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Experimental Analysis of Protective Headgear Used in Defensive Softball Play

By

John Scott Strickland

A Thesis submitted to the Department of Mechanical Engineering in partial fulfillment of the requirements for the degree of Master's of Science in Mechanical Engineering

UNIVERSITY OF NORTH FLORIDA

COLLEGE OF COMPUTING, ENGINEERING, AND CONSTRUCTION

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This Thesis titled Experimental Analysis of Protective Headgear Used in Defensive Softball Play is approved:

Grant Bevill, PhD, PE
Thesis Advisor and Committee Chairperson

Stephen Stagon, PhD
Committee Member

Alexandra Schönning, PhD
Committee Member

Accepted for the School of Engineering:

Osama Jadaan, PhD
Director of the School of Engineering

Accepted for the College of Computing, Engineering, and Construction:

William Klostermeyer, PhD
Dean of the College of Computing, Engineering, and Construction

Accepted for the University:

John Kanter, PhD
Dean of the Graduate School

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ABSTRACT

Every year in the United States, an estimated 1.6 to 3.8 million people sustain sports-related traumatic brain injuries (TBIs), with an appreciable number of these injuries coming from the sport of softball. Several studies have analyzed the impact performance of catcher's masks within the context of baseball; however, virtually no studies have been performed on fielder's masks within the context of softball. Thus, the main objective of the present work was to evaluate the protective capabilities of softball fielder's masks. To better understand the injury mechanisms and frequency associated with softball head/facial injuries, epidemiological data from a national database was reviewed first. Results displayed "struck-by-ball" as the most frequent injury mechanism (74.3%) for all head/facial injuries with a large majority occurring to defensive players (83.7%). With further motivation, the present work focused on testing the impact attenuation and facial protection capabilities of fielder's masks from softball impacts. Testing with an instrumented Hybrid III headform was conducted at two speeds and four impact locations for several protective conditions: six fielder's masks, one catcher's mask, and unprotected (no mask). The results showed that most fielder's masks reduced head accelerations, but not to the standard of catcher's masks. On average, they reduced peak linear and angular acceleration from 40-mph impacts by 36-49% and 14-45%, respectively, while for 60-mph impacts they were reduced by 25-42% and 13-46%, respectively. Plastic-frame fielder's masks were observed to allow facial contact when struck at the nose region at high speed. Observed differences in impact attenuation across fielder's mask designs further suggested influence from specific design features such as foam padding and frame properties. Overall, the results clearly demonstrate that head/facial injuries may be mitigated through the broader use of masks, while further optimization of impact attenuation for fielder's masks is pursued.

CHAPTER 1 – BACKGROUND INFORMATION

1.1 INTRODUCTION

An estimated 1.6 to 3.8 million sports-related, traumatic brain injuries (TBI) occur annually in the United States alone [1]. An appreciable number of these injuries come from softball - a popular sport in the United States, with an estimated 10-12 million people participating annually [2]. However, serious head and face injuries have been observed to be sustained as part of the sport, with rare cases even leading to death. For example, in 2007, a 12-year-old Michigan girl, who was struck in the head by a ball while practicing with her team, sustained a serious brain injury that led to her death a day later [3]. And, in 2016, a 39-year-old Tennessee woman was umpiring for a charity softball game when a foul tip struck her face mask, resulting in a closed head injury that she died from a couple days later [4].

Many elements of softball play involve high-energy events that are capable of causing serious facial and head injuries, including pitching, batting, or running into other players or objects. For example, batted balls include energy from the original pitch as well as the swung bat [5], resulting in ball speeds as high as 99 mph. The kinetic energy associated with a 99-mph ball is comparable to a well-executed karate strike [6], and players, such as the pitcher, may have as little as 0.36 seconds of reaction time before being impacted [7]. Foul tips or "bad hops" present a similar danger, and may provide softball participants, such as catchers and umpires, even less time to react to the impending impact.

It is known that protective headgear exists for all player positions within softball to assist in preventing or mitigating such injuries. However, it is unknown in how well protective headgear for defensive fielder positions (i.e. fielder's masks) perform in attenuating softball impacts (Figure 1). A review of biomechanics literature demonstrates that only the protective equipment available

to batters and catchers have been evaluated, whereas, virtually no studies have been performed for the facemasks available to fielders. In addition, very few, if any, fielder's masks are certified by the National Operating Committee on Standards of Athletic Equipment (NOCSAE), wherein they require that impacts to defensive protective headgear for baseball/softball must not exceed a Severity Index (SI) of 1200, not display permanent damage under their specified test conditions, and impacts must not allow contact with certain facial regions [8].

Figure 1: Examples of fielder's masks for softball. (Left to Right: Steel-frame fielder's mask, Plastic-frame fielder's mask)

Within this context, the main objective of the present work was to evaluate the protective capabilities of fielder's masks. This study was broken down into two components. The first section involved reviewing injury data from a national database to identify the statistical distribution of softball injuries and injury mechanisms. The second section consisted of experimentally evaluating several brands of fielder's masks and comparing them to a catcher's mask and no mask.

1.2 RELEVANT LITERATURE

Several published studies were identified that involve assessment of the impact attenuation capabilities of catcher's masks in the context of baseball. Aside from differences in gameplay between baseball and softball (i.e. ball material, average ball speeds, etc.), these studies nonetheless provide relevant background knowledge for the present study since the types of catcher's masks used in baseball are very similar, if not identical, to those used in softball.

Shain et al. [9] investigated impact performance to the nose region of catcher's masks from baseball impacts ranging from 60-80 mph. This study found peak linear accelerations of 140-180 g with no mask reduced to 16-30 g with a catcher's mask, whereas peak angular accelerations fell from 19,500-27,500 rad/s^2 to 2250-3230 rad/s^2. Significant reductions (upwards to ~85%) in head accelerations were observed (as compared to no mask protection) and it was noted that these reductions placed concussive risks below current established injury thresholds [9].

Beyer et al. [10] expanded on this knowledge by analyzing the impact performance of catcher's masks from baseball impacts at seven locations on the facemask: forehead, lateral forehead, eyebrow, lateral eyebrow, nose, lateral nose, and chin. At higher impact speeds (84 mph), Beyer et al. reported that the most susceptible regions of impact (i.e. chin and eyebrow) exhibited 26-42 g and 1974-5266 rad/s^2 for peak linear and angular accelerations, respectively. It was also reported that catcher's masks reduced head accelerations below established injury thresholds [10].

Schwizer et al. [11] conducted a study in assessing catcher's mask impact performance when foam padding properties were adjusted. Computational methods via finite-element modeling (FEM) were used along with experimental methods (much like those used by Shain et al. and Beyer et al.), and both methods were found in agreement with their results. Schwizer et al. observed that

facemask impacts at higher speeds attenuated better with stiffer foams, while at lower speeds the catcher's mask attenuated better with softer foams [11].

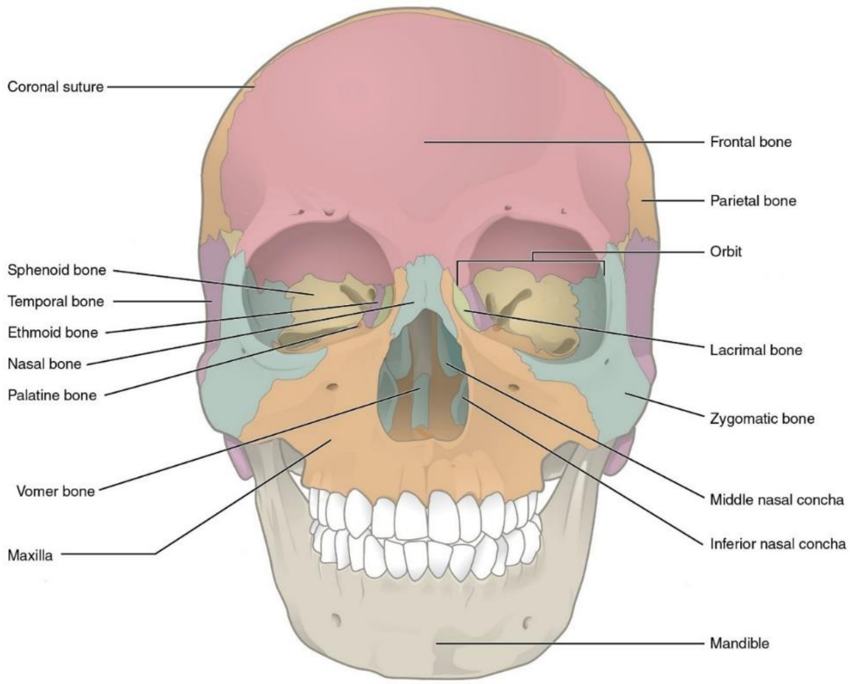
1.3 BACKGROUND – ANATOMY, INJURY, AND METRICS

This section of the thesis provides relevant knowledge to the reader to better understand the context and background of the conducted studies. The following topics are addressed: head anatomy, head injuries and mechanisms, and head injury measurements and criterion. Head anatomy will cover the skull, brain, and relevant surrounding structures, such as the scalp and meninges. These sub-topics will provide information on structural layout, function, and any relevant material/mechanical properties. Head injuries and injury mechanisms will include discussion on various skull/facial and brain injuries, and the physical processes associated with causing them. Lastly, head injury measurements and criterion will present outcome variables that are typically collected during experimental studies, and how they are used to evaluate the severity of an injury.

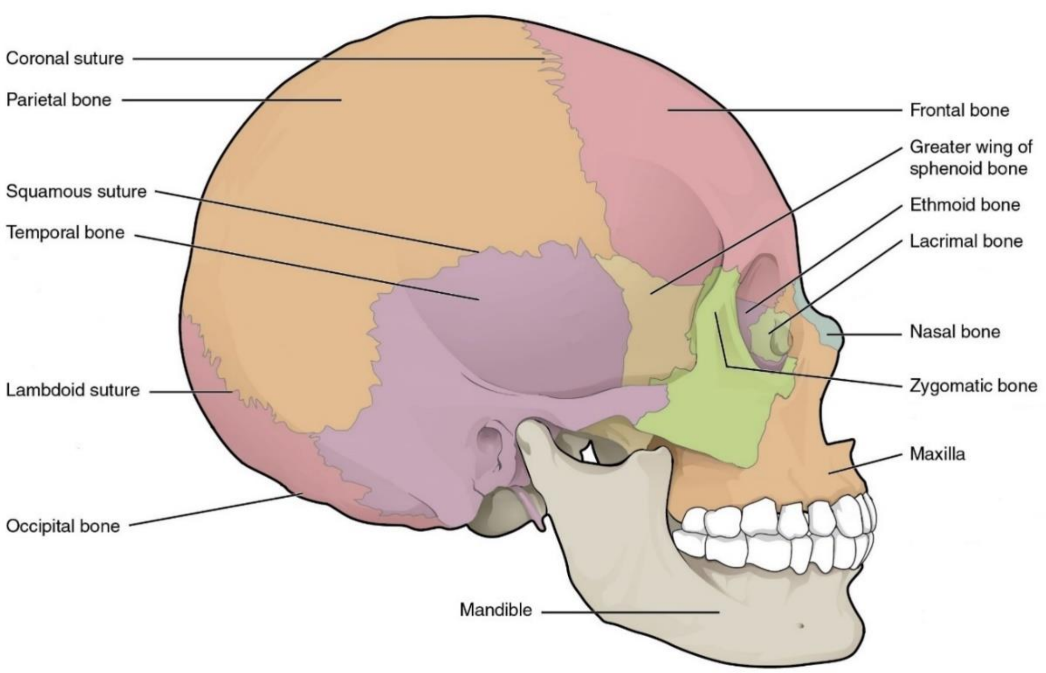
1.3.1 HEAD ANATOMY

1.3.1.1 SKULL

The skull is a sturdy osseous structure that houses and protects the brain. In its sum, the skull consists of 23 bones that subcategorize into two main components: the neurocranium (cranial vault) and viscerocranium (facial skeleton) (Figure 2). The neurocranium is the portion that encases the brain, meninges, and some parts of the cranial nerves and blood vessels by providing a dome-like roof (i.e. skullcap) and a floor (i.e. cranial base). The neurocranium consists of 8 bones: four singular bones (frontal, ethmoidal, sphenoidal, and occipital) and two sets of bilateral bone pairs (temporal and parietal). On the inner surface of the neurocranium, there are several small holes that allow for these blood vessels and nerves to pass through. Additionally, towards the base, there is a large hole that allows for the brain stem to pass through into the spinal cord. The viscerocranium is the facial skeleton that forms the frontal portion of the skull and consists of bones surrounding the mouth, nose, and eye regions. The viscerocranium region consists of 15 bones: three singular bones (mandible, ethmoid, and vomer) and six sets of bilateral bone pairs (maxillae, inferior nasal conchae, zygomatic, palatine, nasal, and lacrimal). This region contains the only free moving joint in the skull structure: the mandible, a bone of which grants the function of breaking up food for consumption [12].



(a)



(b)

Figure 2: The skull shown in an (a) anterior view and (b) lateral view.

(Adapted from Rice University [13])

Many bones across the skull are fused together with sutures (a type of immobile joint consistent of fibrous, interlocking connective tissues). Several bones (i.e. frontal, temporal, sphenoid, and ethmoid bones) are pneumatized, meaning they contain air spaces that serve to decrease their weight. As an individual ages, these air spaces increase, which likely prompts a weakening of the bone against external forces [12].

The mechanical properties of the skull are viscoelastic, which serves to protect the internal soft tissues more effectively [14, 15]. Areas of higher bone volume percentage within the skull have higher elastic moduli and maximum bending stress. For example, the frontal portion of the skull is less porous, thicker, and has a higher bone volume percentage than the parietal portion. As a result, this allows it to take higher forces and absorb more energy before fracture. The frontal, temporoparietal (lateral), and occipital bones have respective peak fracture forces of 4.2-6.2 kN, 2.0-5.2 kN, and 12.5 kN [16, 17, 18, 19, 20]. However, not all mechanical or geometric properties of the skull are the same from person to person due to biological heterogeneity. For example, in a sample of individuals, the thickness, curvature, and porosity of the skull bones can vary, and as such, a range of properties exist [15]. Nonetheless, the skull is better adapted at sustaining higher loads when impacted dynamically as opposed to quasi-statically [21].

1.3.1.2 BRAIN

The brain is the most critical component of the head anatomy since it is the control center for all function in the human body. The brain can be categorized into three parts based on structure and functionality: cerebrum, cerebellum, and brainstem (Figure 3).

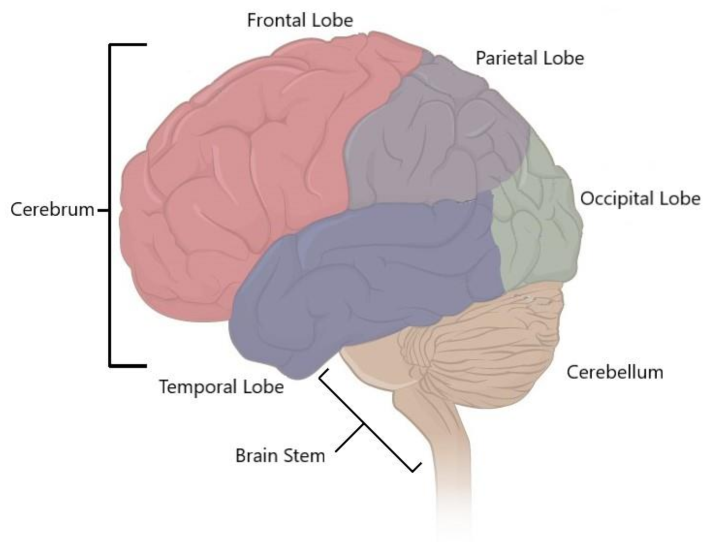


Figure 3: Lateral view of the brain. (Adapted from Rice University [13])

The cerebrum includes two components: the cerebral hemispheres and basal ganglia. The cerebral hemispheres, which are connected by a bundle of axons called the corpus callosum, are the dominant features of the brain, and they are divided up into four lobes: frontal, parietal, temporal, and occipital [12]. Each lobe of the cerebral hemispheres provides specific functions of higher level action, such as thought, sight, and hearing. The frontal lobe is associated with reasoning, planning, parts of speech, movement, emotions, and problem solving. The parietal lobe is associated with movement, orientation, recognition, and perception of stimuli. The temporal lobe is associated with perception and recognition of auditory stimuli, memory, and speech. Lastly, the occipital lobe focuses primarily on visual processing [22]. Covering the cerebral hemispheres are the gyri (folds), sulci (grooves), and fissures (clefts) [12]; the presence of these features grant additional surface area to achieve the necessary capabilities required of the brain in its confined spacing.

The cerebellum is the second largest component of the brain, and is positioned beneath the rear portion of the cerebrum. Primary functions of the cerebellum include coordination of skeletal-

muscle movement, maintenance of balance and posture, movement error detection, and predicting body position during movement. A secondary function of the cerebellum is the modulation of sensations related to anger and pleasure [23].

The brainstem is a crucial component consisting of three main structures: midbrain, pons, and medulla oblongata; all of which are positioned inferior to the cerebrum and cerebellum. The midbrain is involved in pain suppression, alertness, vision, hearing, and body coordination [24]. The pons, which is positioned below the midbrain, acts as a bridge between the cerebrum and cerebellum, and is involved in controlling facial expressions and sideways eye movement, processing sounds, maintaining balance, and chewing. Lastly, the medulla oblongata, which is located below the pons, helps transfer messages between the spinal cord and thalamus. In addition, it helps regulate breathing, heart and blood vessel function, digestion, sneezing, and swallowing [25].

Across the brain, there are two types of tissue: white and gray matter (Figure 4). Gray matter is where neuronal cell bodies and neuropil (i.e. dense areas of axon terminals and dendritic branches for where synapses occur) are contained [26]. White matter inhabits the deeper, bulk regions of the brain and contains bundles of axons that allow for different gray matter parts to connect and communicate with each other. All axons are covered with an insulation, called myelin, to ensure proper neurological communication. Any damage or deterioration to the axons can end up causing disruptions to normal motor, sensory, and cognitive functions [27].

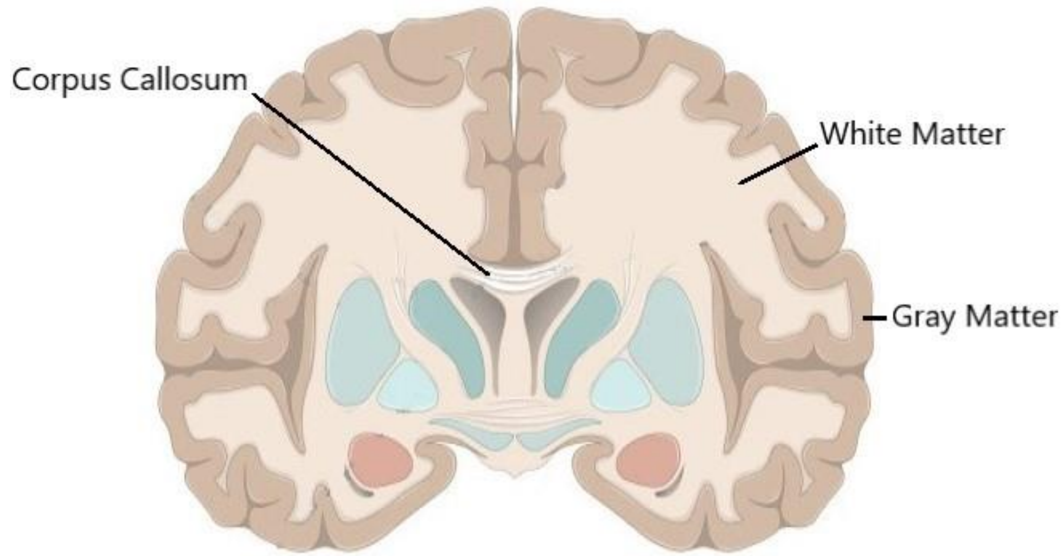


Figure 4: Cross-sectional, anterior view of the brain, displaying white and gray matter. (Adapted from Rice University [13])

In general, the material properties of the brain are anisotropic, non-homogenous, and nonlinearly viscoelastic. The anisotropic aspect is present in the structural geometry and layout of all its various components. The gyri, sulci, and fissures present in the cerebral cortex are example indicators of this property. The non-homogenous property is displayed by the presence of regional differences with respect to white and gray matter. White matter is more fibrous than gray matter, and as such, these regional differences can affect stress/strain fields during an impact event [28]. The nonlinear viscoelastic nature is evident in that brain tissue stiffens as the rate of deformation increases, and that there is a non-linear relationship for brain tissue deformation with respect to force [29, 30]. Finally, it is also important to note that the bulk modulus of the brain is approximately 5 to 6 orders of magnitude larger than its shear modulus, and as such, it more predisposed to deform in shear [31].

1.3.1.3 SURROUNDING STRUCTURES

The surrounding structures of the head include two sub-components: the scalp and meninges. The scalp is a combination of five layers (5 to 7mm thick) which lay above the bone of the skull. It consists of the following: hair-bearing skin, a dense connective tissue layer, a muscle and fascial layer, a loose connective tissue layer, and the pericranium, another dense connective tissue layer. Below the skull is the meninges, whose purpose is to protect and support the brain, form a supporting framework for arteries, veins, and venous sinuses, and provide enclosure for surrounding fluids. The meninges consist of the following structures, from top to bottom: epidural-space, dura mater, subdural space, arachnoidea mater, subarachnoidal space, and pia mater**.** The dura mater is a tough, fibrous membrane, while the arachnoidea mater is a thin membrane that resembles a spider-web. Lastly, the pia mater is a very thin, internal, vasculated membrane that covers the surface area of the brain.

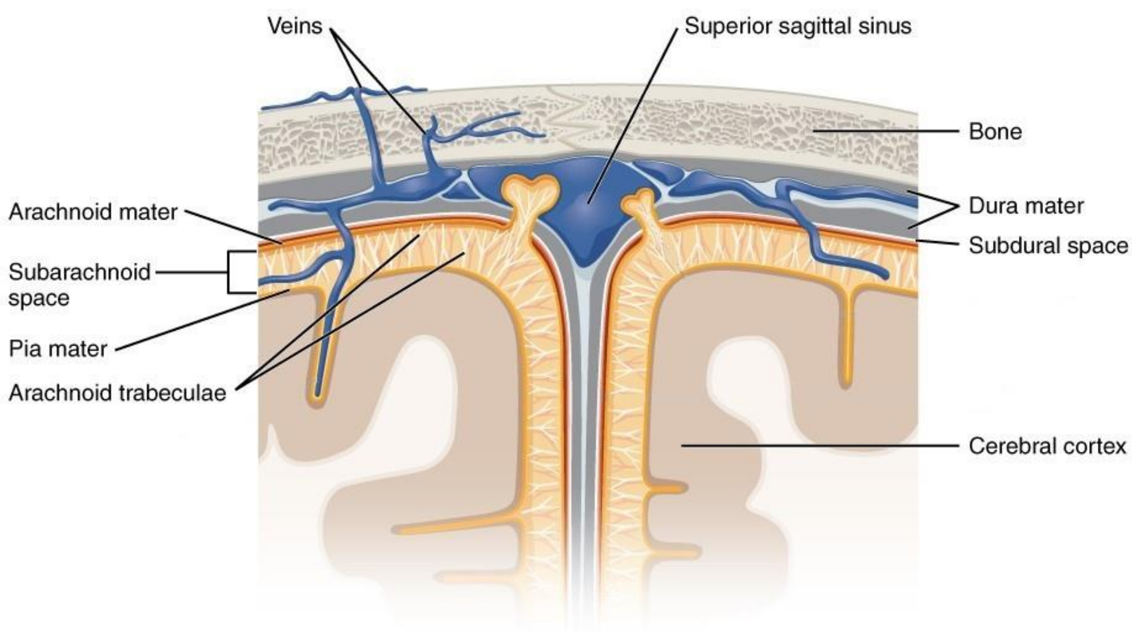


Figure 5: Cross-sectional view of the surrounding structures to the brain.

(Adapted from Rice University [13])

Within the meninges, there are several blood vessels that cross to supply the brain and scalp. The most notable of these blood vessels are the veins that pass through the subdural space. These veins, often called “bridging veins,” are considered of importance since they can be subject to tearing. Within the subarachnoid space and the ventricles of the brain, there is a clear liquid called cerebrospinal fluid (CSF) that fills and circulates through these areas. This fluid surrounds the brain on all sides and serves as a buffer to any mechanical shock, as well as to support the weight of the brain and provide nutrients to it [12, 32].

1.3.2 HEAD INJURIES AND MECHANISMS

In the broader context, head injuries are classified as either open or closed. The role of the injury is determined by whether the dura mater has been punctured (open) or not (closed). There are many mechanisms that can cause head injuries to occur. From an epidemiological perspective, the mechanisms are associated with some of the following causal events: motor vehicle crashes, sports and recreation, assaults and violence, and falls. In terms of the physical processes, head injuries result from either static or dynamic loading. Static loading is generally defined as a load that lasts for more than 200 ms; these loads are typically rare in occurrence. Dynamic loading is the most common with a variety of responses, and can be defined further as either contact or non-contact (i.e., an inertial event) [32].

1.3.2.1 SKULL/FACIAL INJURIES

Skull/facial injuries consist of fractures and/or soft tissue damage. Fractures can be classified into four categories based on injury location: basilar, vault, nasal, and maxillary. Basilar

fractures are around the base of the skull, while vault fractures are around the cranial region. Nasal fractures are at the nose region, as the name implies, and maxillary fractures, which are often considered serious, happen on the facial regions surrounding the nose. Soft tissue injuries commonly involve any contusions (e.g. bruising) or lacerations to the tissue layers sitting above the skull. These injuries are generally of lesser importance since they are rarely life-threatening [32].

These injuries are a result of static or dynamic contact loading to the skull/facial region. Substantial contact loading either to or from an object can cause the skull to deform, resulting in these fractures directly at the site (i.e. bending fracture) or indirectly, where they are oriented in the direction of the force vector (i.e. burst fracture) [32]. At 180 g, the skull is at a 5% risk of fracture, while at 250 g, it is at a 40% risk [33]. However, the force and linear acceleration values associated with skull fractures can vary depending on the impactor surface curvature [17, 34].

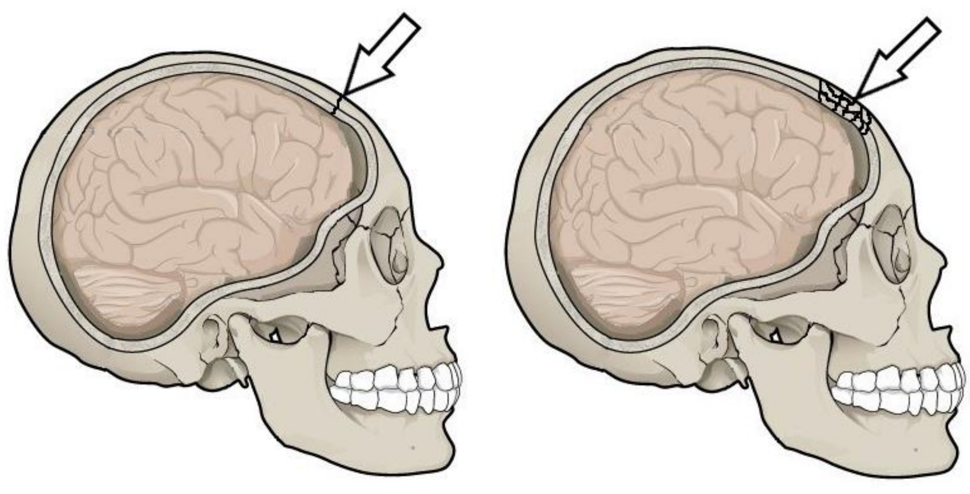


Figure 6: Force applied to skull displaying (a) bending-type fracture and (b) burst-type fracture.

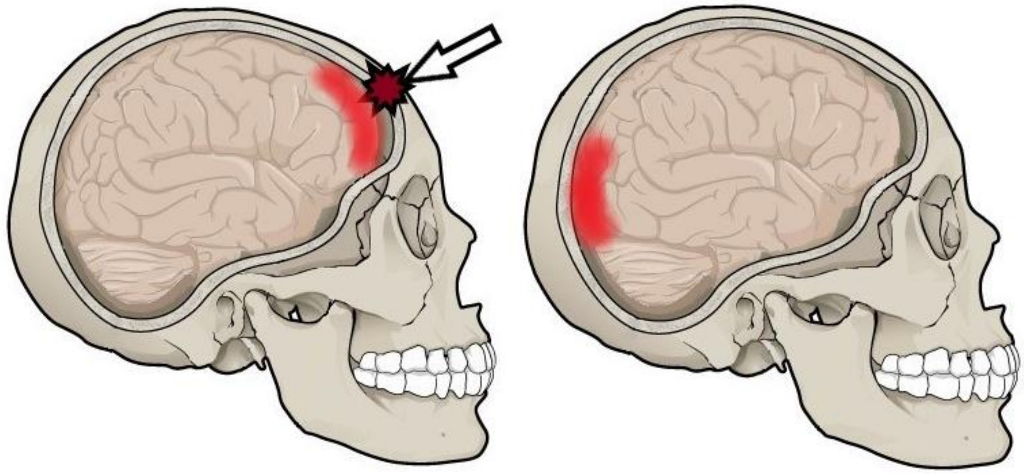
(Adapted from Rice University [13])

Alternatively, soft tissue injuries could be a result if the contact loading is only sufficient enough to damage the tissue layers via a compressive or shearing force. An example injury event for fractures would be a baseball/softball colliding at an individual's face at 90 mph, whereas for soft tissue injuries it may be the forceful impact of an automotive air bag deploying into the face (e.g. bruising).

1.3.2.2 BRAIN INJURIES

In general, there are two types of brain injuries: focal and diffuse. There are several possible mechanisms responsible for causing these brain injuries: static loading, dynamic contact loading, or dynamic non-contact loading. They are typically a result of dynamic impact events with high linear accelerations and/or angular accelerations. However, studies have shown that most brain injuries are likely to occur under rotational motion as opposed to linear [35, 36, 37, 38].

Focal brain injuries are defined by damage to a local region are classified as either contusions or hematomas. Contusions are the most common injury following a head impact, where they either occur at the site of impact (coup contusion) or opposite of it (contre-coup contusion) (Figure 7). Contusions happen because of acceleration-induced movement imparted on the brain causing it to collide with, or glide against, the inner surfaces of the skull [32, 39, 40, 41].



(a) (b)

Figure 7: Force applied to forehead displaying (a) coup contusion (primary impact) and (b) contre-coup contusion (secondary impact). (Adapted from Rice University [13])

Hematomas are defined as brain bleeding, which are further classified by their internal location relative to the meninges (e.g. epidural, subdural, or subarachnoid). Epidural hematomas are often a result of trauma to the skull and the underlying blood vessels. Subdural hematomas can be caused by three different injuries: lacerations of cortical veins and arteries by penetrating wounds, large-contusion bleeding into the subdural space, and tearing of bridging veins between the brain's surface and the dural sinuses [42]. The large collections of blood due to these hematomas can result in a build-up of intercranial pressure (ICP), therefore eventually leading to permanent brain damage and death.

Diffuse brain injuries differ from focal injuries in that damage can occur in multiple locations or over a much larger area. These injuries are typically a result of impact events with high angular accelerations, and they can form a spectrum ranging from mild traumatic brain injury (MTBI) to severe damage on the axons and white matter. MTBIs are the most common,

particularly in sports, and account for about 75% of traumatic brain injuries (TBIs) [43]. Typically, these types of injuries are referred to as mild traumatic brain injury (MTBI), since they involve the individual retaining consciousness post-impact, are fully reversible, and display no detectable problems via medical imaging. However, it's been observed that repeated impacts resulting in MTBIs are often subject to developing into chronic problems (i.e. neurodegeneration), some of which that cannot be diagnosed until post-mortem, such as chronic traumatic encephalopathy (CTE) [44]. The severe end of the concussive spectrum is characterized by immediate loss of consciousness, lack of motor responses, etc. Diffuse axonal injury (DAI) is one such outcome where the axons in the cerebral hemispheres and the white matter are damaged [32].

1.3.3 HEAD INJURY MEASUREMENT AND CRITERIA

Head injury mechanisms and their resulting injuries can be evaluated qualitatively and quantitatively. Various forms of measurements and criteria exist for quantitative measurement - the two simplest and most commonly used being peak rotational acceleration and peak linear acceleration. Linear acceleration measurements are often used in predicting the risk of skull fractures. Rotational accelerations, on the other hand, are thought to play a major role in the occurrence of brain injuries [31, 32].

Although acceleration data provides some meaningful insight on its own, specific criteria have been developed that account for additional aspects of an impact. For example, the Wayne State Tolerance Curve (WSTC) measures head acceleration through a relationship between time duration and linear acceleration from frontal impacts [45]. The WSTC shows how larger magnitudes of linear acceleration can be sustained in shorter impulse durations, while lower magnitudes for longer impulse durations. However, the WSTC itself is limited in that it cannot be

applied to other loading directions and non-contact loading conditions [32]. The WSTC has since then been adapted into other forms, such as the Severity Index (SI) [46] and Head Injury Criterion (HIC) [47]. To further supplement the stock of criteria, other studies have developed their own, such as Head Impact Power (HIP) [48, 49], Rotational Injury Criterion (RIC), and Power Rotational Head Injury Criterion (PRHIC) [50].

The Severity Index, also known as the Gadd Severity Index (GSI), is the first developed head injury criterion of the WSTC, in which it plots the curve on a logarithmic scale to form a straight line. Severity Index is represented by equation (1), where 2.5 is a weighting factor (i.e. slope of the line) determined from the WSTC, *t* is time, and *a* is linear acceleration:

$$SI = \int_{t_0}^{t} a^{2.5}\, dt \tag{1}$$

This criterion is frequently used by the National Operating Committee on Standards in Athletic Equipment (NOCSAE) in certifying protective headgear for sports. Based on NOCSAE standards for baseball/softball headgear, an SI of 1200 is the maximum allowable exposure from an impact [8, 51].

Head Injury Criterion uses linear acceleration and expands upon the SI equation. It is considered one of the most widely used injury criteria, particularly for automotive collision testing. The difference between HIC and SI is that HIC assigns a time limit to the impact duration. For automobiles, the limit is 36 ms [52], while for direct impacts and helmets it is 15 ms [53]. Based on the assigned time limits, HIC is denoted as HIC_{36} or HIC_{15}. Equation (2) represents HIC:

$$HIC = (t - t_0)\left[\left(\frac{1}{t - t_0}\right) \int_{t_0}^{t} a(t)dt\right]^{2.5} \tag{2}$$

In general, suggested threshold values for HIC_{36} and HIC_{15} are 1000 and 700, respectively, for 50th percentile males. Exceeding those values under their associated conditions represents the risk of severe, life-threatening head injury.

Head Impact Power is a criterion that is based on linear and rotational acceleration of the head during impact and on impact duration. The equation (3) for HIP is formulated by using inertial characteristic data of the head along with the equation for the rate of change of kinetic energy for any rigid object:

$$Power = P = \sum m\bar{a} \cdot \bar{v} + \sum I\bar{\alpha} \cdot \bar{\omega} \tag{3}$$

With the characteristic data inputted into the equation, HIP comes out to be the following equation (4):

$$HIP = 4.50a_x \int a_x \, dt + 4.50a_y \int a_y \, dt + 4.50a_z \int a_z \, dt + 0.016\alpha_x \int \alpha_x \, dt + 0.024\alpha_y \int \alpha_y \, dt + 0.022\alpha_z \int \alpha_z \, dt \tag{4}$$

Studies have found that MTBI correlates better with HIP than many other head injury criteria. As such, HIP allows for the evaluation of headgear in the prevention of concussions [48].

RIC and PRHIC are a more recently developed set of criteria. RIC uses the same equation form of HIC but substitutes peak linear acceleration with peak angular acceleration. Since it is similar to HIC, RIC is also able to assign time durations to the impact conditions. It has been observed that RIC has no significant correlation to max angular velocity, and as such this set of criteria is better suited for analyzing mild traumatic brain injury (MTBI). PRHIC on the other hand, utilizes the same equation form of HIC and substitutes the rotational component of HIP for the peak linear acceleration. PRHIC acts as a new predictor for head injuries associated with angular accelerations from six degree-of-freedom device data at the center of gravity in the head.

Significant correlations have been observed with max angular velocity for this criterion, and as such it works best with analyzing severe traumatic brain injury (TBI) [50].

1.3.3.1 INJURY RISK CURVES

The protective capacity of headgear is evaluated within the context of injury risk curves (Figure 8). Injury risk curves are created using logistic regression on impact data where clinically-verified concussions and non-concussions are known. These curves are used in determining the risk of an individual sustaining a head injury from given impact events. Based on the individual's protective condition (e.g. unprotected, helmeted), outcome measurements and their associated risk can vary. For example, a helmeted individual may experience 50 g's of linear acceleration as opposed to 120 g's when unprotected under the same impact mechanisms.

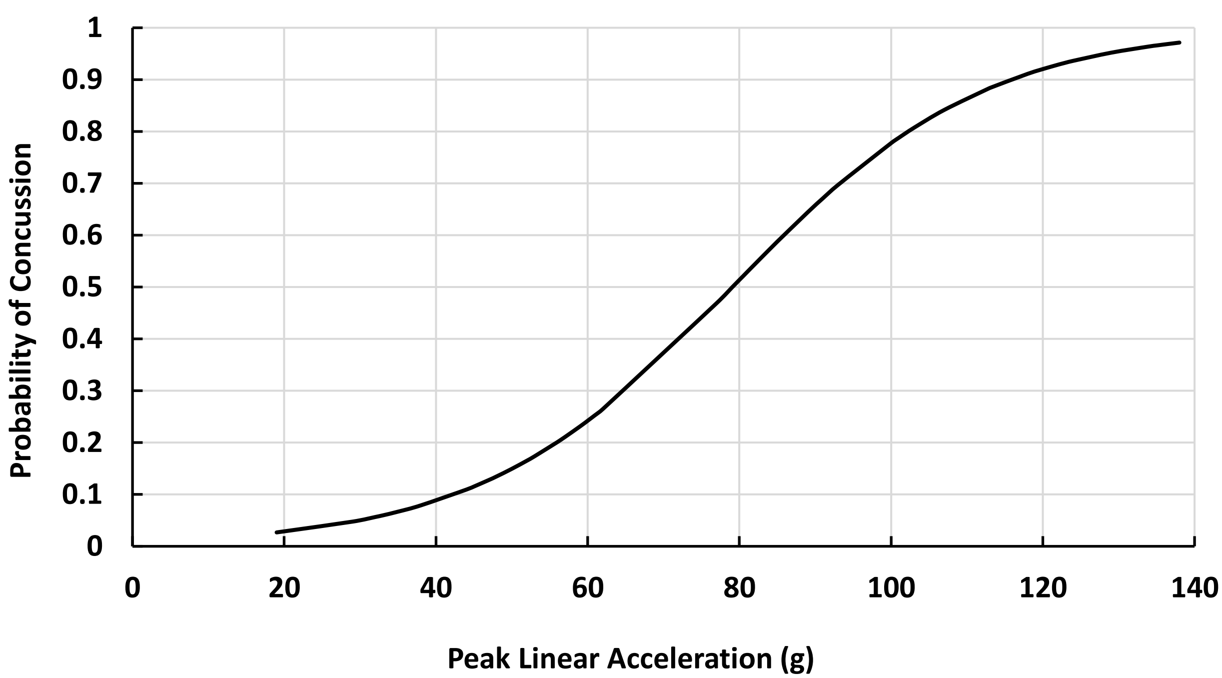


Figure 8: Example injury risk curve from Pellman et al. [54]

A limiting factor in the applicability of any injury risk curve is the foundational data on which the injury risk curve was built. For example, almost all injury risk curves that exist currently in a sports context are based on real-time, or experimentally reconstructed, impacts from football players [54, 55, 56]. As such, impact events that are substantially different from the underlying foundational events may suffer from inaccuracies when predicting the probability of injury.

CHAPTER 2 - EPIDEMIOLOGY OF SOFTBALL HEAD AND FACIAL INJURIES

2.1 INTRODUCTION

A review of the current epidemiological literature on softball-related head injuries reveals that only one study had investigated concussive injuries [54], and no information on the broader spectrum of head and facial injuries was available. The purpose of this epidemiological study was to investigate emergency department (ED) injury data related to softball impacts in order to identify trends regarding the nature and frequency of head and facial injuries sustained during softball play. More specifically, the objective was to evaluate nation-wide emergency department data to determine the statistical distribution of diagnosed head/face injuries sustained during softball play and to identify specific causal mechanisms associated with the injuries. Such data could be helpful during initial triage and/or treatment of injured players, and may also serve to promote broader use of protective headgear for participating players.

2.2 METHODS

The National Electronic Injury Surveillance System (NEISS) was accessed to obtain data regarding head and facial injuries sustained in softball over a five-year timespan (2013-2017). NEISS collects data from approximately 100 hospitals nationwide as a probability sample of all the 5,000+ hospitals in the U.S. with emergency departments. Data collected for each case include the patient's age, sex, race/ethnicity, injury diagnosis, affected body parts, disposition, incident location, and a brief narrative description of the incident. NEISS uses an extrapolating algorithm, based on sample weights and national probability sample, to produce national estimates. The overall system that the Consumer Product Safety Commission (CPSC) employs in NEISS has been well established for many years and is widely used by researchers and government agencies [55, 56, 57].

The database was accessed in May 2018 and queried using the product code "softball" (5034). Codes for the following body parts were used to refine the query such that it was limited to head and facial injuries: head, face, eyeball, mouth, and ear. The narrative description for each entry was further screened to provide insight on use of protective equipment and to determine injury mechanism. Four categories were used to classify protective equipment usage – helmeted (if the narrative specifically identified the player as wearing a helmet or mask at the time of incident), un-helmeted (if the player was specifically identified as not wearing protective headgear), unknown, and equivocal. After initially reviewing the data, six categories of injury mechanism were created, which included: struck by ball, struck by bat, collision with another player, collision with the ground or a fixed object, other (events such as heat exhaustion or a foreign body in the eye), and equivocal/not specified. Sub-categories within the struck-by-ball mechanism group were further created since being struck by a ball proved to be the most common

injury mechanism and many narratives provided additional detail on the contextual setting of the ball strike. These sub-categories included: hit by pitch (all individuals struck by a pitched ball, including batters and catchers), base runner (any offensive player struck by a hit or thrown ball during the act of running bases), defender (fielder) struck by batted ball, defender struck by thrown ball, equivocal defensive play (players identified as being struck in the act of attempting to catch a ball as part of defensive play, but where it was unknown if the ball was hit or thrown), other (for events not involving offensive or defensive play, such as players in the dugout, base coaching, etc.), and equivocal/not specified. Statistical analysis of the data was performed using Stata statistical software (Stata v10, 2007; StataCorp, College Station, TX). This included standard descriptive statistics, Pearson's chi-squared comparisons for categorical variables, Student's T-tests, and analysis of variance (ANOVA) for continuous variables.

To provide insight on whether specific aspects of gameplay were related to particular injury types, the same statistical analyses were performed for sub-categories of injury diagnoses. These sub-categories were limited to injuries that were both frequent (greater than 3% of diagnosed injuries) and relatively severe (contusions and lacerations were excluded). This ultimately resulted in two sub-categories being analyzed – patients with a fracture diagnosis and patients with a head/brain injury diagnosis (where closed head injury (CHI) and concussion diagnoses were grouped together).

Finally, in that the NEISS database uses two separate diagnosis categories for head injuries, those distinctions were maintained within the present analysis. Specifically, when coding events in NEISS, clinicians are instructed to code the most severe and specific diagnosis. If internal head injuries, such as subdural hematomas or cerebral contusions are documented, then a diagnosis of "internal organ injury" is entered (which we use interchangeably with the phrase "brain injury").

Otherwise, if only concussive symptoms are observed, then a diagnosis of "concussion" was to be used.

2.3 RESULTS

During the five-year span, a total of 3,324 injuries involving the face and head were recorded in the NEISS database. Through an algorithm that the NEISS database uses, the weighted annual estimate of these injuries within the United States was calculated to be 121,802. The average age of the injured player was 21.5 years (S.D. = 14.4). Females accounted for 72.1% of the injuries, while males were 27.9%. Regarding disposition, 96.3% of patients were discharged to home from the ED, whereas 2.1% were admitted and 0.5% were transferred to another facility (Table 1).

Table 1: Summary of patient demographics and dispositions.

	Total (n = 3324)	Concussions (n = 589)	Brain Injuries (n = 730)	Fractures (n = 501)
Age	21.5 ± 14.4	18.4 ± 8.7	21.6 ± 16.5	25 ± 14.1
Sex				
Female	2397 (72)	495 (84)	537 (74)	308 (61)
Male	927 (28)	94 (16)	193 (26)	193 (39)
Race				
White	1864	341	423	265
Black	230	29	27	35
Other/Unspecified	1230	219	280	201
Disposition				
Home	3201	576	695	460
Admission/Observation	70	6	23	12
Other/Unspecified	53	7	12	29

The head (45.0%) and face (42.4%) were the most commonly injured body parts, although an appreciable number of mouth (8.8%) and eye (3.0%) injuries were also reported. The most common injury diagnosis was "internal organ" (22.0%; refers to subdural hematomas, cerebral

contusions, and other head injuries that would be considered severe), followed by contusions (18.7%), concussion (17.7%), lacerations (17.1%), fracture (15.1%) and dental injury (2.4%). Other injuries, such as hematoma, nerve damage, puncture, hemorrhage, and avulsion occurred in much smaller frequency (typically much less than 1.0%) (Figure 9).

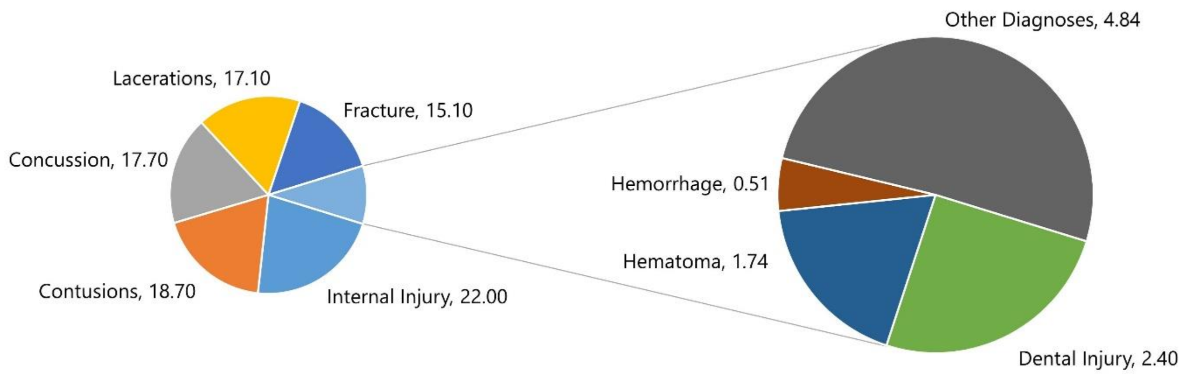


Figure 9: Percentage breakdown of injury diagnoses. The category labeled "Other Diagnoses" includes injury diagnoses that were unspecified or diagnoses with less than 3 documented cases.

The use of a helmet or other protective gear was unknown in 98.3% of cases; only 34 instances (1.0%) of helmet use were recorded, and 12 instances (0.4%) of non-use were recorded. In the cases where helmet use was recorded, females were significantly more likely to have worn a helmet/mask than males (1.3% vs. 0.2%, p=0.002). The most common injury mechanism (Table 2) involved players being struck by a ball (74.3%), followed by collision with another player (8.3%), collision with the ground or a fixed object (5.0%), and being struck by a bat (2.8%). Within the struck-by-ball injury group, most narratives (73.5%) failed to provide sufficient information to understand the context of injury. However, for those that described the aspect of play at the time of ball strike, 83.7% came from defensive play, while 12.3% came during offensive play.

Table 2: Injury mechanisms for all head/facial injuries sustained in softball, and for the sub-category of struck-by-ball events where the precipitating event was described in the narrative.

Injury Mechanism (%)					
Struck by Ball	Struck by Bat	Collision w/ Player	Collision w/ Ground or Object	Equivocal/ Unspecified	Other
74.3	2.8	8.3	5.0	9.7	1.4
Injury Mechanism (%) – Sub-category for Known "Struck by Ball" Event					
Hit by Pitch	Base Runner	Hit by Batted Ball	Hit by Thrown Ball	Equivocal Defense	Other
9.3	3.0	36.8	27.4	19.6	3.9

The sub-group consisting of concussion and CHI diagnoses represented nearly 40% of all reported face/head injuries, with a weighted estimate of 46,056 occurring annually. Disposition statistics were similar to the overall face/head distribution (no metric varied by more than 0.4%). Within this class of injuries, the mean age of an injured player was slightly less (19.8 ± 11.4) and females represented a greater percentage of injuries (78.3%) than in the general data set. Helmet usage was recorded slightly more frequently in this subset – 2.7% of injured players were wearing a helmet, whereas 1.6% were not. Injury mechanisms displayed a small shift away from struck-by-ball injuries (65.9%) to collisions with other players (13.3%) and collisions with the ground or a fixed object (8.4%) (Figure 10) (Table 3). Within the struck-by-ball injury mechanism group with known causal events, 79.2% came from defensive play while 16.8% came from offensive play.

Table 3: Injury mechanisms for CHI and concussion diagnoses, and for the sub-category of struck-by-ball events where the precipitating event was described in the narrative.

Injury Mechanism (%)					
Struck by Ball	Struck by Bat	Collision w/ Player	Collision w/ Ground or Object	Equivocal/ Unspecified	Other
65.9	2.4	13.3	8.4	9.2	0.8
Injury Mechanism (%) – Sub-category for Known "Struck by Ball" Event					
Hit by Pitch	Base Runner	Hit by Batted Ball	Hit by Thrown Ball	Equivocal Defense	Other
10.4	6.4	37.2	30.0	12.0	4.0

The sub-group of fracture diagnoses represented nearly 15.2% of all reported face/head injuries with a weighted estimate of 18,127 occurring annually. Demographic factors were not predictive of fractures. Injury mechanisms appreciably shifted towards struck-by-ball injuries (84.2%), with all other categories decreasing relative to the overall injury group (Table 4). Within the struck-by-ball injury mechanism group with known causal events, 87.5% came from defensive play while 9.4% came from offensive play.

Table 4: Injury mechanisms for fracture diagnoses, and for the sub-category of struck-by-ball events where the precipitating event was described in the narrative.

Injury Mechanism (%)					
Struck by Ball	Struck by Bat	Collision w/ Player	Collision w/ Ground or Object	Equivocal/ Unspecified	Other
84.2	1.6	5.4	1.0	7.8	0.0
Injury Mechanism (%) – Sub-category for Known "Struck by Ball" Event					
Hit by Pitch	Base Runner	Hit by Batted Ball	Hit by Thrown Ball	Equivocal Defense	Other
7.8	1.6	53.1	18.8	15.6	3.1

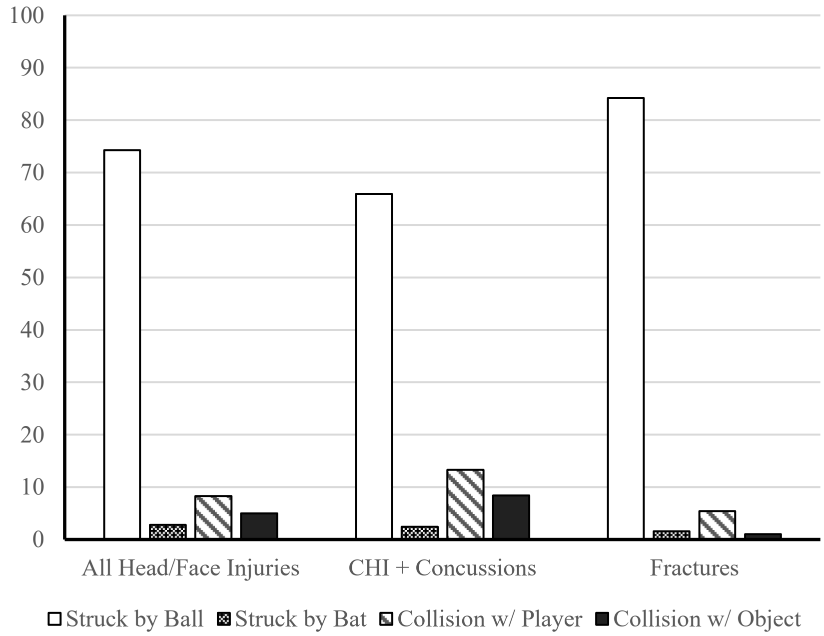


Figure 10: Distribution of injury mechanisms across diagnosis categories.

2.4 DISCUSSION

This study is novel in that it presents data relating to the entire spectrum of head and face injuries sustained across all softball play, and analyzes injury mechanisms in detail. Overall, it is estimated that emergency departments in the United States see 121,802 softball-related head and facial injuries annually. Querying the NEISS database with the same search parameters for different sports reveals that more head and facial injuries occur annually in softball than in ice hockey (35,333) and lacrosse (23,174) combined. In fact, the number of annual softball injuries is on the same order of magnitude as football (396,276) and baseball (255,250) – two sports that traditionally garner a great deal of academic and medical attention. Although the absolute rates at which such injuries occur within each of these sports are unknown (since the total participation

numbers are not known), these data nonetheless indicate that further research into softball injury mechanisms and prevention may be warranted.

With regards to previous studies, we are aware of only one that has examined the epidemiology of softball injuries, however, it only examined concussion diagnoses and was interpreted within the context of fast-pitch versus slow-pitch softball [54]. Although females represented the majority of injuries, we observed an appreciable number of males in the database (accounting for nearly 25% of the injuries), most likely reflective of recreational or slow-pitch play. Most patients were also relatively young (with the average being about 22 years-old), but extremes on each end of the spectrum were observed – the oldest player was 79 and the youngest was five. As such, medical complications associated with treating very old or young patients may be encountered when dealing with softball-related injuries.

Almost all injuries in the database could be classified into one of four categories of causal mechanism: being struck by a ball, struck by a bat, colliding with another player, or colliding with the ground or a fixed object. Within these groups, the overwhelming majority of injuries occurred from being struck by a ball (approximately three times more common than all other mechanisms combined), which is consistent with previous published findings [54]. Based on this observation, we chose to establish sub-categories within the struck-by-ball group, providing unique data on the specific elements of gameplay responsible for such injuries. Several aspects of defensive play were identified as being responsible for the majority of ball impacts, with batted balls representing the plurality. Within the narratives describing defensive players struck by a batted ball, many instances of “line drives” striking pitchers and infielders were encountered, as would be expected. However, descriptions of “bad hops” and players being struck by “fly balls” were also common, indicating that all defensive positions are vulnerable to such impacts. The susceptibility of all defensive

player positions was further confirmed by the appreciable number of injuries caused by thrown balls and missed catches. Finally, although we note that our classification system incorrectly grouped catchers into offensive play, this was done since both the element of gameplay and the mechanism of injury is the same for catchers as for offensive players struck by pitches. As such, the true proportion of defensive injuries is slightly higher than reported here.

The sub-group of concussion and CHI diagnoses represented a large proportion of injuries in the database. Although a concussion diagnosis is reflective of neurological/cognitive dysfunction (i.e. symptom-based) and CHI reflects a non-penetrating head injury (i.e. objective observation), we chose to group these diagnoses together since they are both generally reflective of intracranial trauma and often coexist [32]. While nearly all of these diagnoses resulted in patients being discharged to home, we caution that the NEISS database does not typically contain information on follow-ups and the ultimate course of these injuries is not known. As noted previously, isolated case studies have demonstrated that the course of CHIs from softball impacts can be severe and unfold over the span of days [4], and therefore cautious monitoring and follow-up evaluation of head injuries may be appropriate if persistent symptoms are observed. The frequency of observed CHI and concussion from softball impacts raises additional concerns as seen in other sports with high risk of traumatic brain injuries. For example, in football, those with a history of concussions had an increased chance of sustaining another one compared to those with no history [58]. Cumulative softball-related head injuries may also pose risk for long-term neurodegenerative diseases, such as chronic traumatic encephalopathy (CTE), or other forms of cognitive disorder as seen in observations following repetitive head impacts from football [59]. Therefore, adherence to return-to-play guidelines may be appropriate for both organized and recreational softball play.

Unfortunately, helmet usage was poorly tracked in this data set, and few (if any) conclusions can be directly reached regarding adoption rates or associated efficacy in injury prevention. Nonetheless, most fractures observed in this study were caused by ball strikes experienced during defensive play. The types of masks and protective headgear available to fielders and pitchers are typically designed to prevent contact injuries to the face from a softball [8], but are rarely mandated to be worn. In contrast, protective headgear is required much more often in offensive play and for catchers, particularly so in the case of competitive play (as compared to recreational leagues). This mandated use may partially explain why our study documented relatively few face/head injuries sustained from ball strikes during offensive play. The present data also suggests some areas of helmet testing and design that could benefit from future research. For instance, CHI and concussion represented about 40% of all injuries in the data set, the majority of which were caused by ball strikes in defensive play. However, fielder's masks are less substantial in their construction and padding as compared to batter's helmets and catcher's masks [9, 10, 60, 61], and, to our knowledge, the ability of fielder's masks to attenuate concussive accelerations is not well established. This remains an avenue of future research that should be pursued.

A few limitations were encountered in this study that warrant discussion. First, the data collected by NEISS is solely from patients treated at EDs, and therefore does not account for injuries that were left untreated or those that were treated through means other than the ED. This may include diagnosis and/or treatment by coaches, on-site athletic trainers/physicians, or the patient visiting urgent care clinics or out-patient doctors. In particular, athletes participating in the upper levels of competitive play (e.g., high school and collegiate) likely receive medical care through their trainers/team doctors, and therefore could represent a substantial source of

unaccounted injuries. In addition, several studies have observed that head injuries, particularly traumatic brain injuries, are often left untreated or their diagnosis is missed [1, 62]. What's more, NEISS suffers from imprecise diagnosis codes, and non-expert coders may not understand the difference or importance of the potentially life-threatening "subdural hematoma" versus a superficial "subgaleal hematoma." Given the relative paucity of hospital admissions despite high numbers of reportedly more severe brain injuries in our sample, this was a definite limitation of the data set. Second, the narrative descriptions provided in the NEISS database often lacked sufficient details to fully understand the circumstances under which the injury occurred, thus limiting the statistical power of the analysis. This was specifically encountered in our attempt to quantify helmet/mask usage in the present data set. Third, several of the diagnoses provided by the data set, such as "concussion" and "closed head injury" are not radiology-based or standardized clinical-exam based diagnoses. Despite this, limited conclusions could still be made based on our detailed categorization of injury mechanisms.

CHAPTER 3 – EXPERIMENTAL EVALUATION OF SOFTBALL HEADGEAR FOR DEFENSIVE PLAY

3.1 INTRODUCTION

Acceleration data from studies examining catcher's masks are unlikely to be applicable to fielder's masks, since the structure and composition of these two types of headgear differ from each other (Figure 11). Catcher's masks are typically made with steel frames and have thick padding material (in excess of 1", often) surrounding the face and throat. In contrast, fielder's masks are generally constructed with a minimum amount of material to preserve light-weight mobility and field of vision, while still maintaining enough protective structure to prohibit facial contact from a softball. They are typically constructed using either steel or hard plastic frames along with foam padding at the forehead and chin regions. The padding is often thin (on order of ½") and more compliant than in catcher's masks. Owing to the differences in construction, it is hypothesized that catcher's masks represent a bound in the achievable performance of fielder's masks. However, as discussed above, fielder's masks are much less substantial in their construction and may or may not be capable of reducing concussive risk like catcher's masks would from softball impacts.

Figure 11: Examples of defensive protective headgear for softball. (Left to Right: Hockey-style catcher's mask, Steel-frame fielder's mask, Plastic-frame fielder's mask)

Within this context, the main objective was to evaluate the impact performance of defensive fielder's masks with regards to inertial protection, as well as facial protection (i.e. preventing contact with sensitive areas of the head/face) when subjected to softball impacts. Further, the study aimed to compare the performance of various fielder's mask designs relative to other protective conditions, including unprotected (no headgear/facemask, as a "baseline") and a hockey-style catcher's mask.

3.2 METHODS

The overall test assembly consisted of a Hybrid III 50th percentile male head/neck assembly mounted on a set of low-friction, linear rails. A pneumatic cannon, consisting of a polyvinyl chloride (PVC) barrel, direct-acting solenoid, and a 5-gallon high-pressure air tank, was used to launch softballs at an instrumented headform (Figure 12). Regulation 12-inch softballs (mass = 0.193 kg, COR = 0.44) were used for all impacts (Newell Brands - Worth Sporting Goods, Hoboken, NJ).

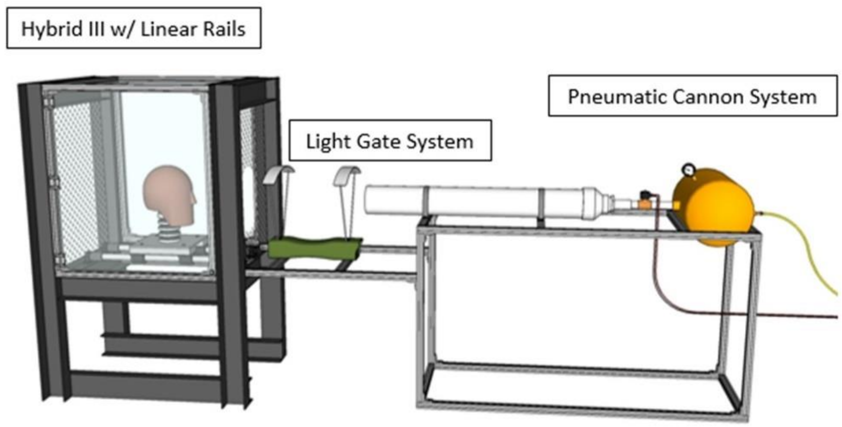


Figure 12: Experimental test set-up used for softball impacts to the Hybrid III head assembly.

Softball launch speeds were chosen to simulate conditions of an overhand thrown ball and batted ball. Average overhand throwing speeds and batted ball velocities for high school and collegiate athletes range from 48 to 55 mph and 55 to 70 mph, respectively [63, 64]. Accordingly, a speed of 60 mph was chosen to simulate batted balls since it falls within the 55-70 mph range and also matches the condition of the National Operating Committee on Standard Athletic Equipment (NOCSAE) tests for softball headgear [8, 51]. For thrown balls, we note that the published speeds are specified for maximum overhand throwing into a net – typically used by coaches and other scouts to evaluate arm strength [63, 64]. Slower throw speeds are likely to occur in game play, so we therefore chose to reduce the average range by approximately 20%, to 40 mph.

Impact locations on the headform were selected based on landmark features present in the headgear (i.e. chin/forehead foam padding, nose guard, etc.) that also corresponded to notable facial landmarks. Specifically, four impact locations were chosen: chin, nose, forehead, and temple. The chin, nose, and forehead impacts were oriented with the headform facing the barrel exit, such that direct anterior-to-posterior impacts would be achieved. Nose impacts were positioned with the center of the ball striking the tip of the nose, while chin impacts were 2.5”

below the nose, and forehead impacts 3” above the nose. Impacts to the temple region were performed with the Hybrid III assembly rotated 90 degrees to achieve direct lateral-to-medial impact conditions. Temple impacts were 3” above the nose and 1.5” posterior to the lateral orbit feature.

The protective headgear that were tested included six fielder’s masks and one catcher’s mask (Table 5). Unprotected impacts, which included the headform wearing no headgear, were conducted as well. Since fielder’s masks were the primary headgear of interest, several different brands were acquired based on variations in design features, such as frame material, mask geometry, and thickness/arrangement of foam padding. Overall, each protective condition (i.e. unprotected, catcher’s mask, etc.) was tested N = 3 times for each impact location and speed. However, one mask type – the ABS fielder’s mask – could not be tested with replication because it sustained extensive damage during several impact conditions and went out of production after this was discovered. The total of number tests conducted was 176.

Table 5: Description of protective headgear used for softball impact testing.

Type	Brand	Model	Size	NOCSAE Certified	Frame Material
Catcher – Hockey Style	Mizuno	G4 Samurai	$7\frac{1}{4}$ - $7\frac{5}{8}$	Yes	Steel
Fielder	Rawlings	RFACE1	OSFM	No	Polycarbonate
Fielder	Markwort (Gameface)	LGFSK	OSFM	No	Polycarbonate
Fielder	Bangerz	HS1800	OSFM	No	ABS
Fielder	Rip-It	RIPDG-A-B	OSFM	No	Steel
Fielder	Mizuno	MFF900	OSFM	No	Steel
Fielder	Schutt	SKU 1221	OSFM	No	Steel

OSFM = One-Size-Fits-Most

Preliminary tests were conducted to ensure that softballs were launching at the desired low and high speeds. The two speeds were verified using a combination of high-speed video (Sony Digital Camera DSC-RX10M3, New York, NY) and light gates (Caldwell Ballistic Precision Chronograph, Columbia, MO), which were capable of measuring speed to 0.25% accuracy. During actual tests, the high-speed video, which collected at 960 frames per second, was further used to ensure impact location and to observe if any facial contact occurred. Ball speeds were measured for each test by using the light gate system. Across N = 50 calibration tests, as well as the N = 176 experiments, the average ($\pm$SD) velocity of low-speed launches was 39.7 ± 0.9 mph, while high-speed tests were 59.8 ± 1.1 mph.

All equipment was inspected for damage after each test. Softballs were checked after every launch for visible damage. If any damage was present, then the softball would be exchanged for a new one. All protective headgear was checked after each test. Any visible damage disqualified headgear from subsequent testing. However, if localized damage was very minor (e.g., a 2-mm dent at chin frame) and unlikely to affect mask performance for impacts at other regions, then it would be allowed for further testing at those unaffected regions (e.g. forehead impact).

The Hybrid III was equipped with three linear accelerometers and three angular rate sensors (DTS 6DX PRO 2K-18K, Seal Beach, CA), all of which were positioned at the center of gravity in the headform. The data acquisition system used a SLICE MICRO (DTS, Seal Beach, CA) to sample acceleration data at 20 kHz with 4 kHz anti-alias filtering. Data was exported in unfiltered form, and all data were post-processed through a custom Matlab program (The Mathworks, Inc, Natick, MA). The program filtered data in accordance with SAE J211 (Instrumentation for Impact Tests); angular accelerations and other injury metrics were calculated in accordance with SAE J1727 (Calculation Guidelines for Impact Testing). Outcome variables available through the post-

processing routine included: peak linear acceleration (PLA), peak angular acceleration (PAA), impact duration (Δt), Head Injury Criterion (HIC) [47], Rotational Injury Criterion (RIC) [50], and Severity Index (SI) [46].

Statistical analyses were carried out to quantify the absolute and relative performance of the protective headgear. For each impact location and speed, the mean and standard deviation were calculated for each outcome variable. Analysis of variance (ANOVA) was conducted for each protective condition (i.e. speed and impact location) with mask type being the factor; post-hoc Bonferroni tests were performed if significance ($p < 0.05$) was detected in the overall ANOVA.

3.3 RESULTS

In general, fielder's masks reduced acceleration metrics relative to the unprotected condition, but not to the same degree as observed in the catcher's mask (Table 6, Table 7). On average, fielder's masks reduced peak linear and angular accelerations during low-speed impacts by 36-49% and 14-45%, respectively, while for high-speed impacts they were reduced by 25-42% and 13-46%, respectively. In comparison, the catcher's mask reduced most inertial metrics by more than 80% relative to the unprotected condition. However, for a few select test conditions and outcome metrics, fielder's masks either failed to outperform the unprotected condition or were able to perform as well as the catcher's mask. Specifically, most fielder's masks did not significantly reduce peak angular accelerations at the forehead and temple ($p > 0.05$) as compared to the unprotected condition (for either launch speed). On the other hand, no statistical difference ($p > 0.05$) was detected in peak accelerations between the fielder's masks and the catcher's mask for four conditions: peak linear acceleration for low-speed forehead impacts ($p = 0.13$), peak

angular acceleration for low-speed nose impacts ($p = 0.30$; $p = 0.72$) and chin impacts ($p = 0.62$), and peak angular acceleration for high-speed forehead impacts ($p = 0.14$).

Some fielder's masks performed notably different relative to other protective conditions, and such effects tended to be manifested uniquely based on test conditions (i.e. anatomic location and impact velocity). For example, the Schutt reduced peak angular acceleration from high-speed forehead impacts comparable to level of the catcher's mask ($p = 0.14$) and was significantly less than three other fielder's masks ($p < 0.05$). In contrast, for the same test condition, the Mizuno performed similar to the unprotected condition ($p = 1.02$) and was significantly higher than three other fielder's masks ($p < 0.05$). During high-speed temple impacts, the Gameface and Rawlings exhibited peak linear accelerations that were not significantly different from the unprotected condition ($p = 1.22$ and $p = 1.09$, respectively), and were significantly higher than the other three fielder's masks ($p < 0.01$).

Table 6: Acceleration-based outcome metrics, expressed as mean (±SD), for low-speed impacts.

		Unpr.	Catcher's	Mizuno	Rip-It	Schutt	Gameface	Rawlings	Fielder's Average
NOSE	PLA	62.3 (3.1)	19.7 (0.8)	35.0 (2.5)	40.4 (2.9)	41.4 (3.2)	31.9* (1.8)	44.5 (1.4)	38.6 (5.1)
	PAA	2648 (134)	941 (270)	1430‡ (314)	1789 (204)	2103† (326)	1524‡ (296)	1833 (182)	1736 (337)
	SI	25.7 (2.1)	4.5 (1.0)	9.1‡ (0.9)	8.0‡ (1.4)	14.4 (2.9)	6.9‡* (0.8)	10.6 (0.8)	9.8 (3.0)
	HIC15	22.1 (1.7)	3.8 (1.0)	7.5‡ (0.7)	7.0‡ (0.9)	12.1* (2.7)	5.9‡ (0.7)	8.5 (0.7)	8.2 (2.5)
	RIC36 (x 10^6)	1.30 (0.14)	0.15 (0.08)	0.28‡ (0.10)	0.47 (0.05)	0.36‡ (0.09)	0.47 (0.15)	0.51 (0.05)	0.42 (0.12)
FOREHEAD	PLA	98.6 (14.5)	16.5 (2.1)	58.0 (2.0)	32.3‡ (4.3)	39.8 (2.6)	46.7 (1.4)	76.8* (2.9)	50.7 (16.3)
	PAA	2532 (230)	638 (124)	2727† (247)	2009† (379)	1451* (75)	2260† (225)	2464† (404)	2182 (513)
	SI	62.4 (13.1)	2.5 (0.7)	23.1 (2.7)	10.0‡ (1.2)	8.5‡ (1.6)	18.2 (0.4)	38.2* (1.8)	19.6 (11.2)
	HIC15	54.8 (10.8)	2.1 (0.6)	19.5 (2.5)	8.1‡ (1.0)	6.8‡ (1.7)	14.9‡ (0.2)	33.5* (1.7)	16.6 (10.1)
	RIC36 (x 10^6)	0.49 (0.11)	0.07 (0.03)	0.65† (0.12)	0.37†‡ (0.14)	0.15‡ (0.01)	0.46† (0.11)	0.50† (0.21)	0.43 (0.21)
TEMPLE	PLA	96.8 (2.7)	24.0 (2.2)	44.4* (3.8)	59.2 (8.2)	50.3 (4.7)	77.0* (1.4)	79.2* (5.4)	62.0 (15.1)
	PAA	3971 (164)	1605 (106)	2705 (145)	3130 (145)	2842 (47)	3474† (225)	3334† (455)	3097 (364)
	SI	62.6 (3.0)	3.7 (0.7)	14.9‡ (3.2)	24.2 (5.9)	16.0 (1.9)	41.3* (1.4)	44.6* (6.6)	28.2 (13.5)
	HIC15	55.7 (3.0)	3.2 (0.6)	13.4‡ (2.9)	21.1 (5.1)	14.3 (1.8)	37.1* (1.4)	39.5* (6.5)	25.1 (12.0)
	RIC36 (x 10^6)	1.51 (0.16)	0.25 (0.04)	0.66‡ (0.09)	0.87 (0.12)	0.72‡ (0.01)	1.12† (0.19)	1.01† (0.34)	0.88 (0.24)
CHIN	PLA	84.2 (2.9)	12.3 (2.5)	42.9 (1.0)	46.3 (4.0)	34.2* (3.5)	40.0 (1.8)	50.9 (1.7)	42.9 (6.3)
	PAA	5163 (113)	1371 (165)	3478 (203)	2123* (243)	1799‡* (349)	2998 (191)	3749* (168)	2829 (808)
	SI	40.3 (2.7)	1.5 (0.6)	12.4 (1.0)	9.5 (2.1)	6.4‡* (2.4)	11.6 (0.9)	17.5* (0.8)	11.5 (4.0)
	HIC15	35.1 (2.5)	1.4 (0.5)	9.7 (0.5)	7.7 (1.8)	5.3‡ (2.1)	9.3 (0.6)	15.3* (0.9)	9.5 (3.6)
	RIC36 (x 10^6)	3.98 (0.25)	0.33 (0.14)	2.08 (0.21)	0.99‡ (0.17)	0.58‡* (0.26)	1.6 (0.20)	2.08 (0.37)	1.47 (0.65)

† = Not significantly different from Unprotected ($p > 0.05$); ‡ = Not significantly different from Catcher's Mask ($p > 0.05$); * = Significantly different from 3+ fielder's masks ($p < 0.05$)

Table 7: Acceleration-based outcome metrics, expressed as mean (±SD), for high-speed impacts.

		Unpr.	Catcher's	Mizuno	Rip-It	Schutt	Gameface	Rawlings	Fielder's Average
NOSE	PLA	119.3 (7.7)	33.4 (2.1)	76.2 (5.7)	72.3 (2.6)	61.9 (1.5)	57.6* (1.6)	75.5 (0.8)	68.7 (8.2)
	PAA	4362 (203)	1137 (130)	1861 (203)	2288 (290)	2491 (175)	2833 (242)	2313 (22)	2357 (370)
	SI	102.5 (11.9)	17.6 (1.8)	43.8 (4.5)	32.5‡ (3.0)	40.9 (2.2)	24.5$^{\ddagger *}$ (0.9)	49.1 (1.4)	38.2 (9.3)
	HIC15	88.4 (11.3)	15.4 (1.6)	36.0 (3.6)	26.3‡ (1.5)	36.3 (2.2)	19.9$^{\ddagger *}$ (0.8)	43.4 (1.4)	32.4 (8.7)
	RIC36 (x 10^6)	3.61 (0.41)	0.30 (0.07)	0.61‡ (0.09)	1.40 (0.28)	0.67‡ (0.20)	1.75 (0.41)	1.24 (0.06)	1.14 (0.49)
FOREHEAD	PLA	173.2 (13.0)	25.0 (0.6)	121.2 (1.8)	89.8 (2.0)	84.5 (2.5)	106.8 (13.2)	134.4* (10.1)	107.3 (20.4)
	PAA	4005 (393)	1472 (273)	4586$^{\dagger *}$ (134)	3169† (258)	2329$^{\ddagger *}$ (396)	3417† (365)	3895† (396)	3479 (825)
	SI	208.6 (22.9)	10.3 (0.6)	112.5 (2.4)	65.6 (3.1)	65.6 (1.6)	89.8 (13.8)	141.7* (16.4)	95.1 (31.3)
	HIC15	173.1 (11.8)	8.8 (0.6)	98.0 (1.9)	55.2 (2.8)	58.4 (2.1)	79.0 (11.0)	126.2* (13.9)	83.4 (28.2)
	RIC36 (x 10^6)	1.57 (0.43)	0.19 (0.05)	2.20$^{\dagger *}$ (0.12)	0.94† (0.18)	0.45‡ (0.16)	1.08† (0.26)	1.45† (0.36)	1.23 (0.64)
TEMPLE	PLA	146.3 (10.9)	51.5 (12.6)	94.1 (1.1)	95.3 (5.4)	99.3 (9.5)	131.8$^{\dagger *}$ (7.6)	131.3$^{\dagger *}$ (8.1)	110.4 (18.9)
	PAA	6079 (184)	2784 (596)	4581 (345)	5377† (218)	4657 (199)	5687† (501)	5471† (668)	5155 (588)
	SI	174.7 (27.9)	19.3 (7.1)	74.1 (6.7)	73.4 (13.4)	69.2 (10.6)	137.2$^{\dagger *}$ (12.0)	143.9$^{\dagger *}$ (12.1)	99.6 (36.0)
	HIC15	156.7 (25.8)	16.1 (5.9)	66.9 (7.6)	65.7 (13.6)	62.0 (9.7)	124.2$^{\dagger *}$ (11.0)	130.6$^{\dagger *}$ (10.6)	89.9 (33.1)
	RIC36 (x 10^6)	4.35 (0.35)	0.79 (0.25)	2.22‡ (0.42)	3.33† (0.27)	2.33‡ (0.24)	3.70† (0.82)	3.34† (0.93)	2.98 (0.80)
CHIN	PLA	127.5 (13.6)	25.3 (0.3)	83.4 (7.1)	91.3 (7.0)	72.0 (3.4)	86.9 (5.3)	95.5 (0.6)	85.8 (9.4)
	PAA	7162 (223)	1978 (123)	6278† (809)	4411 (256)	3304* (447)	5345 (118)	6068† (38)	5081 (1199)
	SI	117.0 (15.7)	7.2 (0.7)	55.4 (7.5)	55.2 (8.1)	40.4 (0.8)	52.1 (5.3)	67.9 (4.9)	54.2 (10.3)
	HIC15	102.7 (12.4)	6.1 (0.8)	47.5 (7.0)	46.1 (7.4)	33.3 (1.8)	45.0 (4.8)	58.4 (4.6)	46.1 (9.5)
	RIC36 (x 10^6)	9.05 (0.22)	0.98 (0.14)	7.76$^{\dagger *}$ (2.11)	4.25 (0.44)	2.06$^{\ddagger *}$ (0.37)	5.20 (0.28)	7.20† (0.25)	5.30 (2.29)

† = Not significantly different from Unprotected ($p > 0.05$); ‡ = Not significantly different from Catcher's Mask ($p > 0.05$); * = Significantly different from 3+ fielder's masks ($p < 0.05$)

With regards to mask damage, only a few masks were damaged in the low-speed impacts, but most experienced some form of damage from high-speed impacts (Table 8). The most common form of damage to the steel-frame fielder's masks was plastic (permanent) deformation of the frame near the location of impact. The polycarbonate frames experienced substantial elastic deformation during impact, but no permanent damage was observed during any tests. The ABS mask also experienced deformation during impact, and significant structural damage was noted in four out of eight test conditions. The catcher's mask was not damaged during any test.

Table 8: Summary of observed damage and facial contact for each mask and test condition.

	Nose		Forehead		Temple		Chin	
	L	H	L	H	L	H	L	H
Catcher's Mask	-	-	-	-	-	-	-	-
Mizuno	-	D (3/3)	D (1/3)	D (3/3)	-	D (2/3)	-	-
Rip-It	D (2/3)	D (3/3)	-	D (1/3)	D (1/3)	D (3/3)	-	D (1/3)
Schutt	D (2/3)	D (3/3)	-	D (1/3)	-	-	-	-
Bangerz	D + FC	D + FC	-	D	-	-	-	D
Gameface	-	LC (3/3)	-	-	-	-	-	-
Rawlings	-	LC (3/3)	-	-	-	-	-	-

(-) = No observed damage, L = Low-speed, H = High-speed, D = Damage, FC = Facial contact, LC = Limited facial contact, (N/3) = Number of instances where damage was observed for N = 3 test trials.

All plastic-frame fielder's masks allowed facial contact at the nose region during high-speed tests (Figure 13). The frame of the ABS mask failed under impact to the nose at both low- and high-speed conditions, allowing forceful contact of the ball/mask structure to the face (observed using high-speed video). During high-speed nose impacts, the polycarbonate masks elastically deformed to a sufficient degree that brief facial contact was observed.

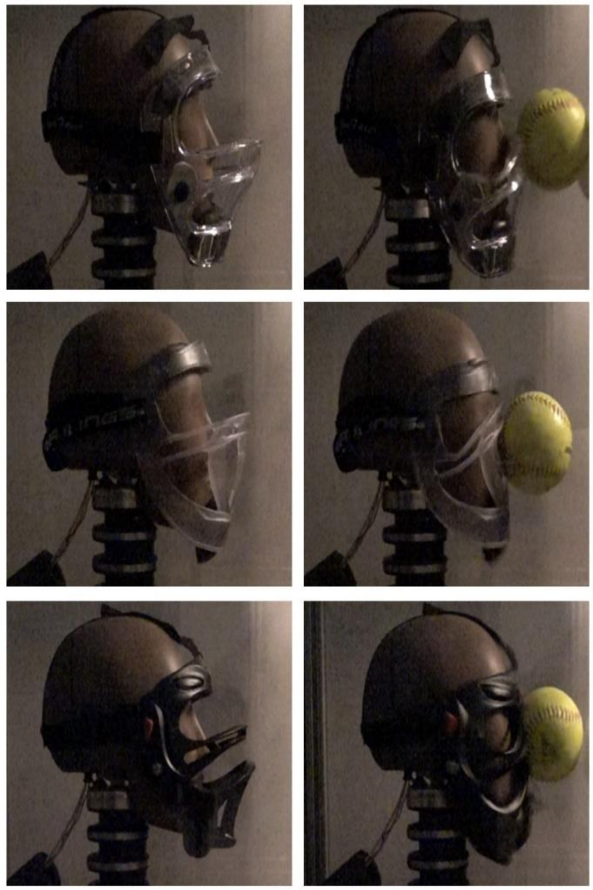


Figure 13: Pre-impact (left) and maximum deformation (right) of the plastic fielder's masks during high-speed nose impacts. Polycarbonate designs are shown in the top and middle rows, and ABS is shown in the bottom row.

3.4 DISCUSSION

The objective of this study was to evaluate the protective capacity of softball fielder's masks, with a focus on their ability to attenuate head accelerations and the concomitant risk of concussion. The masks were compared against a baseline unprotected condition, and a "gold standard" represented by a hockey-style catcher's mask. As seen in the results, the fielder's masks reduced peak accelerations in almost all cases. However, peak angular accelerations for some

fielder's masks displayed no significant reductions under certain impact conditions with respect to unprotected, raising concern about their protective ability in such instances.

We also assessed the masks' ability to prevent facial contact, as this is one of the primary functions of protective defensive fielder's headgear. One source for criteria in assessing facial contact is provided by NOCSAE, wherein a mask fails if any contact is observed with the ocular area of the headform or if the ball/mask structure makes contact with the brow or maxilla/dental regions [8, 51]. We observed that the masks prevented such contact in almost all cases, with notable exceptions for the following: facial contact of the ball and mask structure at the nose region with the ABS fielder's mask, as well as limited contact of the mask structure to the nose region with the polycarbonate fielder's masks. While these plastic-frame fielder's masks are often marketed for their light-weight construction, a stiffer or more substantial frame would be beneficial in reducing the elastic deformation that enabled facial contact to occur.

Across impact locations and speeds, some fielder's masks were observed to perform appreciably different than other designs. Since impact speeds and locations were controlled, these variations are likely attributable to differences in design features of the masks. For example, we observed that most fielder's masks provided little attenuation of peak angular accelerations for high-speed impacts to the forehead. At the same time, most masks tended to consist of similar construction at the forehead – a relatively thin strip of removable foam that wrapped across the interior of the forehead framing. A notable exception was the Schutt, which performed exceptionally well for this impact condition. The Schutt differed from other designs by using a two-part foam pad (the region contacting the wearer being relatively compliant, and the region contacting the mask being relatively stiff), as well as a smooth cloth covering the foam. Qualitatively, the smooth cloth seemed to produce a low-friction interface between the mask and

the head, which could potentially prevent rotational forces in the mask from being transferred to the head. This phenomenon has been observed elsewhere in literature where smooth, low-friction interfaces have been created at the headgear/head interface [65, 66].

While these results provide important information regarding the relative performance of fielder's masks, insight regarding their relative concussive risk can be garnered if the data is analyzed within the context of injury risk curves [28, 67, 68, 69]. As an example, an angular-acceleration-based injury risk curve developed by Pellman et al. [69] was selected (primarily since that curve's underlying data was developed from lab-reconstructed impacts using a 50th percentile Hybrid III, similar to our methodology). Using this curve, the fielder's masks, on average, displayed concussive risks of 5-10% and 3-6% for high- and low-speed nose impacts, respectively, and 7-31% and 3-9% for high- and low-speed forehead impacts, respectively. Higher concussive risks were observed for impacts at the temple and chin regions across all fielder's masks, with the highest reaching up to 67% at the chin region (Figure 14). As a final caveat, we acknowledge that our data could be analyzed using any of the above-referenced risk curves, and that the various available risk curves may provide vastly different estimates of injury probability. However, in that all such risk curves are monotonic in nature, relative rankings would not change by using another curve, only the absolute values of risk.

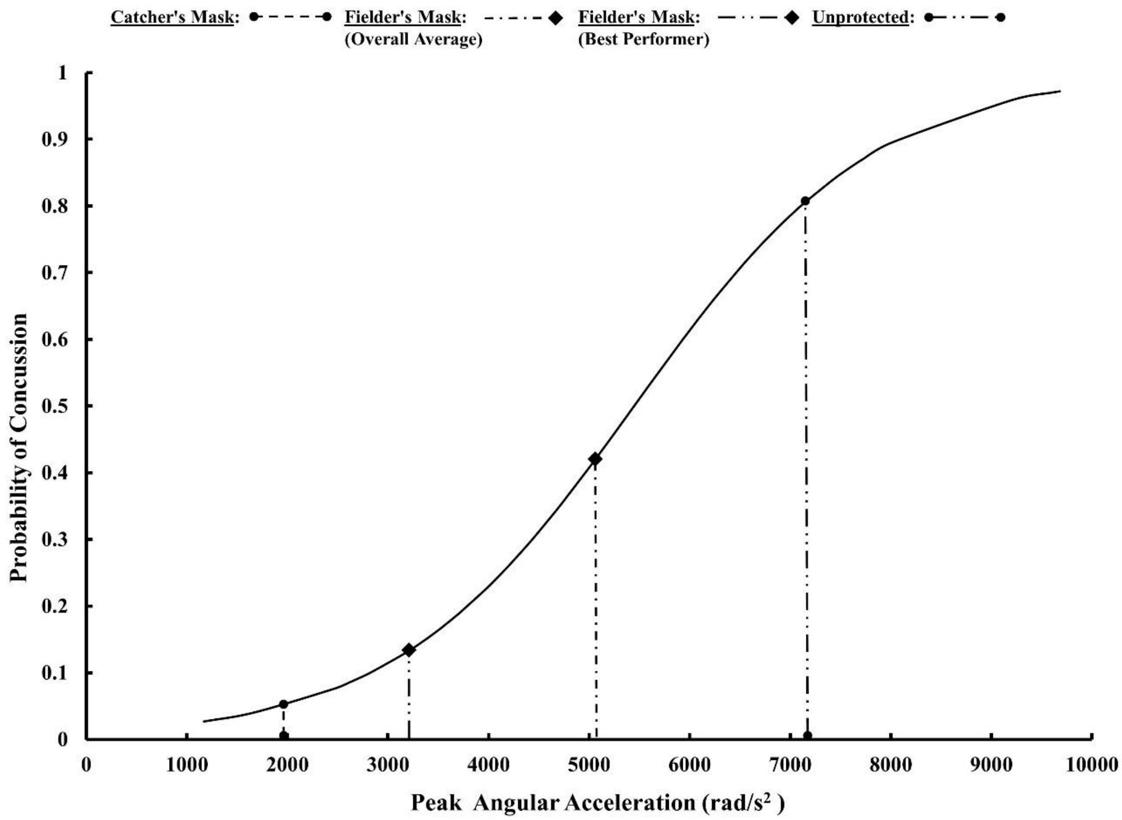


Figure 14: Injury risk curve presenting data for the average values of peak angular acceleration from high-speed chin impacts.

Although our study may provide some limited information regarding the relationship between mask design features and impact attenuation, it was not explicitly designed for this purpose. For example, we observed a wide range of peak angular acceleration values across fielder's masks for high-speed chin impacts. However, there are many factors (i.e. foam padding, frame geometry, mask-head coupling, etc.) involved in the design of these fielder's masks, and parametric evaluations of mask designs would be required to truly detect such effects. Other limitations in our study result from the lack of true bio-fidelity of the Hybrid III, including the lack of an articulated jaw for chin impacts [10] and the difference in material properties of the Hybrid III compared to the actual human head [32]. We also note that anthropomorphic testing devices

(ATDs) exist in various sizes (a 50th percentile male ATD that represents the median of the population, and 5th percentile females and 95th percentile males that represent extremes in the population), and only the 50th percentile male was used in this study. We assert that the 50th percentile male headform serves as the best available option for several reasons. First, the average age of injured softball players is 21.5 years (mature adults) with an approximate 72/28-percent female/male split. Second, anthropometric data indicates there is a relatively small difference between the head sizes of median males and median females (as compared to the variations that occur across the extremes within gender [70]). Third, given the small size of the 5th percentile headform and the large size of the 95th percentile, these ATDs would not even be compatible with all protective headgear evaluated in this study. Finally, utilization of the Hybrid III 50th percentile male headform allows for our results to be compared with most other published sports-impact studies that also use the same headform [9, 10, 11, 60, 71, 72, 73].

CHAPTER 4 - CONCLUDING REMARKS

4.1 CONCLUSIONS

The epidemiological work presented here highlights that numerous head and facial injuries occur from the sport of softball, even though protective headgear exists for all player positions. Most patients were females in their teens and twenties, and the most common injury diagnoses included closed-head injuries, contusions, lacerations, fractures and concussions. Very few players were admitted to hospitals or held for observation. The overwhelming majority of injuries were caused by players being struck by softballs, particularly those in defensive fielder positions, and therefore player safety efforts should be focused in that specific area of play. Overall, it presents novel data related to head/facial injuries sustained in softball and unique information on injury mechanisms, but also highlights the need for more detailed information and research regarding the sport of softball.

The experimental evaluation of fielder's masks revealed that most masks appreciably reduced head accelerations relative to unprotected impacts, but not to the level available with other protective equipment, such as a catcher's mask. Specifically, fielder's masks tended to be more effective at reducing peak linear acceleration than peak angular acceleration. The results suggest that key design features may help certain fielder's masks reduce head accelerations relative to others. Plastic frame fielder's masks were found to be sub-optimal in comparison to steel frame designs based on facial contact and impact attenuation at the forehead, nose, and chin regions. Nonetheless, all masks were preferable to the unprotected condition, thereby providing motivation for defensive players to wear the current available protective headgear whenever possible.

4.2 FUTURE WORK

The findings of this thesis have revealed several avenues for future research. First and foremost, a prime direction is in the investigation of how variations in material properties affect the protective capacity of fielder's masks. Experimentally, factors such as foam properties, low-friction interfaces, and frame geometry were observed to influence acceleration exposures significantly. Finite element modeling would allow for these factors to be parametrically analyzed, and focusing on one element per study could lead towards substantial improvements in mask design or in establishing proper design standards. Efforts could also be focused towards improving plastic frame fielder's masks against facial contact, whilst still maintaining their light-weight feature.

A second avenue of future research is in creating a validated finite-element (FE) model for a fielder's mask. A validated mask model could be applied to a bio-fidelic head FE model (including a brain) to study locations of high stress/strain on brain tissue from softball impacts. Bone or other relevant tissues within the head anatomy could be studied as well. Data from such an application would prove useful in distinguishing relative protective abilities of fielder's masks beyond the standard injury metrics available from ATD testing. In addition to this application, the human-mask models could also be used in supplementing the parametric variation study.

Lastly, a third research avenue is in the development of new injury risk curves based on baseball or softball impact data. The general framework to achieve this is laid out by studies like Pellman et al. [54] and Rowson et al. [55]. Whether accomplished through real-time or laboratory reconstructed methods, these injury risk curves would be novel and provide insight into any differences from a football-based injury risk curve.

4.3 EXPLORATORY WORK

A final avenue of future work involves development of new fielder's masks that improve on deficiencies observed in commercial designs and that capitalize on design principles known to minimize impact severity. Efforts with regards to this research avenue have begun (but not completed) by investigating the benefits of metallization for plastic-type fielder's masks. The process of metallization requires that the base material be struck with a conductive surface, such as titanium or stainless steel, to allow for electroplating of a metallic element, such as nickel. Since the Rawlings was known to allow facial contact, this design was used. However, the existing commercial version of this facemask could not be used since the polycarbonate would not take to the electroplating process. As such, additional masks were made by 3D printing with another type of plastic, Nylon-12, and planned to be struck with a conductive nickel surface and then electroplated with pure nickel (Figure 15).

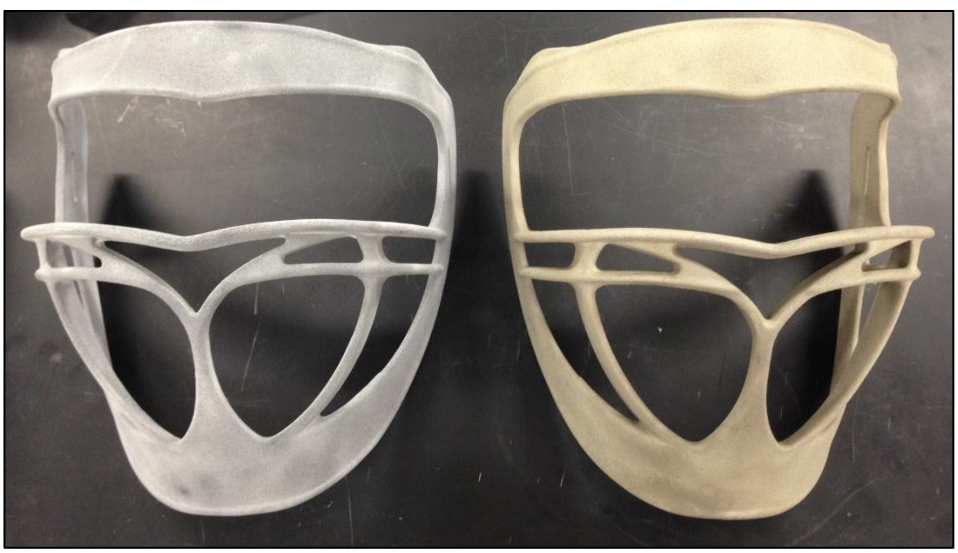


Figure 15: 3D printed, Rawlings fielder's masks without nickel coating (left) and with nickel coating (right).

To conserve time and resources on the electroplating process, efforts were directed towards creating a validated finite-element model for nickel-plated, Nylon-12. This FE model would allow for the prediction of mask deformation as a function of nickel thickness (Figure 16).

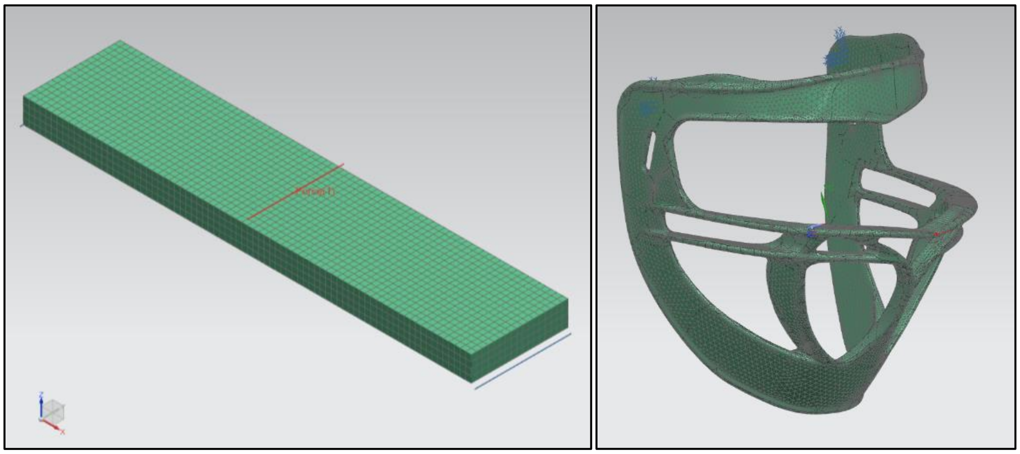


Figure 16: FEM of bar sample (left) and Rawlings fielder's mask (right) with sample test conditions applied.

Thus far, three-point bending tests have been performed for small Nylon-12 bar samples with varying coating thicknesses (0 µm, 50 µm, 100 µm, and 200 µm) (Figure 17).

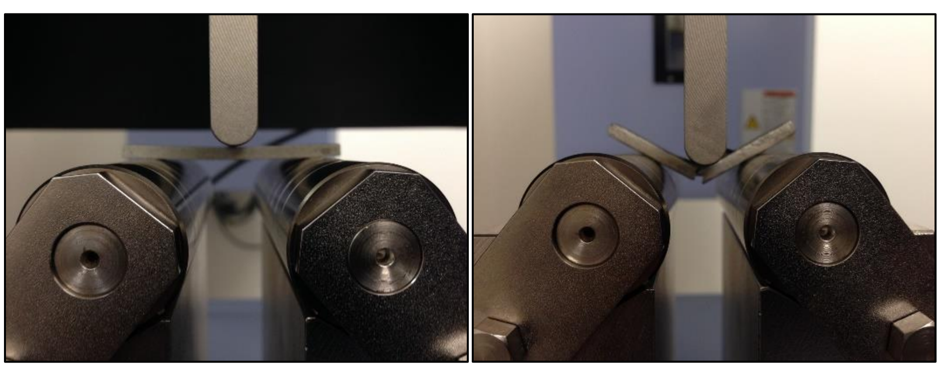


Figure 17: Three-point bend test - nickel-coated bar sample before loading (left) and after coating failure (right).

Results showed an increasing elastic modulus from the 0 µm to the 200 µm samples. However, the data also indicate that the interface bonding between the Nylon-12 and nickel failed at relatively low loading (all tested samples had similar yield points), indicating a flaw with the conductive surface process (Table 9).

Table 9: Results from three-point bend tests.

	0 µm	**50 µm**	**100 µm**	**200 µm**
Average Yield Force (N)	37.68	41.98	40.76	42.50
Average Yield Stress (MPa)	20.85	19.32	17.60	16.25
Average Yield Strain (%)	1.4	0.5	0.2	0.2
Average Elastic Modulus (MPa)	1516	4248	7457	9677

As such, more effective and stronger conductive surface coatings are being pursued, and more three-point bend tests will be conducted with the new samples. This new data will be used to calibrate and validate the FE model, and an optimized metal-coating thickness will be determined such that the mask will prevent facial contact but have a minimum weight. Experimental testing will be used to verify facial contact prevention and provide new acceleration data regarding metallized, plastic frame fielder's masks.

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VITA

John Scott Strickland graduated from the University of North Florida with a Bachelor's of Science in Mechanical Engineering (BSME) and University Honors in the spring of 2017. He decided to continue his education in pursuit of a Master's of Science in Mechanical Engineering (MSME) and was graciously given the opportunity to do so under a Graduate Assistantship with Dr. Bevill. Throughout his two years in the graduate program, he gained valuable skills and knowledge on navigating research studies and engineering projects in a proper and efficient manner. During this time as well, he and his fellow co-author(s) submitted several articles and conference abstracts. He has been proudly accepted and published for articles in the Journal of Applied Biomechanics (JAB) and the Orthopaedic Journal of Sports Medicine (OJSM), and for conference abstracts at the 2018 Conference of Florida Graduate Schools (CFGS) and the 2018 American Society of Mechanical Engineers – International Mechanical Engineering Congress and Exposition (ASME – IMECE).

Publications:

Strickland JS, Bevill GR. Experimental evaluation of softball protective headgear for defensive play. *J App. Biomech.* 2019.

Strickland JS, Crandall M, Bevill GR. A retrospective analysis of softball-related head and facial injuries treated in US emergency departments, 2013-2017. *Orthop J Sports Med.* 2019.

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