system is required to improve results, particularly for simulating struck-by events with larger masses.

Furthermore, workers typically do not work standing perfectly upright, which was the test configuration simulated in this study. In fact, Moriguchi et al. identified that typical work postures involve significant forward flexion of the head/neck[49], which is consistent with epidemiological data from the BLS that indicates the most common head position during injurious accidents was partially looking down[22]. We chose to simulate vertical impacts to an upright head since it is the configuration used ANSI hard hat certification standards[28] and it is also the most common configuration analyzed in literature. From a mechanical perspective, the headband components of a hard hat should play a larger role in impacts with forward-flexed head orientations, and therefore the accurate modeling of these features should be more important for simulating such events. Towards this end, future studies must examine the role of geometry and material models during impacts with different head orientations in order to develop fully validated FE simulations of hard hats.

Finally, although the validation of this finite element model in present study was based solely on PLA results, other injury metrics may also be of interest in other contexts. As an example, while validating HIC measures was not a specific objective of this project, that metric can be calculated for both experimental and computational impacts conducted for this study (Table 13). As seen in Table 13, none of the models consistently predicted the actual experimental outcomes for HIC. The rigid headband model consistently over-predicted HIC and suffered from the most error. The straps-only model was the best predictor for steel but predicted value of HIC from that model did not vary much across the three impactor types. The compliant headband was the best predictor of HIC in two out of the three impact conditions (wood and lead shot) but was the worst predictor for the steel condition. Despite this error, an important mechanistic benefit was identified in the compliant headband model that was not seen in the other two models – it was the only model to correctly rank HIC values (from smallest, lead, to largest, steel), while the other two models appeared to exhibit an impactor-dependent bias. These findings highlight the need for future work pursuing improvements in both the material models and geometric features used in FE models of hard hats, particularly as more complex impact conditions are simulated.

Table 13 - Summary of the HIC Injury Metric Results

Impactor	Experiment	Straps-Only	Rigid Headband Model	Compliant Headband Model
Steel	6.8 ± 0.1	6.4	8.8	10.2
Wood	6.1 ± 0.7	8.1	18.3	7.3
Lead	1.8 ± 0.1	5.6	14.8	4.8

Chapter 5 – Conclusion

5.1 CONCLUSION BY CHAPTER

5.1.1 Chapter 2

This study sought to provide basic biomechanical data on the ability of hard hats to attenuate accelerations associated with traumatic brain injury and to investigate the contribution of high-level design differences (webbing attachments and headband mechanisms) in this regard. Hard hats substantially reduced head accelerations relative to the unprotected condition – often by as much as 95%. Moreover, the parametric comparison between hard hat designs was examined to explore the effect of hard hat features in reducing head accelerations. Ratchet tightening systems and added webbing attachment points did not consistently help attenuate head accelerations, as there were no significant statistical differences in performance across designs for all test conditions. Finally, damage to the helmet suspension was observed in several tests; the presence of failure generally reduced head accelerations for the 4 lbs. tests, but not for 8 lbs.

5.1.2 Chapter 3

The purpose of this study was to experimentally test the impact attenuate performance of hard hats on a forward-flexed headform (30° of simulated neck flexion) as compared to an upright (0° of flexion) position. Hard hats generally reduced injury metrics for both upright and forward-flexed test conditions, supporting their use in all situations with the potential for overhead impacts. While linear acceleration metrics were comparable in upright versus forward-flexed tests, angular accelerations were substantially larger for forward-flexed conditions. Moreover, hard hats failed to attenuate angular accelerations for some forward-flexed conditions. While different hard-hat models afforded varying levels of protection across the range of test conditions studied, there was no condition where results from upright testing correlated to results from forward-flexed testing. Furthermore, decoupling of helmets was also noticed to have a positive effect on attenuation of accelerations.

5.1.3 Chapter 4

This study sought to validate FE models of hard hats against experimental measures of PLA, obtained by simulating struck-by construction accidents using impactor objects with a range of stiffnesses. While many of the simplified models considered here provided reasonable estimates of head acceleration for individual impactors, the best results were achieved by the model that simulated the geometry and components of the hard hat's suspension system with the most detail. Future numerical simulations of struck-by accidents should therefore use hard hat models that account for the mechanical contributions of both the webbing strap system and the deformable behavior of the headband structure whenever possible.

5.4 FUTURE WORK

The studies conducted lead to many new avenues of research that can be executed and that will be extremely beneficial to the construction safety industry. The first study generated interesting questions regarding the effect of the hardhat features in hard hat head acceleration attenuation. For instance, variations in headband attachment performance caused large variability in results due to its failure behavior. Thus, the first path that can be further studied is the design of these features, the material they are made of and the different options commercially available. In the end, all of these areas of interest can potentially optimize such a feature to improve PPE performance.

The second series of experimental tests conducted evaluated the most common impact postures associated with field-reported injuries. The different head orientation was proven to be more injurious and expose users to scenarios of possible TBIs. Further testing should focus on the specific features that could potentially positively influence performance (i.e. decoupling) and the headband material properties that caused this phenomenon. The variety of hard hats available from construction is enormous, but it would bring many benefits to workers' safety if these features (and others) were further examined. Additionally, force transmission tests were not conducted as part of the experiments in this study, however, it is one of the focuses of ANSI/ISEA Z89.1 2014. Thus, it would important to analyze the performance of hard hats regarding different injury metrics and injuries in general with a different head orientation.

The validated finite element model developed here opens up several branches of potential research that can be pursued in the future. Initially, new generations of the validated model can be developed that incorporate even more geometric detail as well as non-linear material models that can capture the failure behavior of suspension components. This model could then be used as an optimization tool, which could accomplish fast, cost effective and nondestructive development/testing of new features. In addition to the benefits associated with designing, this finite element model could also be simulated with a human model rather than the ATD. The implementation of the human head model would enable the analysis of stresses and strains in biological tissues and could thereby provide more detailed information on injury mechanisms and prevention.

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