EVALUATING THE HEAD INJURY RISK ASSOCIATED WITH BASEBALL AND SOFTBALL

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ABSTRACT (ACADEMIC)

More than 19 million children participate in youth baseball and softball annually. Although baseball and softball are not commonly depicted as contact sports in the, according to the U.S. CPSC baseball and softball were responsible for 11.6% of all head injuries treated in emergency rooms in 2009; third most behind only cycling and football. Ball impact has been identified as the leading cause of injury in baseball and softball, with the most frequent injury resulting from a ball impacting the head. Reduced injury factor balls, infield softball masks, batter’s helmets, and catcher’s masks have all been integrated into baseball and softball as a means for preventing serious head injury from ball impact.

The research in this thesis had four objectives: to compare the responses of the Hybrid III and NOCSAE headforms during high velocity projectile impacts, to compare head injury risk across a range of baseball stiffness designed for different age groups, to evaluate the effectiveness of infielder softball masks’ ability to attenuate facial fracture risk, and to describe a novel methodology to evaluate the performance of batter’s helmets and catcher’s masks. Results of these research objectives determined the most suitable ATD headform to evaluate head injury risk for high velocity projectile impacts, provided a framework for determining the optimal age-specific ball stiffness and optimal infield mask design, and disseminated STAR ratings for batter’s helmets and catcher’s masks to the public. The research presented in this thesis can be used to further improve safety in baseball and softball.
EVALUATING THE HEAD INJURY RISK ASSOCIATED WITH BASEBALL AND SOFTBALL

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ABSTRACT (GENERAL AUDIENCE)

Baseball and softball are two commonly played sports, however, they combine to yield some of the highest head injury rates among sports. Safety measures like protective headgear and softer balls have been implemented into the games, but there is currently no metric for comparing different models and brands on their effectiveness at reducing head injury. The research in this thesis provides an evaluation system that compares the effectiveness of protective headgear between different models and brands and their ability to reduce head injury. This research is presented to the public as a purchasing tool and can be used to further improve the safety in baseball and softball.
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CHAPTER 1 – INTRODUCTION: HEAD INJURY RISK ASSOCIATED WITH BASEBALL AND SOFTBALL

OPENING REMARKS

It is estimated that there could be up to 3.8 million sports related traumatic brain injuries (TBIs) that occur each year. More than 19 million children participate in youth baseball and softball annually. Although baseball and softball are not commonly depicted as contact sports, according to the U.S. Consumer Product Safety Commission (CPSC) baseball and softball were responsible for 11.6% of all head injuries treated in emergency rooms in 2009; third most behind only cycling and football.1,7 Ball impact has been identified as the leading cause of injury in baseball and softball, with the most frequent injury resulting from a ball to the head.4 Ball to head impacts cause 25% of the annual deaths associated with youth baseball and softball, which have the highest fatality rate of all sports between the ages of 5-14.6

Scarcely known, softball has a greater overall injury rate than baseball because the injury rate while fielding batted balls is higher.3 This is likely because softball fields are smaller than baseball fields. The reduction in field size decreases the amount of time fielders have to react to a batted ball and increases their chances of sustaining an injury due to ball impact. One of the most common injuries seen from ball impact in softball is facial fracture.

RESEARCH OBJECTIVES

The U.S. CPSC reported that 36% of all baseball and softball injuries could be prevented, or reduced in severity with the use of safety equipment.2 As a means for preventing serious head injury from ball impact, reduced injury factor balls, infield softball masks, batter’s helmets, and catcher’s masks have all been integrated into baseball and softball. Safety equipment for baseball
and softball is evaluated using anthropomorphic test device (ATD) headforms. ATD headforms
determine the forces and accelerations experienced by a player during a loading event and can be
correlated to injury risk. The research in this thesis had 4 research objectives: to compare the
responses of the Hybrid III and NOCSAE headforms during high velocity projectile impacts, to
compare head injury risk across a range of baseball stiffness designed for different age groups, to
evaluate the effectiveness of infielder masks’ ability to attenuate facial fracture risk, and to
describe the methodology of a new evaluation system for batter’s helmets and catcher’s masks.
Accomplishing these objectives will help improve the safety of baseball and softball by
suggesting the most suitable ATD headform for evaluating protective headgear and by providing
a framework for optimizing age-specific ball stiffness and infield mask design. In addition, age-
specific STAR ratings for batter’s helmets and catcher’s masks will be disseminated to the public
as a purchasing tool, which will provoke manufacturers to design batter’s helmets and catcher’s
masks that reduce concussion risk.

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CHAPTER 2 - COMPARISON BETWEEN THE HYBRID III AND NOCSAE HEADFORMS DURING HIGH VELOCITY PROJECTILE IMPACTS

ABSTRACT

Surrogate headforms are often used to evaluate head injury risk. Two of the most common ATD headforms are the Hybrid III and the NOCSAE. The Hybrid III and NOCSE headforms have been tested and compared to prove that they are capable of recording head accelerations for impact durations ranging from 10 to 15 ms. However, the two headforms have yet to be compared for high velocity projectile impacts. To simulate high velocity projectile impacts, a customized pitching machine was used to propel baseballs and softballs at 15, 25, and 35 m/s into the center of the headforms’ foreheads. Linear acceleration and angular velocity were collected. Analysis of these data showed that during high velocity projectile impacts the NOCSAE headform displayed an oscillatory response for linear accelerations. These oscillations caused an increase in peak linear resultant acceleration of 51%, 55%, and 62% for baseball impacts and 27%, 49%, and 45% for softball impacts at 15, 25, and 35 m/s respectively, compared to the Hybrid III. These differences corresponded to a 84%, 91%, and 93% increase in SI values for baseball impacts and a 56%, 86%, and 77% increase in SI values for softball impacts at 15, 25, and 35 m/s respectively, compared to the Hybrid III. Using an FFT, high frequency peaks were seen ranging from approximately 500 to 1500 Hz and 1500 to 2300 Hz for 15 m/s impacts; approximately 500 to 1500 Hz, 1500 to 2500 Hz, and 2500 to 3200 Hz for 25 m/s impacts; and approximately 750 to 1500 Hz and 1500 to 3000 Hz for 30 m/s impacts. Filtering was not able to correct the high frequency content. Neither ARS signals presented high frequency velocities when analyzed. This work suggests that a Hybrid III headform should be used when examining the biomechanical response during high velocity projectile impacts.
Keywords: head impacts, baseball, softball, linear, rotational, acceleration, frequency, biomechanics

INTRODUCTION

Anthropomorphic test devices (ATD) are humanoid dummies used to determine the forces and accelerations experienced by the body during a traumatic event. Injury predictors are acquired using load cells, potentiometers, angular rate sensors (ARS), and linear accelerometers. ATD headforms are extensively used to study different head impact scenarios. Two commonly used ATD headforms are the Hybrid III and the National Operating Committee on Standards for Athletic Equipment (NOCSAE).² ³ ⁵

Although both headforms have been tested to ensure that their biomechanical response is biofidelic, they do present structural differences.³ The Hybrid III headform was designed for automotive testing, but has been used for helmet testing. It was modeled after an average American man, constructed using an aluminum shell with a hollow brain cavity for instrumentation, and covered with a vinyl skin.⁵ The NOCSAE headform was developed to evaluate protective athletic headgear and modeled after the shape and size of an average American football player.³ The NOCSAE headform consists of a nylon shell, filled with glycerin, and covered with urethane.⁵ A shaft insert from the bottom of the headform is used for instrumentation. The NOCSAE headform is usually preferred for evaluating athletic headgear because it mimics the contours of the human head, providing a more realistic fit.²

Both headforms have been used to study sports related head impacts, which cause an estimated 3.8 million traumatic brain injuries annually.⁶ Instrumented athletes, volunteer experiments, and cadaver tests have helped increase our knowledge of head injury and led to the
development of head injury risk curves. These curves have been used to evaluate the effectiveness of helmets in the laboratory.

A headform comparison between the NOCSAE and Hybrid III for impact durations commonly seen in helmeted sports (10-15 ms) found miniscule differences between the two, validating the use of either headform for helmet testing. However, the two headforms have yet to be compared for high velocity projective impacts (1-5 ms), which are commonplace in sports like baseball and softball. Previous studies have used the Hybrid III headform to analyze head impacts in baseball and softball, even though the NOCSAE headform is used to certify protective head equipment worn in these sports. Preliminary laboratory testing determining the difference in biomechanical response between the two headforms during baseball and softball impacts showed unexpected oscillations in the NOCSAE’s acceleration pulses. These results provoked a further evaluation of the headforms to determine if the NOCSAE headform was capable of capturing high velocity projectile impacts. The objective of this study was to compare the respective responses of the Hybrid III and NOCSAE headforms during high velocity projectile impacts.

METHODS

To simulate projectile head impacts in sports, baseballs and softballs were projected using a combination pitching machine. The dual wheeled, electric motor-driven machine (Jugs Sports Combination Pitching Machine Model SR3616-681-7, Tualatin, OR) was customized and anchored to the floor to reduce unwanted vibration. Each wheel had an independent speed dial with digits ranging from 0 to 100 that could be set to obtain desired speeds. The wheels were pressurized to 17 psi and their speed dials maintained at least a 35 digit offset from one another to prevent the ball from knuckling, as specified in the manual. In order to minimize ball loading
differences, custom ball holders were constructed to load the balls with the same orientation into the pitching machine. The ball velocity was calculated over the final 10.16 cm of the baseball mount using a dual laser velocity gate sensor (Velocity Timer Model 1204, KME Company, Troy, MI) and was repeatable within ± 3% of the desired velocity.\textsuperscript{1} The customized pitching machine possessed an accuracy within a 0.635 cm radius circle, which was verified from a previous study using the same machine.\textsuperscript{1} The baseballs used for testing were manufactured by Rawlings (model ROMLB) with a COR of 0.55. The softballs used for testing were also manufactured by Rawlings (model C12RYLAH) and had a COR of 0.47.

Linear accelerations and angular velocities of the head were evaluated using the response of a two commonly used surrogate headforms. A 50\textsuperscript{th} percentile male Hybrid III headform and a medium NOCSAE headform were attached to a 50\textsuperscript{th} percentile Hybrid III neck and mounted to a 16 kg sliding table that mimicked the inertial properties of the upper torso. Note that the NOCSAE headform was modified to allow attachment of a 50\textsuperscript{th} percentile Hybrid III neck.\textsuperscript{3} To collect linear acceleration and rotational velocity data, the headforms were instrumented with three linear accelerometers and an ARS at the center of gravity of the headform.\textsuperscript{3,12} Data acquisition was conducted using a TDAS Slice Pro (DTS, Seal Beach, CA) system with a sampling rate of 20 kHz. The experimental setup is depicted in Figure 2.1.
Figure 2.1. Anchored customized pitching machine that propels projectiles into either a 50th percentile male Hybrid III headform, or a medium NOCSAE headform. The headforms are connected to 50th percentile Hybrid III neck and mounted to a 16-kg sliding table that replicates the upper torso.

The headforms were positioned vertically, 35.56 cm away from the launcher and impacted at the center of the forehead, 8.0 cm above the tip of the nose (Figure 2.12.2). The center of the forehead was chosen because it was the flattest, most repeatable location to evaluate head injury predictors. The balls were projected at 15, 25, and 35 m/s. Three trials were conducted at each velocity and the average linear accelerations and rotational velocities were taken from each headform.

Figure 2.2. Baseballs and softballs impacted the Hybrid III and NOCSAE headforms in the center of the forehead. The center of the forehead was 8.0 cm above the tip of the nose. The forehead was chosen because it was the flattest and most reliable location on the headform to evaluate head injury predictors.

Data collected were processed according to SAE J211. The linear accelerometers at the center of gravity of both headforms were filtered at CFC 1000. The ARS signals were filtered at
CFC 180 for the Hybrid III and CFC 155 for the NOCSAE.³ A custom-written MATLAB (Mathworks, Natick, MA) code was used to calculate the desired metrics, peak resultant linear acceleration, peak resultant angular velocity, and Severity Index (SI). A fast Fourier transform (FFT) was performed on the linear resultant accelerations and the angular velocities to detect high frequency peaks of linear resultant acceleration magnitude and angular velocity magnitude respectively in the headforms.

RESULTS

The acceleration pulses seen in baseball and softball impacts generally settled by 6 ms and had a majority of the signal in the first 2 ms. Analyzing the unfiltered linear resultant accelerations and unfiltered linear x accelerations as a function of time for baseball and softball impacts at 15, 25, and 35 m/s revealed an oscillatory response in the NOCSAE headform (Figures 2.3, 2.5, and 2.7). Oscillations in the NOCSAE headform’s linear acceleration pulses resulted in an increase of peak linear resultant acceleration of 51%, 55%, and 62% for baseball impacts at 15, 25, and 35 m/s respectively, compared to the Hybrid III headform. For softball impacts, the NOCSAE headform displayed a 27%, 49%, and 45% increase in peak linear resultant acceleration at 15, 25, and 35 m/s respectively, compared to the Hybrid III. These differences corresponded to a 84%, 91%, and 93% increase in SI values for baseball impacts and a 56%, 86%, and 77% increase in SI values for softball impacts at 15, 25, and 35 m/s respectively, compared to the Hybrid III. Only the linear x accelerations were analyzed because the impact occurred along the x axis, contributing to a majority of the linear resultant. The filtered linear accelerations displayed the same oscillatory responses as the unfiltered data, but were slightly lower magnitudes. For baseball impacts, filtering linear resultant accelerations yielded a reduction in peak linear acceleration of only 4% for a 35 m/s impact in the Hybrid III
headform, but reduced the peak linear acceleration magnitudes of the NOCSAE headform 10%, 18%, and 23% for 15, 25, and 35 m/s impacts respectively. This corresponded to a reduction in SI values of 0%, 6%, and 10% for the Hybrid III and 17%, 20%, and 31% for the NOCSAE at 15, 25, and 35 m/s respectively. Filtering softball impacts reduced the peak linear resultant acceleration magnitudes 2% at 25 and 35 m/s for the Hybrid III headform. For the NOCSAE headform, peak linear acceleration magnitudes were reduced 9%, 11%, and 13% for filtered softball impacts at 15, 25, and 35 m/s respectively. Corresponding SI reduction for both headforms were 0%, 3%, and 4% for the Hybrid III and 13%, 19%, and 24% for the NOCSAE at softball impacts of 15, 25, and 35 m/s respectively. The filtered and unfiltered peak linear resultant acceleration values and corresponding SI values for baseball and softball impacts are presented in Tables 2.1 and 2.2.

An FFT was conducted on the linear resultant acceleration traces to provide a comparison between headforms in the frequency domain (Figures 2.4, 2.6, and 2.8). The majority of the linear resultant magnitude occurred at approximately 0 Hz. For the Hybrid III, the linear resultant magnitude exponentially decayed from approximately 2500 g to 0 g over a span of 1500 to 2000 Hz. The linear resultant magnitude of the NOCSAE headform decayed similarly, but displayed peaks at higher frequencies. For ball impacts at 15 m/s the NOCSAE headform illustrated two distinct peaks ranging from approximately 500 to 1500 Hz and from approximately 1500 to 2300 Hz. The NOCSAE headform displayed three high frequency peaks during the 25 m/s impact, ranging from approximately 500 Hz to 1500 Hz, 1500 to 2500 Hz, and from 2500 to 3200 Hz. At 35 m/s the NOCSAE headform presented two peaks, ranging from approximately 750 to 1500 and from 1500 to 3000 Hz. The high frequency peaks seen in the NOCSAE headform for this range of velocities correlates to high frequency accelerations in the time domain. High frequency
accelerations influence the biomechanical response, producing unrealistic head acceleration and SI values. Higher frequency peaks were not seen in the ARS signals of the NOCSAE and Hybrid III headforms. Like the peak linear accelerations, filtering had little effect on the frequency responses and ARS signals for baseball and softball impacts at each velocity.

Figure 2.3. Displays the unfiltered linear resultant and linear x acceleration traces for baseball and softball impacts at 15 m/s. The oscillatory response of the linear accelerations for the NOCSAE headform resulted in peak linear acceleration values approximately double the Hybrid III’s.
Figure 2.4. Transforming the unfiltered linear resultant accelerations at 15 m/s from the time domain to the frequency domain using an FFT showed two high frequency peaks ranging from approximately 500 to 1500 Hz and from approximately 1500 to 2300 Hz.
Figure 2.5. Illustrates the unfiltered linear resultant and linear x acceleration traces for baseball and softball impacts at 25 m/s. The oscillatory response of the linear accelerations for the NOCSAE headform resulted in peak linear acceleration values approximately double the Hybrid III’s.
Figure 2.6. Transforming the unfiltered linear resultant accelerations at 25 m/s from the time domain to the frequency domain using an FFT displayed three high frequency peaks ranging from approximately 500 to 1500 Hz, 1500 to 2500 Hz, and 2500 to 3200 Hz.
Figure 2.7. Depicts the unfiltered linear resultant and linear x acceleration traces for baseball and softball impacts at 35 m/s. The oscillatory response of the linear accelerations for the NOCSAE headform resulted in peak linear acceleration values approximately double the Hybrid III’s.
Figure 2.8. Transforming the unfiltered linear resultant accelerations at 35 m/s from the time domain to the frequency domain using an FFT presented two high frequency peaks ranging from approximately 750 to 1500 Hz and from approximately 1500 to 3000 Hz.

Table 2.1. Filtered and unfiltered peak linear resultant accelerations and SI values with standard deviations for the Hybrid III and NOCSAE headforms during baseball impacts at 15, 25, and 35 m/s. The differences in peak linear resultant acceleration and SI increased between the headforms as the velocity increased.

<table>
<thead>
<tr>
<th>Velocity (m/s)</th>
<th>Headform</th>
<th>Unfiltered Peak Linear Resultant (g)</th>
<th>SI</th>
<th>Filtered Peak Linear Resultant (g)</th>
<th>SI</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>SI</td>
<td></td>
<td>SI</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>Hybrid III</td>
<td>91 ± 6</td>
<td>38 ± 5</td>
<td>91 ± 5</td>
<td>38 ± 5</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>186 ± 10</td>
<td>232 ± 24</td>
<td>167 ± 12</td>
<td>193 ± 16</td>
</tr>
<tr>
<td>25</td>
<td>Hybrid III</td>
<td>187 ± 6</td>
<td>188 ± 14</td>
<td>186 ± 6</td>
<td>177 ± 11</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>416 ± 26</td>
<td>2098 ± 83</td>
<td>340 ± 20</td>
<td>1675 ± 65</td>
</tr>
<tr>
<td>35</td>
<td>Hybrid III</td>
<td>326 ± 5</td>
<td>620 ± 12</td>
<td>313 ± 3</td>
<td>553 ± 13</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>851 ± 125</td>
<td>8409 ± 2156</td>
<td>656 ± 97</td>
<td>5831 ± 1456</td>
</tr>
</tbody>
</table>
Table 2.2. Filtered and unfiltered peak linear resultant accelerations and SI values with standard deviations for the Hybrid III and NOCSAE headforms during softball impacts at 15, 25, and 35 m/s. The differences in peak linear resultant acceleration and SI increased between the headforms as the velocity increased.

<table>
<thead>
<tr>
<th>Velocity (m/s)</th>
<th>Headform</th>
<th>Unfiltered</th>
<th>Filtered</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Peak Linear</td>
<td>SI</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Resultant (g)</td>
<td></td>
</tr>
<tr>
<td>15</td>
<td>Hybrid III</td>
<td>118 ± 3</td>
<td>79 ± 3</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>162 ± 23</td>
<td>178 ± 56</td>
</tr>
<tr>
<td>25</td>
<td>Hybrid III</td>
<td>200 ± 9</td>
<td>247 ± 21</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>390 ± 43</td>
<td>1727 ± 509</td>
</tr>
<tr>
<td>35</td>
<td>Hybrid III</td>
<td>302 ± 16</td>
<td>675 ± 56</td>
</tr>
<tr>
<td></td>
<td>NOCSAE</td>
<td>550 ± 30</td>
<td>2970 ± 632</td>
</tr>
</tbody>
</table>

DISCUSSION

Comparison of the Hybrid III and NOCSAE headforms during high velocity projectile impacts showed disagreement in linear accelerations and SI values. The NOCSAE headform presented high frequency accelerations that produced unrealistic peak linear acceleration values. The differences in peak linear resultant accelerations between the Hybrid III and NOCSAE headforms increased with velocity (Figure 2.9). The differences in SI values between the two headforms shows an inconsistency in determining head injury metrics. For example, the maximum SI value the Hybrid III recorded during baseball impacts was 620 at 35 m/s. Contrastingly, the NOCSAE headform generated a maximum SI value of 8,409 during the same impact. Filtering was not able to correct for the high frequency accelerations, which suggests that the structural differences between the headforms is the cause for disparity.

The ARS signals were also compared and analyzed, but presented no signs of high frequency velocities influencing the responses. This analysis suggests that the NOCSAE
headform should not be used to evaluate athletic equipment that experiences high velocity projectile impacts with impact durations ranging from 1-5 ms.

Figure 2.9. Illustrates the differences in unfiltered linear resultant accelerations as a function of velocity for the Hybrid III and NOCSAE headforms (left: baseball, right: softball). As the velocity increases, the differences in peak linear resultant accelerations between the headforms increases. On average the NOCSAE headform produced at least double the linear resultant acceleration than the Hybrid III.

While this study was able to evaluate the ability of the Hybrid III headform and the NOCSAE headform to capture a high velocity projectile impact, there were a few limitations. The test setup was not the same as the NOCSAE standard used in baseball and softball testing. NOCSAE uses a headform rigidly attached to a slider table that is instrumented with a tri-axial accelerometer, instead of three linear accelerometers. It is unclear if similar high frequency content is observed in that test setup. In addition, the instrumentation in the NOCSAE headform used was in a slightly different location because the headform had to be modified to attach a 50th percentile Hybrid III neck, which could be a cause for the resonance seen.
CONCLUSIONS

The use of ATD headforms to collect head injury data allows injury risks to be calculated for traumatic events. Although the NOCSAE headform is typically used to evaluate protective head equipment in athletics, the Hybrid III headform better measured the response of high velocity projectile impacts that are commonly seen in baseball and softball. During high velocity projectile impacts, the NOCSAE headform displayed an oscillatory response for linear accelerations in the time domain, which translated to high frequency peaks in the frequency domain. High frequency peaks indicated a contribution of high frequency accelerations in the acceleration pulses. High frequency accelerations create unrealistic acceleration pulses that could lead to an inaccurate representation of what baseball and softball players would experience during high velocity projectile impacts. It is suggested that head protection designed for baseball and softball should be tested with a Hybrid III headform over a modified NOCSAE headform to accurately capture the loading event.

REFERENCES

5 Kendall, M., E. S. Walsh and T. B. Hoshizaki. Comparison between hybrid iii and hodgson–wsu headforms by linear and angular dynamic impact response. Proceedings of the Institution of
12 NOCSAE. Standard projectile impact test method and equipment used in evaluating the performance characteristics of protective headgear, faceguards or projectiles. National Operating Committee on Standards for Athletic Equipment: NOCSAE DOC (ND) 021-98m09, 2009.
CHAPTER 3 - HEAD INJURY RISK ASSOCIATED WITH BASEBALL STIFFNESS AS A FUNCTION OF PLAYER AGE

ABSTRACT

The majority of head injuries in baseball are due to ball impact. To reduce injury risk, standard baseball stiffness varies between age groups. The objective of this study was to compare head injury risk across a range of baseball stiffness (RIF1, RIF5, RIF10, Youth, HS/College and Pro) designed for different age groups. To simulate baseball impacts, a customized pitching machine was used to propel baseballs from 15 m/s to 30 m/s in 5 m/s increments. The balls impacted the center of the forehead of a 50th percentile Hybrid III headform. The headform was connected to a Hybrid III neck, mounted on a 16-kg sliding table, positioned vertically and instrumented with a 9 accelerometer array in a 3-2-2-2 configuration. To account for head size differences between ages, acceleration data collected from the Hybrid III were transformed using geometric scaling laws. Skull fracture risk and concussion risk were compared between ball types at each impact velocity. Analysis of these data show that the youth ball, age 13-14, produced the highest skull fracture and concussion risk across the velocity range, however at age matched velocity the professional level (Pro ball) yielded the greatest skull fracture and concussion risk and the safety balls used for 5-8 year olds (RIF 1) yielded the lowest skull fracture and concussion risk. This study provides framework for determining optimal age-specific ball stiffness.

Keywords: head impacts, baseball, head injury risk, linear, rotational, acceleration, biomechanics
INTRODUCTION

It is estimated that in the United States there are more than 19 million children that participate in youth baseball annually. Baseball players between the ages of 5-14 sustain the highest fatality rate of all sports, with approximately one in four annual deaths resulting from an impact from the ball to the head. Ball impact has been identified as the leading cause of injury in baseball, with the most common injury being from the ball to the head. A pitcher throwing a ball toward the head of a batter and striking the unprotected head is one specific scenario that can result in head injury, and is the interest of this study. These impacts can lead to concussion, skull fracture, and in some instances death. Development of reduced injury factor (RIF) balls have provoked rule changes to specify certain ball stiffness to different age groups as a way to mitigate injury. RIF baseballs range from levels 1-10, with 1 being the softest and 10 being the hardest. Previous studies on the effect baseball hardness has on injury risk have shown that a softer ball reduces the potential for head injury. RIF 1, RIF 5, and RIF 10 balls have been specified for age groups 5-8, 7-10, and 9-12 respectively. In addition, there is a youth style ball for ages 13-14, a high school and college style ball, and a professional ball for ages 14 and higher.

A baseball must be certified by the National Operating Committee on Standards for Athletic Equipment (NOCSAE) prior to its use in the field of play. The requirements are as follows: weigh between 5.0 and 5.25 ounces, have a circumference within 9 to 9.25 inches, and a coefficient of restitution (COR) value between 0.45 and 0.55. Depending on the ball compression type (low, medium, and high) the compression deflection value at 0.25 in displacement must not exceed 45 lbs., be within 75-150 lbs., or be within 200-350 lbs. respectively.
Few studies have investigated age specific head injury risk as a function of baseball stiffness. The objective of this study was to compare head injury risk across a range of baseball stiffness designed for different age groups. A better understanding of these risks will help aid in the optimization of age-specific ball stiffness.

METHODS

To simulate a baseball being pitched toward a batter’s head, baseballs were projected by a baseball pitching machine. The pneumatic wheeled, electric motor-driven machine (Jugs Sports Combination Pitching Machine Model SR3616-681-7, Tualatin, OR) was customized and anchored to the floor to reduce unwanted vibration. Each wheel had an independent speed dial with digits ranging from 0 to 100 that could be set to acquire desired speeds. The wheels were pressurized to 17 psi and their speed dials maintained at least a 35 digit offset from one another to prevent the ball from knuckling, as specified in the manual. In order to minimize ball loading differences, custom ball holders were constructed to load the balls with the same orientation into the pitching machine. The ball velocity was calculated over the final 10.16 cm of the baseball mount using a dual laser velocity gate sensor (Velocity Timer Model 1204, KME Company, Troy, MI) and was repeatable within ± 3% of the desired velocity. The customized pitching machine possessed an accuracy within a 0.635 cm radius circle, which was verified from a previous study using the same machine. The baseballs tested were all manufactured by Rawlings with a COR of 0.55. The models used for each age group were ROMLB (Pro) for the professional level, R100-H2 (HS/Col) for the high school and collegiate level, RDYB1 (Youth) for 13 to 14 year olds, RIF10L (RIF 10) for 9-12 year olds, RIF5L (RIF 5) for 7-10 year olds, and Level 1 Training Baseball (RIF 1) for 5-8 year olds and are listed in the order from hardest to softest.
Head injury risk as a function of ball type was evaluated using the response of a surrogate headform. A Hybrid III anthropomorphic test device (ATD) 50th percentile male head and neck were mounted to a 16 kg sliding table that mimicked the inertial properties of the upper torso. To collect linear and rotational head accelerations, the Hybrid III headform was instrumented with a nine accelerometer array (Endevco 7264B-2000, Irvine, CA) in a 3-2-2-2 configurations. Data acquisition was conducted using a TDAS Slice Pro (DTS, Seal Beach, CA) system with a sampling rate of 20 kHz. The experimental setup is depicted in Figure 3.1.

![Figure 3.10. Anchored customized pitching machine that projects baseballs into a 50th percentile male Hybrid III head and neck system that is mounted to a 16-kg sliding table. The baseballs impact the Hybrid III headform in the center of the forehead, 8.0 cm above the tip of the nose.](image)

The headform was positioned vertically, 35.56 cm away from the launcher. Impact location was at the center of the forehead, which was 8.0 cm above the tip of the nose (Figure 2.1). The center of the forehead was chosen because it was the flattest, most repeatable location to evaluate head injury risk by ball type. The balls were tested from the softest ball, RIF 1, to the hardest ball, Pro, over a range of velocities from 15 m/s (33.55 mph) to 30 m/s (67.1 mph) in 5 m/s increments. Each ball was tested three times at each velocity. This range of velocities was chosen because it resembles the real world plate velocities seen by the different age groups. Table 3.1 shows the average plate velocity by age group, or age matched velocity. These
velocities were calculated by taking the average pitch speed by age and reducing it by 9%, which has been found to be the average velocity lost from the pitcher’s mound to the batter’s box.\textsuperscript{1,2}

Table 3.1. The average plate velocity for each age group is the impact velocity players would encounter. The average plate velocity was calculated by subtracting 9% of the average pitch velocity for each age group, accounting for the average velocity lost from the pitcher’s mound to home plate.

<table>
<thead>
<tr>
<th>Age Group</th>
<th>5-8</th>
<th>7-10</th>
<th>9-12</th>
<th>13-14</th>
<th>High School-College</th>
<th>Professional</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ball Type</td>
<td>RIF 1</td>
<td>RIF 5</td>
<td>RIF 10</td>
<td>Youth</td>
<td>HS/Col</td>
<td>Pro</td>
</tr>
<tr>
<td>Average Plate Velocity (m/s)</td>
<td>18.3</td>
<td>18.3</td>
<td>22.4</td>
<td>26.4</td>
<td>32.6</td>
<td>36.2</td>
</tr>
</tbody>
</table>

Data collected were processed according to SAE J211. The acceleration data used to compute linear accelerations were filtered using channel frequency class (CFC) 1000 and the acceleration data used to compute rotational accelerations were filtered at CFC 180.\textsuperscript{9} A custom-written MATLAB (Mathworks, Natick, MA) code was used to calculate the desired metrics, peak resultant linear acceleration and peak resultant angular acceleration. To account for differences in head size between age groups, geometric length scaling using average circumferential head length was performed to best represent the head accelerations experienced by each age group.\textsuperscript{7} Differences in head size can either increase or decrease the head accelerations experienced. Since these data were collected from an ATD representing a head size from the oldest age group (largest head), the accelerations seen by the smaller heads were increased as a result of scaling. Equations 1, 2, and 3 were used to scale the peak resultant linear and rotational accelerations using average circumferential head length of each age group. Here \( \lambda_L \) is the length scaling factor, \( L_z \) is the average head circumference of each age group, \( L_i \) symbolizes the circumferential length of the Hybrid III head, \( \alpha_z \) and \( \alpha_z \) represent the scaled linear and rotational accelerations respectively, and \( \alpha_i \) and \( \alpha_i \) are the measured linear and rotational accelerations of the instrumented Hybrid III head respectively.
\[ \lambda_L = \frac{L_2}{L_i} \]  
\[ a_s = \lambda_L^{-1}a_i \]  
\[ \alpha_s = \lambda_L^{-2}\alpha_i \] (eq. 1, eq. 2, eq. 3)

Table 3.2 displays the scaling factors used. Data were not scaled for the HS/Col ball and Pro ball because the Hybrid III was an accurate representation of their head size. Skull fracture risk was calculated using scaled peak linear acceleration and the methodology of Mertz et al for each ball type as a function of velocity.\(^6\) Concussion risk was calculated with equation 4 by inputting scaled peak linear acceleration, \( a \), and scaled peak rotational acceleration, \( \alpha \), for each ball type as a function of velocity.\(^{12}\) An ANOVA was conducted for the effect ball type had on skull fracture risk, p-value of 0.05, and Tukey’s Honest Significant Difference test post hoc.

\[ CP = \frac{1}{1 + e^{-(-10.2 + 0.0433 \cdot x + 0.000275 \cdot x^2 - 0.00000092 \cdot x^3 + 0.00000002 \cdot x^4)}} \] (eq. 4)

Table 3.2. Average head circumference lengths, in centimeters, for each age group and the calculated length scaling factor, \( \lambda_L \). The length scaling factor was calculated by dividing the average circumferential head length of each age group by the circumferential head length of the instrumented headform, the Hybrid III. The 50\(^{th}\) percentile Hybrid III accurately represented the head circumference of the High School/College and Professional age groups, so the scaling multiplier was one. These scaling factor can then be applied to obtain the estimated accelerations experienced at each age group.

<table>
<thead>
<tr>
<th>Age Group</th>
<th>5-8</th>
<th>7-10</th>
<th>9-12</th>
<th>13-14</th>
<th>50(^{th}) % Hybrid III</th>
</tr>
</thead>
<tbody>
<tr>
<td>Avg Head Circumference (cm)</td>
<td>51.75</td>
<td>52.60</td>
<td>53.35</td>
<td>54.55</td>
<td>57.20</td>
</tr>
<tr>
<td>( \lambda_L )</td>
<td>0.90</td>
<td>0.92</td>
<td>0.93</td>
<td>0.95</td>
<td>1.0</td>
</tr>
</tbody>
</table>

RESULTS

Table 3.3 shows the average scaled and unscaled linear and rotational accelerations for each ball type at the tested velocities. The Youth ball produced the maximum scaled linear peak acceleration for each velocity, 104, 142, 208, and 272 g’s, respectively. The RIF 1 ball yielded
minimum scaled linear peak acceleration values for each velocity, 46, 61, 87, and 121 g’s respectively, however the RIF 5 ball was almost identical to the RIF 1. It can also be seen that the RIF 10 ball produced the maximum scaled rotational accelerations, 3246 and 4506, at 15 and 20 m/s respectively. The youth ball produced the maximum scaled rotational acceleration, 5731, at 25 m/s and 6922, at 30 m/s. The minimum scaled rotational acceleration values at 15 and 25 m/s resulted from the RIF 5 ball, while the RIF 1 ball yielded the minimum at 20 m/s and the HS/Col ball yielded the minimum at 30 m/s.

The skull fracture risk as a function of velocity is presented in Figure 2. From 15 to 20 m/s there was little difference in skull fracture risk between the ball types, however, above 20 m/s the Youth ball resulted in the highest risk of skull fracture. Across all velocities, there was no difference in skull fracture risk between the RIF 1 and RIF 5 balls. Statistical comparison between ball types can be found in Table 3.4. Evaluating concussion risk as a function of velocity by ball type generated two distinct ball groupings (Figure 3.2). The Pro, HS Col, Youth, and RIF 10 balls all were very similar across the velocity range and concussion risk approached 100% at 30 m/s. Conversely, the RIF 5 and RIF 1 ball types were almost identical across the velocity range, but only produced a concussion risk of about 60% for a 30 m/s impact.
Table 3.3. Displays the average scaled and unscaled peak linear accelerations (PLA) with their respective standard deviations (PLA SD) and the average scaled and unscaled peak rotational accelerations (PRA) with their respective standard deviations (PRA SD) for each ball type and velocity. The Youth ball had the highest PLA across all velocities and the RIF 5 and RIF 1 balls had the lowest PLA, with very little differences between them.

<table>
<thead>
<tr>
<th>Ball Type</th>
<th>Velocity (m/s)</th>
<th>Avg PLA (g)</th>
<th>PLA SD</th>
<th>Avg Scaled PLA (g)</th>
<th>Scaled PLA SD</th>
<th>Avg PRA (rad/s²)</th>
<th>PRA SD</th>
<th>Avg Scaled PRA (rad/s²)</th>
<th>Scaled PRA SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pro</td>
<td>15</td>
<td>85</td>
<td>5</td>
<td>85</td>
<td>5</td>
<td>3096</td>
<td>192</td>
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<td>9</td>
<td>6911</td>
<td>168</td>
<td>6911</td>
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<tr>
<td>HS/Col</td>
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<td>7</td>
<td>99</td>
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<td>8</td>
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<td>Youth</td>
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<td>7</td>
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<td>8</td>
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<td>72</td>
<td>6922</td>
<td>80</td>
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<td>81</td>
<td>5</td>
<td>86</td>
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<td>295</td>
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<td>117</td>
<td>11</td>
<td>4203</td>
<td>663</td>
<td>4506</td>
<td>762</td>
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<td>198</td>
<td>5</td>
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<td>38</td>
<td>6855</td>
<td>44</td>
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<tr>
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<td>47</td>
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<td>63</td>
<td>6</td>
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<td>106</td>
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<td></td>
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Figure 3.211. The skull fracture risk of each ball type was calculated using the scaled PLA (left). The youth ball produced the highest risk, 58%, while the RIF 5 and RIF 1 balls resulted in approximately 0% risk across the range of tested velocities. The concussion risk (right) was calculated by inputting scaled accelerations into the bivariate risk function and showed two distinct groupings between balls. In one group, the Pro, HS/Col, Youth, and RIF 10 balls all had very similar risk values across the velocities, maxing out at approximately 100%. In the other group, the RIF 5 and RIF 1 balls were almost identical with a maximum risk value of 58%.
Table 3.4. A connecting letters report showing the differences between ball types at each velocity for skull fracture risk. If the ball types are not connected by the same letter, they are considered to be different from one another, p-value of 0.05.

<table>
<thead>
<tr>
<th>Velocity (m/s)</th>
<th>Ball Type</th>
<th>Mean (%)</th>
<th>Group</th>
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<td>15</td>
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<td>A</td>
</tr>
<tr>
<td></td>
<td>HS/Col</td>
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<td>A</td>
</tr>
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<td></td>
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<td>B</td>
</tr>
<tr>
<td></td>
<td>Pro</td>
<td>0.01</td>
<td>C</td>
</tr>
<tr>
<td></td>
<td>RIF 5</td>
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<td>C</td>
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<tr>
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<td>RIF 1</td>
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<td>C</td>
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<td>B</td>
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<td>Pro</td>
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<td>B</td>
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<td></td>
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<tr>
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<td>A</td>
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**DISCUSSION**

Evaluating the head injury risk by ball type on a bare Hybrid III headform showed that ball type greatly influenced injury risk. It was found that at 15 m/s the Youth ball was different than the RIF 10, Pro, RIF 5, and RIF 1 balls and the HS Col ball was different than the Pro, RIF 5 and RIF 1 balls. For 20 m/s the HS Col ball was the only ball that was different from the rest. The Youth, Pro, and RIF 10 balls were all different from one another and the HS Col, RIF 5, and RIF 1 ball types at 25 m/s. At 30 m/s the Youth ball is different from the HS Col, RIF 10, RIF 5, and RIF 1 balls; the Pro ball is different than RIF 10, RIF 5, and RIF 1 balls; and the HS Col ball was different than the RIF 10, RIF 5, and RIF 1 ball types.
Across the tested velocity range there were two distinct groups of ball types (higher severity and lower severity). The higher severity group consisted of the Pro, HS/Col, Youth, and RIF 10 balls, while the lower severity group contained the RIF 5 and RIF 1 balls. The youth ball produced the highest skull fracture risk and concussion risk across the tested velocities and was most similar to the Pro ball. At 30 m/s the youth ball produced a scaled PLA of 272 g, a skull fracture risk of 58% and a concussion risk of 99%, while the Pro ball produced a scaled PLA of 262 g, a skull fracture risk of 50%, and a concussion risk of 99%. The HS/Col ball and the RIF 10 ball were also similar to one another across velocities, with the only major difference in skull fracture risk between 25 m/s and 30 m/s. The HS/Col ball had higher risk values, 5% and 38% for these two velocities, 25 m/s and 30 m/s, compared to the RIF 10’s risk values of 3% and 9% respectively. The difference in risk at 30 m/s corresponded to a difference in linear acceleration of 49 g. For all ball types in the higher severity group a concussion risk of approximately 100% was achieved at 30 m/s. The balls in the lower severity group were almost identical to one another, yielding 0% skull fracture risk across the velocity range and a maximum concussion risk value of 58% at 30 m/s. This risk value corresponded to a scaled PLA of 121 g. The rotational accelerations measured provided no differences by ball type across the range of velocities.

Scaling the accelerations of each ball type for their respective age group had little effect on the risks, since the maximum increase for linear acceleration was 10% and the maximum increase for rotational acceleration was 22%. The scaling process slightly increased the magnitude of the risk functions, but did not affect the overall trend of these data. As a result of scaling, the risks at age matched velocity, or the average plate velocity each age group encounters, was able to be analyzed. When analyzing skull fracture risk and concussion risk for each ball type at age matched velocities, it was found that risk decreases by age group, as would
be expected. However, even at age matched velocity the concussion risk for the 13-14 year olds (Youth) only decreased 5% (100% to 95%). In addition, the skull fracture risk increases by approximately 25% between the 9-12 age group (RIF 10) and 13-14 (Youth) age group, which is the greatest increase from one age group to the next.

While the study was able to evaluate the effect that different ball types had on skull fracture and concussion risk, there are a few limitations that should be acknowledged. First, the use of length scaling to scale the head accelerations from the Hybrid III to the respective head size of each age group assumes constant mass density and modulus of elasticity between subjects. In addition, only one impact location was tested, the forehead. Different locations of the head will have different curvatures that could affect the head injury risk of each ball type. Another limitation is that PLA is only a correlate for head injury mechanisms. Also, the concussion risk function utilized is based off of adult concussion parameters. There is evidence to suggest that concussion risk is different in youth athletes, so a youth specific concussion risk function could alter these results. Additionally, the impact duration of these impacts were between 1-2 ms, which is significantly shorter than the duration seen in football impacts, 10-15 ms. This difference in loading rate could affect the thresholds and parameters used for estimating concussion risk and would provide the need for the development of a loading rate specific risk function.

CONCLUSIONS

Different stiffness baseballs are used for different age groups/ levels of play in order to reduce injury risk. To test the effective skull fracture risk and concussion risk associated with each ball type, baseballs were projected into the forehead of a Hybrid III head for a range of velocities, 15-30 m/s, that resemble real world scenarios. Peak linear acceleration was used to
calculate the skull fracture risk and it was found that the youth ball produced the highest risk. However, when analyzing the age specific risk it was found that the skull fracture risk decreased from the highest to lowest level of play, which was expected. Concussion risk was also analyzed using peak linear and peak rotational accelerations. Similar to skull fracture risk, the youth ball yielded the highest concussion risk across the velocity range, but when considering the concussion risk at age matched velocity it was found that the professional, collegiate, and high school levels had the highest concussion risk.

The information provided in this study can be used to further improve age specific baseball design. Future studies can be used to further determine the injury risk associated with different baseball types and brands, which when coupled with improved knowledge of human tolerance for high loading rate head impacts, such studies can help to improve injury prevention in baseball.

REFERENCES

8 NOCSAE. Standard performance specification for newly manufactured baseballs. 027-12m17a, 2017.
CHAPTER 4 - DO INFIELD SOFTBALL MASKS EFFECTIVELY REDUCE FACIAL FRACTURE RISK?

ABSTRACT

Softball has a higher injury rate than baseball with a majority of injuries occurring while fielding batted balls. Infielder masks exist and are intended to reduce head injury risk, but are rarely worn. The objective of this study was to evaluate the effectiveness of infielder masks’ ability to attenuate facial fracture risk over a range of designs and materials. To simulate batted ball impacts, a customized pitching machine was used to propel softballs at the average batted ball velocity of female high school players, 24.6 m/s. The balls impacted locations centered over the maxilla and zygoma bones of a FOCUS headform. The headform was attached to a 50th percentile Hybrid III neck and secured to a 16-kg sliding table. The FOCUS headform is equipped with 10 tri-axial titanium force plates: the eyes, the right and left maxilla, the right and left zygoma, the right and left frontal bone, the nasal bone, and the mandible. Facial fracture risk of each facial bone was compared between masks and impact locations using peak resultant forces. Analysis of these data show that the mask material and the distance between the mask and the impacted facial bone were key factors in determining a mask’s performance. It was found that a metal mask with a distance greater than or equal to 35 mm away from the maxilla and a distance of 25 mm or more away from the zygoma was effective at reducing facial fracture risk. Plastic masks performed worse because they excessively deformed to allow ball contact with the face, generating high forces. This study provides framework for determining the optimal infield mask design for reducing facial fracture.

Keywords: head impacts, softball, facial fracture, injury, biomechanics
INTRODUCTION

Approximately two million girls between the ages of 12 and 18 play softball annually. Compared to baseball, softball has a greater overall injury rate, especially because the injury rate while fielding batted balls is higher. This is likely due to the difference in field size between baseball and softball. Softball fields are smaller than baseball fields. For comparison, a regulation baseball field’s pitching mound is 60.5 feet away from home plate and there are 90 feet in between each base, while a softball field’s pitching mound is only 43 feet away from home plate with 60 feet in between the bases. The reduction in field size decreases the amount of time fielders have to react to a batted ball and increases their chances of sustaining an injury due to ball impact. One of the most common injuries from ball impact is facial fracture. The highest percentage of facial fractures occur in the midface region (zygoma, orbital, nasal, and maxilla), so this region was the area of interest for this study.

The United States Consumer Product Safety Commissions (CPSC) reported that 36% of all baseball and softball injuries could be prevented, or reduced in severity with the use of safety equipment. Since 2006, it has been a requirement that all softball batters wear a batting helmet with a facemask, however, infielder masks are still not required in the sport. It is difficult for high school and collegiate leagues to mandate the use of infielder masks because the National Operating Committee on Standards for Athletic Equipment (NOCSAE) does not certify them. NOCSAE has not developed a standard for a facemask only device because they believe the risk of getting hit in the face by a batted ball is the same as getting hit in the head by a batted ball. In order for NOCSAE to certify a mask of this type, it would have to be heavily padded and have a shell that encompasses the head, similar to a catcher’s mask.
There is a lot of discussion about requiring infield masks in softball, and some states have already required certain positions to use an infielder’s mask during high school play, but there is still a lack of data supporting the facemask’s ability to reduce injury. The objective of this study was to evaluate the effectiveness of infielder masks’ ability to attenuate facial fracture risk. A better understanding of these masks’ performances will help determine the benefits of wearing an infielder’s mask and aid in optimizing mask design.

METHODS

To simulate a batted softball being driven toward an infielder’s face, softballs were projected using a softball pitching machine. The dual wheeled, electric motor-driven machine (Jugs Sports Combination Pitching Machine Model SR3616-681-7, Tualatin, OR) was customized and anchored to the floor to reduce unwanted vibration. Each wheel had an independent speed dial with digits ranging from 0 to 100 that could be set to acquire the desired speed. The wheels were pressurized to 17 psi and their speed dials maintained at least a 35 digit offset from one another to prevent the softball from knuckling, as specified in the manual. In order to minimize potential differences in softball positioning during loading, custom ball holders were constructed to create a homogeneous loading orientation for all impacts. The softball velocity was calculated over the final 10.16 cm of the softball mount using a dual laser velocity gate sensor (Velocity Timer Model 1204, KME Company, Troy, MI) and was repeatable within ± 6% of the desired velocity. The customized pitching machine possessed an impact location accuracy within a 0.635 cm radius circle, which was verified from a previous study using the same machine. The softballs used to test the infield masks were 12 inches in circumference, weighed 7.0 ounces, and were manufactured by Rawlings (model C12RYLAH). The infield masks tested were the All-Star: Vela, Bangerz: HS-6500, Champro: The Grill,
Defender Sports: Defender Sports Shield, Markwort: Game Face (large and medium), Rawlings: Fielders Mask, Schutt: Fielders Guard, and Schutt: Titanium Fielders Guard (Figure 4.1). These infielder masks were chosen because they represented a range of designs and materials used for infielder masks on the market.

Figure 4.12. Infield masks tested: Top row left to right - All-Star: Vela, Bangerz: HS-6500, Champro: The Grill. Middle Row left to right - Defender Sports: Defender Sports Shield, Markwort: Game Face Large, Markwort: Game Face Medium. Bottom row left to right – Rawlings: Fielders Mask, Schutt: Fielders Guard, Schutt: Titanium Fielders Guard. The infield masks tested represent a subset of the infield masks on the market, but encompasses various mask designs and materials used. The materials used for infielder masks are polycarbonate, steel, and titanium.
Facial fracture risk for each infield mask was evaluated using the response of a surrogate headform. A Facial and Ocular Countermeasures for Safety (FOCUS) headform and a 50\textsuperscript{th} percentile male Hybrid III neck were affixed to a 16 kg sliding table that mimicked the inertial properties of the upper torso. The FOCUS headform is equipped with ten tri-axial titanium force plates that measure the loading patterns of the eyes and facial bones. The facial bones modeled in the FOCUS headform are the right and left frontal bone, the right and left zygoma, the right and left maxilla, the nasal bone, and the mandible.\textsuperscript{7} For this study, the eye data were excluded and only the facial bone data were collected. Data acquisition was conducted using a TDAS Slice Pro (DTS, Seal Beach, CA) system with a sampling rate of 20 kHz.

The two locations chosen for impact testing were centered over the maxilla (Location A) and the zygoma (Location B) of the headform, since they were bones that are commonly fractured from softball impact.\textsuperscript{1} Locations are referenced from a zero location on the FOCUS headform. The zero location is when the FOCUS headform is centered in front of the launcher with no rotations, and set so the middle of the muzzle is located at the tip of the nose. Both locations are 35.56 cm away from the launcher and tilted 10 degrees forward, towards the launcher. For the maxilla impact location the y and z translations on the slider table from the zero location were -1.5 cm and -0.5 cm respectively and the headform was rotated -15 degrees about the z axis. For the zygoma impact location, the y and z translations from the zero location were 1.5 cm and 1.5 cm respectively and the head was rotated -55 degrees about the z-axis. These translations and rotations correspond with SAE J211 coordinate system of the head. The experimental setup and impact locations are depicted in Figure 2.14.2.

The masks were tested at an impact speed equivalent to the average batted ball speed of female high school softball players, 24.6 m/s.\textsuperscript{10} This speed corresponded to the right wheel set to
67 on the speed dial and the left wheel set to 27 on the speed dial. Four masks of each type were tested. Two of the four masks were used for the maxilla location (A), while the other two were used to test the zygoma location (B). Each mask was only impacted once, totaling two trials at each location. Because there were no fitting directions from the manufacturers, each mask was positioned on the head using best judgment. The chin pad was positioned at the base of the chin and the forehead padding was adjusted to lie superior to the eyes. The facemask was centered by making sure the headform had a clear line of sight and was not tilted toward one side or the other. The straps were adjusted to assure a snug fit around the headform, preventing any mask from falling off the headform during testing.

A Phantom high speed camera (Vision Research, Wayne, NJ) was positioned perpendicular to the balls trajectory to ensure correct impact location. The Phantom camera was set to a sampling rate of 1000 fps in order to capture the entirety of the impact. Prior to testing, the distance between the masks and the impacted facial bones were measured using a dial caliper to analyze the effect of mask distance on facial fracture risk.

A bare headform forehead impact was also conducted at 24.6 m/s on a Hybrid III headform as a reference for the force experienced by a player not wearing a mask. Using the conservation of momentum, Equation 1, the force was back calculated.

\[ m\Delta v = \int F dt \]  

In Equation 1, \( m \) is the weight of the softball, \( \Delta v \) represents the change in ball velocity from before to after the impact, \( F \) is the force experienced by the headform, and \( dt \) symbolizes the duration of the impact. The duration of the impact was acquired from the linear resultant acceleration pulse of the TDAS Slice Pro system and the change in velocity was obtained using the Phantom high speed camera.
Data collected were processed according to SAE J211 and filtered using channel frequency class (CFC) 300.\textsuperscript{5} Peak resultant force for each facial bone was calculated for each test. Facial fracture risk was calculated using the nonparametric model developed by Cormier et al.\textsuperscript{6,11}

![Image](image_url)

Figure 4.213. Top: Anchored customized pitching machine that projects softballs into a FOCUS headform attached to a 50th percentile male Hybrid III neck that is mounted to a 16-kg sliding table. Bottom Left: Maxilla impact location (A) with translations and rotations from the reference location. Bottom Right: Zygoma impact location (B) with translations and rotations from the reference location.

To determine if the impact locations, the mask material, or the distance between the mask and the impacted facial bone had an effect on mask performance, an ANCOVA was conducted using JMP Pro 13 (SAS, Cary, NC). The log transformation of the peak resultant force was used
as the response variable in the ANCOVA because the risk values were zero heavy data and generated a non-normal distribution. Mask material, the distance between the mask and the impacted facial bone, and the impact location were used as predictors in the analysis. A p-value of \( \leq 0.05 \) was considered significant.

**RESULTS**

Table 4.1 displays the average force and average non-parametric fracture risk of the nasal bone, the right maxilla, the right zygoma, and the right frontal bone for each mask at each impact location. The left maxilla, left zygoma, and left frontal bone were not included because they did not produce any fracture risk and the mandible was excluded because its risk values were relatively low.

Forces on facial bones varied by mask and location. The right zygoma and the right frontal bones yielded the highest forces during a zygoma impact (B) for the tested masks (Figure 4.3). However, there was a lot of variance in force within these impact configurations, indicating that some masks reduce force in these areas better than others. The forces tended to be greater in facial bones that were closer to the impact location than facial bones on the contralateral side of the impact.

The average non-parametric fracture risk also varied by mask and location, but produced very similar trends to the average force (Figure 4.4). The right zygoma during a zygoma impact (B) generated the highest fracture risk and facial bones on the contralateral side of the headform than the impact sustained zero fracture risk. Overall, the masks were successful at reducing facial forces and facial fracture risk. For comparison, a player not wearing an infield mask would encounter forces around 11,604 N, which was determined though a bare headform Hybrid III
test. Even at impact configurations that were the most severe, the variance in fracture risk and force was wide enough to support the claim that wearing a mask reduces fracture risk.

Table 4.5. Average force and average non-parametric fracture risk of the nasal, right zygoma, right maxilla, and right frontal bone for each mask at each impact location. The Bangerz mask sustained the highest fracture risk value for the right maxilla during a maxilla impact (A) and zygoma impacts (B) yielded the highest fracture risk value for the right zygoma across all masks. The left maxilla, left zygoma, and left frontal bones produced zero fracture risk and the mandible had relatively low fracture risks. Location A is a maxilla impact and location B is a zygoma impact.

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<td>0.00</td>
<td>50.52</td>
<td>0.00</td>
<td>124.54</td>
<td>0.00</td>
<td>1104.96</td>
<td>0.13</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>118.62</td>
<td>0.05</td>
<td>1357.32</td>
<td>0.87</td>
<td>125.01</td>
<td>0.00</td>
<td>1691.55</td>
<td>0.30</td>
</tr>
</tbody>
</table>
Figure 4.314. Illustrates the average force experienced by each facial bone across all mask types and location. The right zygoma and the right frontal bone experienced the largest forces during a zygoma impact (B). The bones contralateral to the impact yielded relatively low forces. Location A is a right maxilla impact and location B is a right zygoma impact.

Figure 4.415. Displays the average non-parametric fracture risk seen by each facial bone across all mask types and by impact location. The right zygoma during a zygoma impact (B) generated the highest fracture risk and facial bones contralateral to the impact produced no fracture risk. Location A is a right maxilla impact and location B is a right zygoma impact.
Each mask tested possessed its own unique design. Table 4.2 displays the differences by the mask material and by the distances between the facemask and the impacted facial bone. Mask material encompassed polycarbonate, steel, and titanium, but was dichotomized so that all polycarbonate masks were classified as plastic and steel and titanium masks were classified as metal.

Table 4.6. Displays the material of each mask and the distance between the mask and the headform at the maxilla and zygoma. The material category “metal” includes both steel and titanium masks and the material category “plastic” represents polycarbonate masks. Plastic masks had greater distances between the mask and the maxilla when compared to metal masks.

<table>
<thead>
<tr>
<th>Mask</th>
<th>Material</th>
<th>Zygoma Distance (mm)</th>
<th>Maxilla Distance (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>All-Star</td>
<td>Metal</td>
<td>23.30</td>
<td>46.62</td>
</tr>
<tr>
<td>Bangerz</td>
<td>Metal</td>
<td>16.35</td>
<td>26.47</td>
</tr>
<tr>
<td>Champro</td>
<td>Metal</td>
<td>29.50</td>
<td>45.00</td>
</tr>
<tr>
<td>Defender Sports</td>
<td>Plastic</td>
<td>19.77</td>
<td>47.29</td>
</tr>
<tr>
<td>Markwort Large</td>
<td>Plastic</td>
<td>24.97</td>
<td>61.74</td>
</tr>
<tr>
<td>Markwort Medium</td>
<td>Plastic</td>
<td>16.48</td>
<td>55.10</td>
</tr>
<tr>
<td>Rawlings</td>
<td>Metal</td>
<td>22.63</td>
<td>40.13</td>
</tr>
<tr>
<td>Schutt Steel</td>
<td>Metal</td>
<td>22.90</td>
<td>35.80</td>
</tr>
<tr>
<td>Schutt Titanium</td>
<td>Metal</td>
<td>23.04</td>
<td>35.11</td>
</tr>
</tbody>
</table>

An ANCOVA between the log force, the mask material, the impact locations, and the distance between the mask and the impacted facial bone was conducted. The parameter estimates table was used to determine if distance, material, and impact location were contributing factors to mask performance. Distance and material produced p-values of 0.0353 and 0.0021, showing that both distance and material effect mask performance. Impact location yielded an insignificant p-value (0.6931) showing that there was insufficient sample evidence to suggest that impact
location had an effect on mask performance. There may be collinearity between impact location and distance that explains why impact location is not significant.

Figures 4.5 and 4.6 display the relationship between the average force and the distance between the mask and the impacted facial bone for each mask material (metal and plastic). Figure 4.5 presents the relationship between the average force and the distance by material for the right maxilla during a maxilla impact (A). Regardless of material, masks with a maxilla distance greater than 35 mm above the impacted facial bone resulted in an average force under 1,000 N. Figure 4.6 illustrates the relationship between the average force and the distance by material for the right zygoma during a zygoma impact (B). Metal masks with a distance greater than 25 mm above the zygoma yielded average forces less than 1,000 N. More mask samples provide a better understanding of the relationship between the average force and distance from the mask to the impacted facial bone for each mask material. Since force is a predictor of fracture risk, it is likely that fracture risk is affected by the same variables (material and distance from the mask to the face).
Figure 4.5.16. Shows the average right maxilla force as a function of the distance between the mask and the maxilla for each mask material at the maxilla impact location (A). The linear regression line for the metal masks displays a negative correlation with distance for a maxilla impact (A). The linear regression line for the plastic masks shows that there is little correlation between mask distance and average force for plastic masks in maxilla impacts (A).

Figure 4.6. Displays the average right zygoma force as a function of the distance between the mask and the zygoma for each mask material at the zygoma impact location (B). The linear regression line for the metal masks displays a steep negative correlation with distance for a zygoma impact (B). The linear regression line for the plastic masks shows that there is little correlation between mask distance and average force for plastic masks in zygoma impacts (B).
DISCUSSION

Evaluating the performance of infield masks using the FOCUS headform showed that masks do effectively reduce facial fracture risk. If a player was not wearing an infield mask, they could experience facial forces around 11,604 N. With infield masks, no facial forces exceeded 3,000 N (74% reduction in force) and a majority of forces were below 1,000 N (≥ 91% reduction in force). Impacts to the maxilla (A) yielded much lower forces and fracture risks than impacts to the zygoma (B), which produced the highest forces and fracture risks of all facial bones. For zygoma impacts (B), the right zygoma and the right frontal bone sustained the most severe forces and fracture risk values. However, even at these severe impact configurations the variance in force within the tested mask set varied from 762 N to 2,953 N for the right zygoma during a zygoma impact (B) and 663 N to 2,749 N for the right frontal bone during a zygoma impact (B). This corresponded to a fracture risk range from 59% to 100% for the right zygoma and a fracture risk range from 0% to 64% for the right frontal bone. These variations in facial force and fracture risk suggest that mask performance is different based on different designs, and if the proper mask is worn facial fracture risk can be immensely reduced.

Looking into the design aspect of infield masks, it was found that mask material and the distance from the mask to the impacted facial bone were significant factors in mask performance. For all masks, the zygoma location (B) had a smaller distance between the mask and the headform than the maxilla location (A), supporting the finding that a greater distance leads to a greater reduction of force. From analysis during a maxilla impact (A), if the distance was greater than 35 mm, the average force experienced was less than 1,000 N, regardless of mask material. However, a majority of the metal masks were able to reduce the average force more than plastic masks at this distance. During a right zygoma impact (B), if the distance for a metal mask was
greater than 25 mm the average force was reduced to under 1,000 N. Distance did not seem to
effect the average force at the zygoma for plastic masks.

It is believed that plastic masks performed worse than metal masks, even though their
distances between the mask and the impacted facial bones were greater than or equal to metal
masks, because the material properties of the plastic masks allowed significant intrusion. High
speed footage depicted that as the softball engaged plastic masks, the masks deformed to the
point where the ball contacted the headform (Figures 4.7 and 4.8). If the ball is still able to
contact the head through a mask, a higher amount of energy will be transferred into the head,
instead of being dispersed to the mask, generating greater forces on facial bones. None of the
masks broke upon impact from the ball, but there was permanent deformation of the metal masks
at the impact site. From these data, the best infield mask to reduce facial fracture risk seems to be
a metal mask that has a clearance distance greater than 35 mm at the maxilla and 25 mm at the
zygoma.

Figure 4.7. Shows the mask deformation comparison between metal and plastic for a maxilla impact (A). The left
two pictures show the Champro mask (metal) before impact and during maximum ball intrusion. The right two
pictures illustrate the Markwort Large mask (plastic) before impact and during maximum ball intrusion. The plastic
mask deforms much more than the metal mask allowing the softball to contact the face.
While the study was able to evaluate the effect infielder masks had on facial fracture risk, there are a few limitations that should be acknowledged. First, force is only a correlate for predicting facial fracture. Knowing the force and the area engaged by the ball during impact would allow for pressure to be calculated, which is a better predictor of facial fracture. Furthermore, bare headform reference testing was not able to be conducted on the FOCUS headform for a direct comparison because of the possibility of breaking instrumentation. Another limitation is that the fracture risk calculated may not accurately represent the fracture risk of female high school softball players, since the model was developed using data from male cadavers ranging in age from 41 to age 94. However, by using male cadavers in this age range to establish fracture risk thresholds, it creates a more conservative fracture risk curve because their bones are likely weaker than female high school athletes. In addition, the FOCUS headform is only made in one size, which may not accurately represent the head size of a female softball player. Finally, the number of trials and sample size were small. Conducting more trials would allow standard deviations to be calculated for each location. Adding other plastic and titanium masks to the sample would allow for a better analysis of the relationship between the average force and the distance between the mask and the impacted facial bone and enable additional analyses to be done to determine if the type of metal has an effect on mask performance.
CONCLUSION

Infielder masks are used to help reduce facial fracture risk in softball. To test if these masks can effectively reduce facial fracture risk, softballs were projected into the maxilla and zygoma of a FOCUS headform at 24.6 m/s, representing the average batted ball speed for female high school softball players. Peak force was used to calculate facial fracture risk for each facial bone at both impact locations using a previously developed nonparametric risk model. It was found that infield masks do effectively reduce facial fracture risk. Mask material and the distance between the mask and the impacted facial bone were significant predictors of mask performance. Analysis of these factors justifies that a metal mask with a distance of 35 mm or more above the maxilla and a distance of 25 mm or more above the zygoma is the optimal mask design.

Although these masks did not eliminate the risk of facial fracture, they did reduce it. These data show that infield masks do effectively mitigate facial fracture risk and should be used to help prevent tragic injuries that could lead to facial reconstructive surgery, or in some cases death. Future studies can be conducted to determine if head accelerations are reduced while wearing an infielder’s mask, which, when coupled with more trials and a greater sample size, can help to improve injury prevention in softball.

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CHAPTER 5 - BASEBALL AND SOFTBALL STAR: A METHODOLOGY FOR ASSESSING THE BIOMECHANICAL PERFORMANCE OF BATTER’S HELMETS AND CATCHER’S MASKS

ABSTRACT

Batter’s helmets and catcher’s masks are used in baseball and softball to prevent serious head injuries. This paper presents a novel methodology for evaluating the ability of batter’s helmets and catcher’s masks to mitigate concussion risk. Helmets are evaluated using a STAR formula that considers linear and rotational accelerations to determine injury risk. An exposure value of one is used because head impacts in baseball and softball are rare events. Batter’s helmets and catcher’s masks are impacted at three different locations at two different velocities. Different ball types are used at each velocity so age-specific STAR ratings can be achieved. Age-specific STAR ratings will be broken into two age groups: a 14 and under age group and a high school-college age group. Each helmet model tested will be assigned two STAR ratings, one for each age group. STAR ratings are then disseminated to the public as a purchasing tool for consumers.

Keywords: head impacts, baseball, softball, head injury risk, batting, catching, concussion, biomechanics

INTRODUCTION

It is estimated that there could be up to 3.8 million sports related traumatic brain injuries (TBIs) that occur each year. More than 19 million children participate in youth baseball and softball annually and although baseball and softball are not commonly depicted as contact sports, according to the U.S. Consumer Product Safety Commission (CPSC) baseball and softball were responsible for 11.6% of all head injuries treated in emergency rooms in 2009; third most behind...
only cycling and football.\textsuperscript{4,18} Ball impact has been identified as the leading cause of injury in baseball and softball, with the most common injury resulting from a ball to the head.\textsuperscript{8} Ball to head impacts cause 25\% of the annual deaths associated with youth baseball and softball, which have the highest fatality rate of all sports between the ages of 5-14.\textsuperscript{10} Use of a catcher’s mask starting in the 1880s, and the requirement of batting helmets in the early 1950s greatly improved the safety of the sport by reducing fatalities and severe head injuries.

Batter’s helmets and catcher’s masks must be certified by the National Operating Committee on Standards for Athletic Equipment (NOCSAE) prior to their use in the field of play. NOCSAE certifies baseball and softball helmets using a pneumatic cannon to project balls into a rigidly mounted NOCSAE headform and a twin wire drop tower for drop testing. NOCSAE’s test standard for batter’s helmets requires three samples of each helmet model.\textsuperscript{12} The helmets are positioned on a medium NOCSAE headform that is positioned less than or equal to 24 inches away from the muzzle of the pneumatic cannon. Two of the helmet samples are impacted with softballs at 55 mph in five locations (front right, front boss, right side, right rear boss, and rear), then the third helmet sample is impacted twice with a baseball at 60 mph. For the baseball impact, one impact is at the location that produced the highest Severity Index (SI) during the softball test sequence and the other impact is at a random location that is selected to exploit any weaknesses of the helmet. The SI threshold is not allowed to exceed 1,200 during any of the impact tests.

Catcher’s masks undergo a different certification from NOCSAE that requires seven samples of each model.\textsuperscript{13} Of the seven models, two will be drop tested, two will be impacted with a softball, one will be impacted with a baseball, and all seven will be subjected to faceguard testing. Drop tests are conducted using a twin wire drop tower that drops helmets onto a half
cylinder steel anvil in a right side, right rear boss, rear, and random orientation. Projectile softball tests are conducted at 55 mph, striking the masks on the right side, right rear boss, rear, and a random location. The projected baseball is tested at 60 mph. The baseball impact occurs at the location that produced the highest SI from the softball testing, as well as at a random location that is selected to exploit any flaws of the helmet. For faceguard testing, five of the helmets are impacted once with a baseball at 70 mph in either a vertical, vertical with a 45° midsagittal rotation, or a random orientation to exploit design weaknesses. The remaining two helmets are impacted with a softball at 70 mph at any location. In order to achieve NOCSAE certification, catcher’s masks must sustain a peak SI below 1,200 during all tests and prevent any contact between the mask and ocular area.

The objective of this study is to describe the methodology of a new evaluation system for batter’s helmets and catcher’s masks. This evaluation system will provide a quantitative metric for the ability of batter’s helmets and catcher’s masks to reduce the risk of concussion. Like previous Summation of Tests for the Analysis of Risk (STAR) evaluation systems, Baseball and Softball STAR will define realistic impact conditions to test in the laboratory and disseminate the results to the public so consumers have a meaningful metric to use when purchasing protective safety equipment.\textsuperscript{16,17}

\section*{METHODS}

\textit{Baseball and Softball STAR Equation}

The STAR equation (Eq. 1) being used to evaluate the differences in protective headgear worn in baseball and softball is very similar to the STAR equation used in Hockey STAR, which evaluates risk by considering both linear and rotational accelerations.\textsuperscript{16}

\begin{equation}
Baseball and Softball \text{ STAR} = \sum_{L=1}^{3} \sum_{V=1}^{1} E(L, V) \ast R(a, \alpha)
\end{equation}
In Baseball and Softball STAR, $L$ represents the impact location; $V$ represents the projectile velocity; $E$ represents the exposure as a function of location and velocity; and $R$ represents the risk of concussion as a function of linear ($\alpha$) and rotational ($\alpha$) head acceleration. This STAR equation is used to evaluate both batter’s helmets and catcher’s masks to produce age-specific STAR ratings.

The laboratory test matrices are comprised of 3 impact locations and 1 impact velocity for each age group, totaling 6 test conditions per helmet. For batter’s helmets, two helmets of each model are purchased and undergo the test matrix, resulting in 12 tests per helmet model. For catcher’s masks, four helmets of each model are purchased. Each mask is impacted at all three locations, but only for the specified velocity. Individual catcher’s mask locations are not tested at each age groups’ velocity because of the possible effects on mask performance caused by deformation of materials. Combined, catcher’s masks total 12 tests per mask model.

**Impact Locations**

The three locations for Baseball and Softball STAR are different for batter’s helmets and catcher’s masks. For batter’s helmets the STAR locations are the side, top, and rear boss of the helmet (Figure 5.1) and for catcher’s mask’s the STAR locations are the central nose, lateral eyebrow, and offset forehead (Figure 5.2). STAR locations were selected based on real world injuries.\(^6\)\(^7\) Batter’s helmet impacts are primarily to the side of the helmet because of the natural orientation of the batter. Trichotomizing helmet impacts of clinically diagnosed concussion yielded the three impact locations (side, top, and rear boss).\(^6\) Catcher’s masks are predominately impacted on the front of the mask as a result of the position’s requirements. Previous studies determined the most frequently impacted locations on the mask, which aided in the selection of the STAR locations (central nose, lateral eyebrow, and offset forehead).\(^7\) In addition to being
commonly impacted areas, these locations were selected to have a separation distance of at least one diameter of a softball to avoid the effects of structural deformation that is caused by the prior impact.

Figure 5.19. STAR locations for batter’s helmets. Each location was selected by aggregating clinically diagnosed concussions in the literature. The headform is translated and rotated for each location to represent a realistic head position during impact.

Figure 5.20. STAR locations for a traditional style catcher’s mask (left) and a hockey style catcher’s mask (right). The impact locations are the central nose, lateral eyebrow, and offset forehead. The headform is translated and rotated to resemble realistic head positions for catcher during an impact.
Ball Type and Test Velocity

The velocity for Baseball and Softball STAR is dependent on the age group being tested. For age-specific STAR ratings, helmets need to be tested at age-specific velocities and with a ball type that corresponds to that age group. Baseball and Softball STAR will be split into two age groups: 14 years of age and under and high school through college. The balls used for testing were all manufactured by Rawlings and certified by either NOCSAE or the American Softball Association (ASA). For baseball specific testing the ball models and velocities used for the two age groups were the RDYB1 (Youth) ball at 55 mph for ages 14 and under and the R100-H2 (HS/Col) ball at 75 mph for the high school-college age group. The Youth ball was selected for the 14 and under age group because previous testing of baseball stiffness showed that it displayed similar results to other youth specific balls, but was the most severe of the youth specific baseballs tested. The HS/Col ball was chosen for the high school and collegiate age group because it is the most common ball model used for this age group.

The ball models and velocities used for softball specific testing are the R12RYSA (RIF 10) ball at 45 mph for ages 14 and under and the C12RYLAH (Standard) ball at 55 mph for the high school and collegiate age group. Like baseball, the softballs chosen for testing were selected based on their popularity of use in their respective age groups. The impact velocities for baseball and softball were chosen because they were the average pitch velocities experienced by players of each age group. The sport specific velocities tested are the same for batter’s helmets and catcher’s masks, even though catcher’s mask impacts are typically slower than pitch velocity because of the energy lost during a foul tip. The change in ball velocity from a foul tip has proven difficult to quantify so, catcher’s mask impacts are conducted at the same velocities as batter’s helmets to represent a worst case scenario.
Exposure and Risk

The exposure value for Baseball and Softball STAR is assumed to be one. This is because head impacts in baseball and softball are rare events and there is currently no field data to suggest that impacts occur to one location more than the others. The incidence of concussion is acquired by multiplying the exposure value by the risk values, which is calculated using the bivariate risk function (Eq. 2). Incidence of concussion is summed at all the STAR locations to obtain a STAR value.

\[
R(a, \alpha) = \frac{1}{1+e^{-(a+0.0453+\alpha+0.00875-0.000000301)}}
\] (2)

Baseball Helmet Impact Device

To simulate on field impacts of a ball being pitched toward a batter’s head, or a foul tip off a bat redirecting into a catcher’s mask, a pitching machine is used to project balls in the laboratory (Figure 5.3).\textsuperscript{11} The dual wheeled, electric motor-driven machine (Jugs Sports Combination Pitching Machine Model SR3616-681-7, Tualatin, OR) was customized and anchored to the floor as a way of reducing unwanted vibration. Each wheel possesses an independent speed dial with digits ranging from 0 to 100 that are set to produce desired speeds. The wheels are pressurized to 17 psi and their speed dials maintain at least a 35 digit offset from one another to prevent the ball from knuckling, as specified in the manual. In order to minimize ball loading differences, custom ball holders were constructed to load the balls into the pitching machine with the same orientation. The ball velocity is calculated over the final 10.16 cm of the muzzle using a dual laser velocity gate sensor (Velocity Timer Model 1204, KME Company, Troy, MI) and is capable of obtaining a velocity ± 3% of the desired velocity.\textsuperscript{7} The customized
pitching machine possesses an accuracy within a 0.635 cm radius circle, which was verified from a previous study using the same machine.7

Head injury risk for batter’s helmets and catcher’s masks are evaluated using the response of a surrogate headform. A Hybrid III, anthropomorphic test device (ATD), 50th percentile male head and neck are mounted to a 16 kg sliding table (Biokinetics, Ottawa, Ontario, Canada) to mimic the inertial properties of the upper torso. A Hybrid III headform is used instead of a NOCSAE headform because of the high frequency noise seen in a NOCSAE headform for this test setup during high loading rate impacts, which are associated with baseball and softball. Linear and rotational head accelerations are collected by instrumenting the Hybrid III headform with a nine accelerometer array (Endevco 7264B-2000, Irvine, CA) in a 3-2-2-2 configuration.15 Data acquisition is conducted using a TDAS Slice Pro (DTS, Seal Beach, CA) system with a sampling rate of 20 kHz.

![Anchored customized pitching machine that projects balls into a 50th percentile male Hybrid III head and neck system that is mounted to a 16-kg sliding table to simulate the effective mass of the torso during impact. The sliding table has 5 degrees of freedom, (translation in the x, y, and z axes and rotation about the y and z axes) allowing any location of the helmet to be impacted. Left: Batter’s helmet setup. Right: Catcher’s mask setup.](image)

In an effort to recreate a realistic head position for batter’s helmets, player’s head position during ball impact was analyzed through video analysis for each STAR location. Screenshotting players at this instance allowed for a side by side comparison of head positions in
the laboratory, optimizing the transformation for each impact location. Head positioning for catchers was also analyzed and recreated in the lab for a realistic impact scenario. \(^2\) Table 5.1 specifies the transformations for each batter’s helmet STAR impact location from the reference location and Table 5.2 specifies the transformations for each catcher’s mask STAR impact location from the reference location. The reference location is identified as the Hybrid III headform positioned directly in front of the launcher with no rotations and the center of the muzzle aligned with the tip of the nose.

Table 5.7. Transformations from the reference location to recreate realistic head positions for batter’s helmets. The reference location is defined as the Hybrid III headform positioned directly in front of the launcher with the center of the muzzle aligned with the tip of the nose. The polarity of the transformations are in accordance with SAE J211 head coordinate system.

<table>
<thead>
<tr>
<th>Location</th>
<th>X translation (cm)</th>
<th>Y Translation (cm)</th>
<th>Z Translation (cm)</th>
<th>Y rotation (°)</th>
<th>Z rotation (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Side</td>
<td>- 35.6</td>
<td>- 1.1</td>
<td>- 4.3</td>
<td>+ 10.0</td>
<td>- 80.0</td>
</tr>
<tr>
<td>Top</td>
<td>- 35.6</td>
<td>- 10.5</td>
<td>+ 2.1</td>
<td>+ 15.0</td>
<td>- 50.0</td>
</tr>
<tr>
<td>Rear Boss</td>
<td>- 35.6</td>
<td>- 11.3</td>
<td>- 5.4</td>
<td>+ 10.0</td>
<td>- 100.0</td>
</tr>
</tbody>
</table>

Table 5.8. Transformations from the reference location to recreate realistic head positions for catcher’s masks. The reference location is defined as the Hybrid III headform positioned directly in front of the launcher with the center of the muzzle aligned with the tip of the nose. The polarity of the transformations are in accordance with SAE J211 head coordinate system.

<table>
<thead>
<tr>
<th>Location</th>
<th>X translation (cm)</th>
<th>Y Translation (cm)</th>
<th>Z Translation (cm)</th>
<th>Y rotation (°)</th>
<th>Z rotation (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nose</td>
<td>- 35.6</td>
<td>+ 1.3</td>
<td>- 9.8</td>
<td>+ 10.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Eyebrow</td>
<td>- 35.6</td>
<td>+ 5.0</td>
<td>- 6.0</td>
<td>+ 10.0</td>
<td>0.0</td>
</tr>
<tr>
<td>Forehead</td>
<td>- 35.6</td>
<td>- 4.3</td>
<td>- 2.8</td>
<td>+ 10.0</td>
<td>0.0</td>
</tr>
</tbody>
</table>

Data Collection

Data collected are processed according to SAE J211. The acceleration data for computing linear accelerations are filtered using channel frequency class (CFC) 1000 and the acceleration data for computing rotational accelerations are filtered at CFC 180.\(^{15}\) A custom-written MATLAB (Mathworks, Natick, MA) code is used to calculate the desired metrics: peak resultant linear acceleration and peak resultant angular acceleration. The peak linear and peak rotational
accelerations are averaged between identical helmet models for each impact condition. The average accelerations are then used to calculate the STAR value for each helmet.

EXEMPLAR RESULTS

A batter’s helmet was used to demonstrate the age-specific STAR values in Baseball and Softball STAR. The helmet was impacted at three locations and at two different velocities. The impact velocities tested were 55 mph, representing the 14 and under age group, and 75 mph, representing the high school-college age group. This illustrative test differs from the actual Baseball and Softball STAR testing in that only one helmet was tested. In practice, two helmets of the same model would undergo the test matrix and their acceleration values would be averaged at each location before calculating risk.

In order to determine the age-specific STAR value, the Youth ball was used to impact the helmet during the 55 mph impacts and the HS/Col ball was used to impact the helmet during the 75 mph impacts. Linear and rotational acceleration pulses at the three impact locations (front, top, and rear) are displayed in Figures 5.4, 5.5, and 5.6. The impacts at 75 mph generated much higher linear and rotational accelerations than at 55 mph, as expected. The front and rear impact locations yielded similar acceleration values and the top impact location produced the least severe acceleration values.
Figure 5.422. Shows exemplar linear and rotational acceleration trends for a tested helmet at both 55 and 75 mph for the front impact location. The 55 mph profile illustrates the 14 and under age group and the 75 mph profile depicts the high school-college age group. The maximum linear and rotational acceleration values for each profile are used to determine the age-specific STAR value of the helmet.

Figure 5.5. Shows exemplar linear and rotational acceleration traces for a tested helmet at both 55 and 75 mph for the top impact location. The 55 mph profile illustrates the 14 and under age group and the 75 mph profile depicts the high school-college age group. The peak linear and peak rotational acceleration values for each profile are used to determine the age-specific STAR value of the helmet. The top impact location yielded the lowest peak acceleration values.
Figure 5.623. Shows exemplar linear and rotational acceleration pulses for a tested helmet at both 55 and 75 mph for the rear impact location. The 55 mph profile illustrates the 14 and under age group and the 75 mph profile depicts the high school-college age group. The maximum linear and rotational acceleration values for each profile are used to determine the age-specific STAR value of the helmet.

The peak linear and peak rotational acceleration values were extracted from the acceleration pulses to calculate the concussion risk value at each impact location using Eq. 2 (Tables 5.3 and 5.4). In order to obtain the STAR values for each impact location, the calculated risk was multiplied by the exposure value. Exposure was assumed to be one since baseball and softball impacts are rare events and no literature suggests that one location is impacted more than the others. STAR values were then summed across all locations for each velocity to obtain the age-specific STAR value. For the tested helmet, a STAR value of 0.021 was achieved for the 14 and under age group and a STAR value of 1.937 was achieved for the high school-college age group. STAR values will be used to determine the age-specific STAR rating of a helmet, ranging from 1 to 5 stars. Lower STAR values equate to a lower risk of concussion and a better performing helmet. Therefore, a low STAR value will correspond to a high STAR rating.
Table 5.9. Peak linear and peak rotational acceleration values were used to calculate risk for the 14 and under age group. The Youth ball was projected at 55 mph to obtain these values. This ball and velocity combination resembles the plate scenario players 14 and under would encounter. The risk is multiplied by exposure to obtain a STAR value, which is summed across locations to generate the age-specific STAR value of the helmet for players 14 and under.

<table>
<thead>
<tr>
<th>Velocity</th>
<th>Location</th>
<th>PLA</th>
<th>PRA</th>
<th>Risk</th>
<th>Exposure</th>
<th>STAR</th>
</tr>
</thead>
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<tr>
<td>55</td>
<td>Front</td>
<td>72.2</td>
<td>2691</td>
<td>0.007</td>
<td>1.00</td>
<td>0.007</td>
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<tr>
<td></td>
<td>Top</td>
<td>27.0</td>
<td>1800</td>
<td>0.001</td>
<td>1.00</td>
<td>0.001</td>
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<tr>
<td></td>
<td>Rear</td>
<td>60.3</td>
<td>4001</td>
<td>0.013</td>
<td>1.00</td>
<td>0.013</td>
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</table>

Table 5.10. Peak linear and peak rotational acceleration values were used to calculate risk for the high school-college age group. The HS/Col ball was propelled at 75 mph to obtain these values. This ball and velocity combination resembles the plate scenario players in high school and college would experience. The risk value is multiplied by the exposure to obtain a STAR value, which is summed across locations to generate the age-specific STAR value of the helmet for high school and college players.

<table>
<thead>
<tr>
<th>Velocity</th>
<th>Location</th>
<th>PLA</th>
<th>PRA</th>
<th>Risk</th>
<th>Exposure</th>
<th>STAR</th>
</tr>
</thead>
<tbody>
<tr>
<td>75</td>
<td>Front</td>
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<td>7063</td>
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<td>1.00</td>
<td>0.934</td>
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<td></td>
<td>Top</td>
<td>107</td>
<td>3128</td>
<td>0.044</td>
<td>1.00</td>
<td>0.044</td>
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<tr>
<td></td>
<td>Rear</td>
<td>177</td>
<td>8014</td>
<td>0.960</td>
<td>1.00</td>
<td>0.960</td>
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</table>

1.937

CONCLUSIONS

This paper outlines a novel methodology for comparing the performance of various batter’s helmets and catcher’s masks models. The foundation of these methods is rooted in Football and Hockey STAR, but is more similar to Hockey STAR because it utilizes linear and rotational acceleration to determine concussion risk. A pitching machine was modified for laboratory testing and a Hybrid III headform is used as a surrogate headform to minimize high frequency noise. The exposure value is assumed to be one since baseball and softball impacts are rare occurrences and no data suggests that one location is impacted more than others. Consumers will use the batter’s helmet and catcher’s masks STAR ratings as a tool for purchasing, which will provoke manufacturers to advance batter’s helmet and catcher’s mask design to reduce concussion risk.
REFERENCES

5 ASA. Asa bat and ball certification program. ASA:
14 NOCSAE. Standard performance specification for newly manufactured baseballs. 027-12m17a, 2017.
CHAPTER 6 – CONCLUDING REMARKS

RESEARCH SUMMARY

The research in this thesis explores the effectiveness of protective head equipment used in baseball and softball and their abilities to mitigate head injury risk. Comparison of two commonly used ATD headforms, the NOCSAE and Hybrid III, suggested that the modified NOCSAE headform should not be used when evaluating high velocity projectile head impacts because of resonance in the headform. Using the Hybrid III headform, age-specific baseball stiffness was evaluated on their effectiveness of reducing skull fracture risk and concussion risk. It was found that the youth ball yielded the highest head injury risk across all velocities, but when compared at age matched velocity it was third behind the professional and collegiate-high school balls.

A FOCUS headform was used to determine if infield softball masks effectively mitigate facial fracture risk. By impacting the zygomatic bone and maxilla bone at the average batted ball speed of a high school softball player, the data suggests that a metal mask with a clearance distance of 35 mm above the maxilla and 22 mm above the zygoma reduces facial fracture risk the best. Comparison to a bare headform control impact determined that infield masks do effectively reduce facial fracture risk.

A STAR methodology for comparing the performance of various batter’s helmets and catcher’s masks models was also developed. Using impact locations based on clinically diagnosed concussion and video analysis to determine head position, realistic impact configurations are used to evaluate helmets. Age-specific impact velocities and ball types are used in the STAR evaluation to yield age-specific STAR ratings for batter’s helmets and catcher’s masks.
Although the mechanisms of head injury in baseball and softball are still not well understood, the research in this thesis includes the most recent methods for evaluating protective head equipment. It is anticipated that the research in this thesis can be used as the framework for improving safety in baseball and softball. These studies help to determine the optimal age-specific baseball stiffness and suggests design modifications for optimizing infield mask design. In addition, age-specific STAR ratings will be disseminated to the public as a tool for purchasing batter’s helmets and catcher’s masks. Publicizing STAR ratings will provoke manufacturers to design batter’s helmets and catcher’s masks that reduce concussion risk.

**PUBLICATION OUTLINE**

Table 6.11. Outline of publication detail for chapters 2-5 presented in this thesis.

<table>
<thead>
<tr>
<th>Chapter</th>
<th>Title</th>
<th>Journal</th>
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<tr>
<td>2</td>
<td>Comparison Between the Hybrid III and NOCSAE Headforms During High Velocity Projectile Impacts</td>
<td>Plan to submit to <em>The Journal of Sports Engineering and Technology</em></td>
</tr>
<tr>
<td>3</td>
<td>Head Injury Risk Associated with Baseball Stiffness as a Function of Player Age</td>
<td>Accepted for submission at the 2018 Rocky Mountain Bioengineering Symposium</td>
</tr>
<tr>
<td>4</td>
<td>Do Infield Softball Masks Effectively Reduce Facial Fracture Risk?</td>
<td>Plan to submit to <em>Annals of Biomedical Engineering</em></td>
</tr>
<tr>
<td>5</td>
<td>Baseball and Softball STAR: A Methodology for Assessing the Biomechanical Performance of Batter’s Helmets and Catcher’s Masks</td>
<td>Virginia Tech Helmet Ratings</td>
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