The Etiology of Impact Related Concussions for Catchers and Umpires in Baseball

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Abstract

The information presented herein attempts to quantify the conditions surrounding concussive impacts from foul tips to the masks of catchers and umpires in baseball. Media reports of such occasions were researched on video and pitch speed data from the Pitch F/X system recorded to suggest speeds and locations at which impacts occur. To evaluate mask performance, a pneumatic-wheel, electric-motor driven pitching machine was utilized to shoot baseballs at the instrumented head of a Hybrid III dummy. Head accelerations were calculated from a 3-2-2-2 accelerometer array to allow for comparisons of linear and angular kinematics. 6 common masks (2-piece traditional-style and 1-piece hockey-style) were tested at 7 locations at 60 mph to determine the severity of each location. The center-eyebrow and chin locations were further tested at 84 mph. Speed and location data were used to evaluate a large sample of 25 masks to explore possible performance differences between manufacturer models, mask types and cage styles. The results of this study showed no significant difference between hockey-style and traditional-style mask performance. Titanium caged masks, although lighter than their steel counterparts, experienced higher linear accelerations. However, all masks experienced linear and angular accelerations well below commonly accepted injury thresholds. Yet, concussive injury has still occurred in the players and umpires that wear these masks. The work presented here can be used to help better understand these thresholds and influence the design, construction and evaluation of a new generation of masks that decrease the risk of concussions to the wearer.

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Chapter 1 – Introduction: Concussions in Major League Baseball Catchers and Umpires from Baseball Impacts

Opening Remarks

While much research has focused on concussions in populations of football players, very little work has delved into the sport of baseball to examine concussive injuries. While not generally seen as a contact sport, concussions are reported in baseball players and can lead to lost playing time. Specifically, catchers and umpires behind home plate experience numerous high energy impacts throughout the course of their careers. One mechanism of concussion for these positions is through a fouled ball that travels directly into the faceguard or helmet of the wearer. Various reports in the mass media have highlighted incidents of such occurances with catchers such as Mike Matheny, and umpires like Kerwin Danley. Both have experienced high speed impacts to the faceguard from baseballs, and have had to deal with the lingering effects ever since. Mike Matheny was forced to retire after symptoms from a concussion suffered when a foul ball struck his mask did not improve over time [1]. Kerwin Danley was knocked unconscious and off his feet when a pitched fastball was missed by the catcher, impacting him directly in the faceguard [2]. Their stories highlight the severity of these impacts and the longer term effects that can come about from them. This creates an opportunity to explore an area of sports injury biomechanics that is not currently well understood, with implications for a large population of people.

Research Objectives

To investigate the concussions caused by ball impacts to the head, a methodical approach was taken to understand the circumstances surrounding such phenomenon. The first section of this research investigates the speeds and locations with which catchers and umpires typically experience foul balls to the mask by analyzing real world scenarios where such injuries have occurred. In the second section of this work, these speeds and locations are employed to experimentally discover the most severe impact conditions that could realistically be experienced by a catcher or umpire during the course of a baseball game. The final section of this research applies the results of the first two sections to evaluate mask performance as a whole by testing a large sample of masks under actual conditions that have led to injury.

The goal of this work is to supply the information necessary to help elevate mask design and engineering to higher levels of safety and protection. Furthermore, the information presented herein can be used by athletes and coaches to help select the proper equipment for their needs.

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Chapter 2 - Analysis of Concussions in Major League Baseball Catchers and Umpires from Baseball Impacts

Abstract

Objective: To determine the locations and speeds with which catchers and umpires experience concussive impacts from a foul ball to the mask in baseball.

Design: Mass media reports were searched for incidents in which a catcher or umpire suffered a clinically diagnosed concussion or concussive symptoms from a foul ball directly to the mask in Major League Baseball. Such incidents were evaluated on video through MLB.TV and pitch data was extracted from the Pitch F/X database.

Setting: None.

Patients: None.

Interventions: None.

Main Outcome Measurements: The speed of a baseball as it crosses the plate, and the locations on a mask in which impacts occur.

Results: Ten concussive impacts were examined, four related to catchers and six related to umpires. Seven regions were suggested to represent the various locations at which impacts are experienced. Four of these locations fall along the mid-sagittal plane of the head, at points on the forehead, eyebrow, nose and chin. Three locations correspond longitudinally with the forehead, eyebrow and nose targets but three inches laterally. Velocity data from the Pitch F/X database indicates that the median impact speed is 84 mph.

Conclusions: This study presents seven locations and a target speed of 84 mph for further mask testing and evaluation of concussion risk. This information can be used to compare current product line performance for an understanding of differences (if any) between manufacturers, mask types and styles.

Key Words: Catcher Mask, Umpire Mask, Pitch F/X, Baseball Impact, Concussion

Introduction

Baseball and softball injuries have been estimated to cause more emergency room visits in the United States than any other sport.[1] The National Electronic Injury Surveillance System (NEISS) suggests that 26% of sports related head injuries come from baseball or softball.[2] Impact from a ball has been reported to be the leading cause of head injury in baseball and softball.[3] One mechanism of these impacts to the head occurs when a batter fouls a pitched ball directly backwards into the mask of the catcher or umpire. These impact scenarios have been documented to cause concussions for the mask wearer.[4] In order to investigate this mechanism of injury, a thorough understanding of the conditions surrounding this type of event is necessary. This includes knowing the typical range of speeds in which the ball hits the mask, as well as the locations at which impact occurs.

For a catcher or umpire mask to be certified for use, it must pass the National Operating Committee on Standards for Athletic Equipment (NOCSAE) specification for catcher's helmets with a faceguard.[5] The standard specifications require a mask to pass a drop tower test for the helmet and projectile tests for the faceguard. Faceguards are evaluated with baseballs and softballs at 70 mph to determine if contact between the cage and the face occurs during an impact. Impact location is chosen based on preset positions and any positions thought to exploit weaknesses in the cage. Pass/fail determination is made based on the faceguard's ability to avoid contact by the cage with a no contact area around the ocular region, and limited contact by non-structural components (such as padding) to an area around the mouth/nose. These tests are primarily designed to protect against skull and facial fracture, and do not address injuries such as concussion.

In a study designed to investigate the ability of a catcher mask to attenuate head accelerations from impacts, Shain et al. (2010) addressed concussions from fouled balls and how well masks reduced head accelerations.[4] Masks in this study were tested with impacts aimed at the nose of a Hybrid III dummy head, and speeds varied between 60 and 80 mph. Although shown to reduce head accelerations to levels significantly below accepted injury thresholds, masks have not eliminated the risk for concussions.

Recent advances in ballpark technology have allowed for expansion on the radar gun readings previously relied upon for pitch speed measurements. Radar guns cannot distinguish between the release speed of the ball from the pitcher's hand and the speed at which it crosses the plate. A more in depth measurement system known as Pitch F/X (Sportvision, Inc. *Mountain View, CA*), computes detailed information on each individual pitch thrown in a game such as instantaneous speed, movement, spin and location through the use of 3 cameras positioned throughout the ballpark.[6] The system was introduced in 2007, and provided limited data for its first season. It was installed in all 30 Major League Baseball (MLB) ballparks for the 2008 season, and the database is readily available to the public for download and analysis. The system has previously been used for quantitative and statistical studies,[6, 7] and utilized by broadcast networks to provide in-game statistical analysis and pitch speed reporting. This novel technology provides a unique opportunity to quantify the impact conditions associated with concussive impacts at the elite level in baseball.

The objective of this study was to determine the typical range of plate speeds for which umpires or catchers experience a foul ball to the mask causing concussive symptoms, as well as the locations on the mask at which these impacts occur. Better understanding of concussive impact conditions will give new insight and allow more in-depth laboratory experiments to investigate this mechanism of injury.

Methods

To create a database of relevant impact incidents, the mass media was searched for any reports of an occasion in which a catcher or umpire was struck in the mask by a foul ball during a MLB game throughout the 2008, 2009 and 2010 seasons. For each occasion, the necessary game details were recorded along with notation as to whether or not concussive symptoms were reported. Video copies of the games were obtained from the MLB (MLB.TV) and analyzed for close up footage and additional camera angles (if possible) of the impact incident.

The video footage was examined to determine the type of equipment worn by the catcher or umpire (2-piece traditional-style or 1-piece hockey-style) as well as the manufacturer and model

type if possible. Furthermore, the impact location on the mask was judged based on the numbering pattern described in Figure 1. This numbering pattern partitions the face into four anatomic regions: forehead, eyebrow, nose and chin. Each region is then further broken down into left, right and center areas, except for region number ten (chin area). By examining the locations of the impacts, conclusions could be drawn as to the general distribution of impacts over the mask of an umpire or catcher.
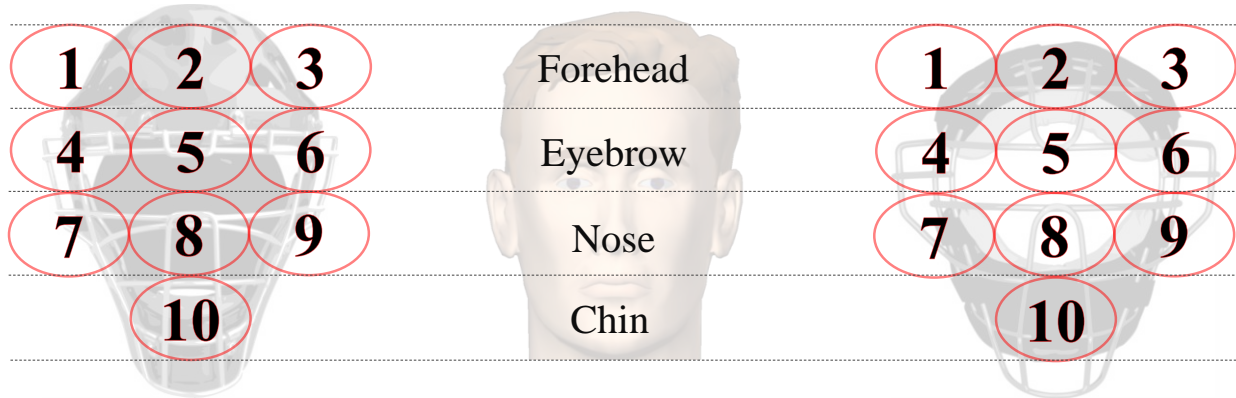
Figure 1: Numbering pattern for determining impact location. The left side graphic represents a 1-piece hockey-style mask, while the right side graphic represents a 2-piece traditional-style mask. The middle graphic shows the breakdown of the anatomical regions overlaid onto a human face.

The vertical spacing between regions correlates with specific areas of the mask construction that were easy to identify impacts with from the video. The forehead region is generally where the mask begins to contour along the top of the head. The eye region represents the top part of the vision gap in the cage just above the eyes at the eyebrow. The nose region represents the bottom part of the vision gap of cage directly in front of the nose. The chin region represents the bottom parts of the cage/mask structure in front of the chin padding.

To best approximate the impact speed of a baseball from a foul tip, the plate speed of the baseball was examined. Information describing the pitch characteristics for each impact was extracted from the Pitch F/X database. Game situation data (inning, batter, pitcher, count, catcher) from the video was accessed through Brooks[8] and used to determine the exact pitch thrown for each impact incident. Initial pitch release speed and plate speed were recorded.

The impact data were sorted to include only the impacts in which concussive symptoms were reported or a concussion was clinically diagnosed. The consequent data were tabulated and

analyzed to determine a range of plate speeds and impact locations for impacts known to cause concussive symptoms.

Results

Table 1 displays the impact characteristics collected for each of the 10 impacts that were reported on in the mass media for the 2008 – 2010 MLB seasons in which a concussion was clinically diagnosed or concussive symptoms reported. The traditional-style classification refers to the 2-piece system (faceguard and skull cap), while the hockey-style refers to the 1-piece system (helmet with faceguard rigidly attached).

Table 1: Impact information for all catcher and umpire impacts referenced in the mass media.

Impact Number	Position	Release Speed m/s (mph)	Plate Speed m/s (mph)	Impact Location	Equipment Style
1	Umpire	39.7 (88.8)	35.7 (79.9)	7	Traditional
2	Umpire	39.6 (88.6)	37.2 (83.2)	10	Traditional
3	Umpire	41.9 (93.7)	38.6 (86.4)	2	Hockey
4	Umpire	42.9 (96.0)	38.5 (86.1)	7	Traditional
5	Umpire	43.5 (97.2)	39.2 (87.6)	9	Traditional
6	Umpire	41.0 (91.8)	37.7 (84.4)	8	Traditional
7	Catcher	33.7 (75.4)	31.6 (70.6)	9	Traditional
8	Catcher	38.6 (86.3)	35.8 (80.1)	5	Traditional
9	Catcher	41.6 (93.1)	38.0 (85.1)	5	Traditional
10	Catcher	41.2 (92.2)	37.5 (83.9)	2	Hockey

Table 2 displays a statistical summary of the impact incident data. The average plate speed was 37 m/s (82.7 mph). The median impact speed was 37.6 m/s (84.2 mph). Figure 2 portrays the impact location distribution. Each impact is represented by a single baseball, and the numbering follows the pattern established previously in Figure 1. The longitudinal region experiencing the largest number of impacts was the center or mid-sagittal area (60%). The anatomic region with the most impacts was the nose area (50%).

Table 2: Statistical summary of baseball plate speeds for impacts that resulted in concussive symptoms or a diagnosed concussion. All speeds are given in m/s (mph).

Number of Impacts	Average	Maximum	Minimum	Median	Standard Deviation	Coefficient of Variation
10	37 (82.7)	39.2 (87.6)	31.6 (70.6)	37.6 (84.2)	2.21 (4.95)	5.88%
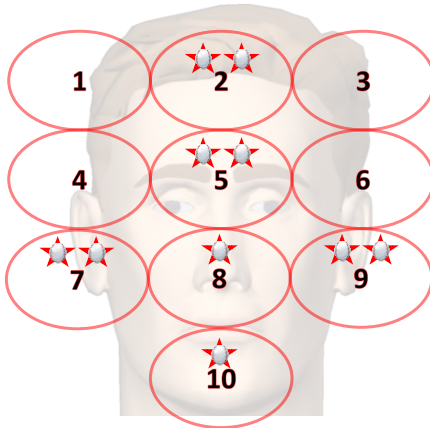
Figure 2: Location distributions for impact incidents.

Discussion

The distribution of plate speeds for concussive impacts was left skewed (Figure 3), indicating the median (84 mph) is a better approximation of the data. The relationship of a larger number of concussions with increased speed is intuitively expected, as baseballs traveling with more speed carry more energy, therefore increasing the risk of injury. As pitch speed generally increases with age and level of competition, this distribution may be unique to the MLB and caution should be taken when trying to translate trends to lesser levels.
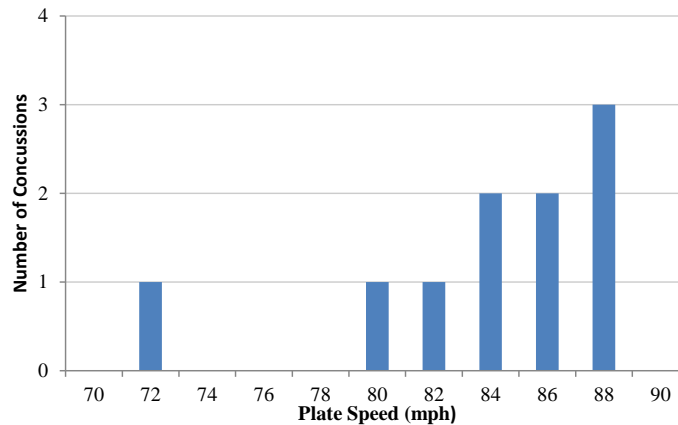
Figure 3: Histogram showing the distribution of the impact speeds for concussive impacts.

It is reasonable to extend the median value further into a range of values; therefore a target speed within 37.1 – 38.4 m/s (83 – 86 mph) should best approximate the real world data collected in this study. To our knowledge, no previous work has looked at catcher mask performance at this range of speeds. For helmet evaluation, Jones and Mohan (1984) examined higher speed tests between 40.2 and 44.7 m/s (90 and 100 mph) to determine how various types of helmets for many sports attenuate impacts[9]. Shain et al. (2010) looked at concussion injuries by evaluating catcher mask performance at slightly lower speeds between 26.8 m/s and 35.8 m/s (60 and 80 mph).[4] Other studies have performed baseball tests at similar speeds but only to evaluate the Hybrid III dummy in comparison to other dummy models.[3, 10]

The video results suggest that impacts occur in several locations; therefore they should be evaluated and compared with regards to severity. The most frequent impacts occurred vertically along the center of the face, and horizontally aligned with the nose. Figure 4 portrays generalized impact locations relative to anatomic landmarks of the face. This location distribution assumes symmetry across the mid-sagittal plane of the human head.
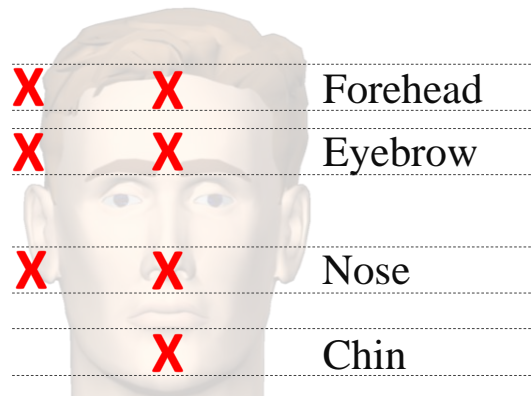
Figure 4: Suggested impact locations for further testing superimposed onto a human head.

The median plate speed of 84 mph is considerably higher than the NOCSAE standard test speed for faceguards of 70 mph. For locations, NOCSAE standard dictates that impacts with a baseball are to be done with the headform in a position perpendicular to the trajectory of the ball, a position 45° rotated from the mid-sagittal plane, and any position in which the center of the

impact occurs around the "no contact" or ocular area. In these positions, the baseball is to be aimed at the widest opening on the faceguard, at the material structure, and any point to exploit locations that may cause failure. Furthermore, softballs are aimed at two locations selected to exploit any area of possible failure. With these guidelines, it is possible to cover the same locations suggested by Figure 4. However, the NOCSAE standard does not require that the same locations be consistently tested from mask to mask, rather there is some freedom in selecting the exploited regions.

While this study provides a good approximation of the range of speeds and location for impacts to an umpire or catcher's mask, there are a few limitations that must be acknowledged. First, the sample size was limited because only impacts that were given attention in the mass media were investigated and analyzed, and the constraints of the Pitch F/X system restricted this data set to MLB impacts between the 2008 to 2010 seasons. To discover events that occurred that were not covered by the mass media, one would need to examine the video of every game played throughout a season. With 30 teams playing 162 games per season and approximately 135 pitches per game per team,[11] it may not be practical to obtain a full collection of impact incidents. However, this data would allow for a better understanding of the true impact location distribution, as well as the range of plate speeds of the pitched baseball.

The second main limitation of this study is that the plate speed of the baseball, not necessarily the impact speed, is reported. When a pitched baseball contacts a swung bat resulting in a foul ball, there is an associated speed loss and change of trajectory. This speed loss is not readily quantified and difficult to determine from the available video. Additionally, consistent side camera angles are necessary to determine the change in trajectory and the corresponding angle between the baseball and mask upon impact. The change in trajectory is further complicated by the pitched ball incoming at various angles (due to the pitcher residing on a raised mound as well as from movement due to different pitch types) and the batter swinging with different swing paths. Further work to understand the change in ball kinematics after partial impact with a bat could provide a more accurate indication of the speed in which the mask of a catcher or umpire is impacted, as well as the trajectory of the ball after contact with the bat. Regardless, the plate speed provides a better approximation than the release speed, as the speed at release dropped between 6 and 11% by the time the ball reached the plate.

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The results put forth in this study have applications towards mask design and evaluation. Shain et al. (2010) showed that catcher masks reduced head accelerations considerably, but it is still unknown what causes concussions from ball impacts to the head. Future work must be done to evaluate masks at the speeds and locations indicated. By knowing this information, engineering analyses can be used to evaluate and influence product design.[12,13] Testing can be done to determine a relative severity between the impact locations, in order to facilitate mask-to-mask evaluation at the most severe and possibly injurious locations. The results of such studies can be combined with the increasing knowledge of human tolerance to head impact to improve injury prevention.[14-16]

Conclusion

A series of 10 events in which a catcher or umpire experienced a foul ball to their mask were analyzed through both video and reported Pitch F/X data. These events comprised of occasions where concussive symptoms were reported or a concussion was clinically diagnosed. It was found that the impacts were well distributed across the face, and the median plate speed was approximately 37.6 m/s (84 mph). From these results further testing can be done to understand which locations on the mask cause the highest probability of concussion at speeds within the range of 37.1 − 38.4 m/s (83-86 mph). These data could provide insight into the etiology of concussion and can be used for experimental design to evaluate mechanisms of concussions in baseball.

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Chapter 3 - The Effect of Baseball Impact Location on Head Accelerations: Experiments with Catcher and Umpire Masks

Abstract

Ball impact to the mask from a foul tip is one way catchers and umpires experience concussions in baseball. These impacts have been shown to occur in a variety of locations at various speeds. In the first part of this study, seven locations on six different masks were tested at 60 mph to determine the most severe locations. Masks were fitted to a Hybrid III anthropomorphic test dummy head instrumented with a 3-2-2-2 accelerometer array and baseballs were projected with a modified pitching machine. Results showed no difference in response between locations in linear acceleration measurements. Impacts to the center-eyebrow and chin were higher in angular acceleration magnitude than other locations. Next, the center-eyebrow and chin locations were tested again on different masks of the same model at 84 mph. At this speed no difference was seen in linear or angular acceleration between locations. The results of this study provide data to support mask evaluation and comparison at the center-eyebrow and chin locations at a speed of 84 mph. These conditions simulate ball impacts that cause concussion in Major League Baseball. Performance comparisons can be used to improve mask design and safety for catchers and umpires at all levels.

Word Count: 199

Introduction

Head injuries in sports are occurring at an epidemic level (Greenwald et al., 2008). Some estimates suggest that upwards of 3.8 million sports related traumatic brain injuries (TBIs) occur every year (Langlois et al., 2006). Of these cases, 75% involve mild traumatic brain injury or MTBI, commonly referred to as concussion (Centers for Disease Control and Prevention (CDC), 2003). In the year 1995, head injuries in baseball comprised 18.5% of all competitive sports related head injury as reported by the National Electronic Injury Surveillance System (NEISS) (Thurman et al., 1998). Ball impact has been stated as the leading cause of head injury in baseball and softball (Heald & Pass, 1994). A foul ball deflection into the mask of a catcher or umpire is a specific scenario in which ball impact can cause head injury (Shain et al., 2010). Although the true exposure of catchers and umpires to these incidents is unknown, such impacts can cause loss of playing time and lead to early retirement.

The exact mechanism of concussion for ball impacts is unknown, and to our knowledge only one published study has investigated the occurrence of concussions from ball impacts to catcher masks. Shain et al. (2010) investigated the impact attenuation abilities of catcher masks on a Hybrid III dummy head (Shain, et al., 2010). They found that masks greatly reduced both linear and angular accelerations when compared to no mask at all. However, head accelerations with a mask were considerably less than acceleration levels typically associated with concussions. Furthermore, only four masks were tested with impacts targeted at the nose at speeds ranging from 60 – 80 mph.

Head acceleration is a commonly reported metric used as a predictor of head injury because it relates to the inertial response of the brain. Linear and angular accelerations of the head have been investigated in experimental studies on cadavers, animals and human volunteers to create injury metrics relating acceleration to injury potential. Both metrics are thought to produce different mechanisms of injury and therefore are typically investigated independent of each other (King et al., 2003; Unterharnscheidt, 1971). Linear acceleration injury criterion was developed off the Wayne State Tolerance Curve (Gurdijan et al., 1966). From this curve, injury metrics such as the Gadd Severity Index (SI), and Head Injury Criterion (HIC) have been derived (Gadd,

1966; Versace, 1971). However, these metrics are used to predict skull fracture, and thought to correlate with severe concussion.

Angular acceleration has been less studied, however scaled primate tolerance limits have been proposed by several researchers (Davidsson et al., 2009; Margulies & Thibault, 1992; Ommaya, 1985). More recent work has been geared towards relating head acceleration measured in football players to sports related concussion, and has given new insight to the head accelerations associated with injury (Duma & Rowson, 2011; Rowson et al., 2011; Rowson et al., 2009). Although the mechanism of injury for concussions from ball impacts is still unknown, suggested injury criterion and thresholds based on head accelerations have proved useful in quantifying the risk of injury. As such, linear and angular head acceleration are the metrics focused on in this study.

Previous work has investigated 10 concussive foul ball impacts experienced by catchers and umpires and found that typical ball speeds at the crossing of the plate ranged between 83 and 86 mph (Chapter 1). Furthermore, these impacts were approximated to seven different locations on the face. Comparison of the linear and angular accelerations of the head from impacts to these locations to injury thresholds in the literature will enable an understanding of the mechanism behind MTBI related injury. This knowledge can lead to further work in the evaluation of, and comparison between, the various mask types available for purchase by the consumer. Additionally, results from this study can be used in mask design and construction, providing specific regions of interest based on injury data. As such, the objective of this study was to evaluate the response of catcher and umpire masks to impacts at these different regions to discover the most severe impact location(s).

Methods

To simulate a baseball projecting towards the mask of a catcher or umpire after partial contact with the bat, baseballs were propelled by a baseball pitching machine. The pneumatic wheel, electric-motor driven machine (Jugs Sports, *Tualatin, OR*) was modified and anchored to the floor, reducing unnecessary vibration. The machine was capable of reaching a target velocity

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within +/- 3%, and having accuracy within a ¼ inch radius circle. These performance standards were set forth by the National Operating Committee on Standards for Athletic Equipment (NOCSAE) (NOCSAE, 2009b). Velocity was calculated over the final 3 to 4 inches before impact from high speed video shot at 3802 fps (Phantom V9, Vision Research *Wayne, NJ*), and accuracy was determined using carbon transfer paper to find the impact center as shown in Figure 5.
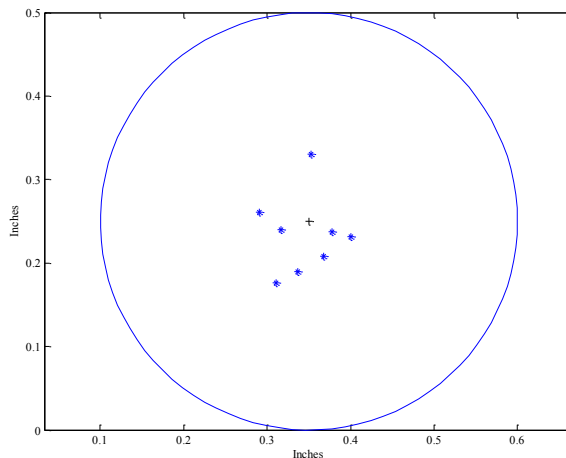
Figure 5: Sample accuracy chart of the pitching machine at 84 mph. The drawn circle represents the desired ¼ inch target on an arbitrary coordinate system (measured in inches), with the aim of the machine set to the center via laser. Impact locations were determined by finding the centroid of the digitized impact footprint and marked with an *.

While the NOCSAE standard for a projectile launching device makes no mention of ball spin, by its nature this machine causes the ball to be launched with some rotation. The speed on the machine is set by adjusting the rotation speed of the two wheels responsible for projecting the ball. As per the manufacturer's recommendation, the wheels were not set to the same level, to avoid a loss in accuracy from the knuckling of the ball as it skids between the two wheels. As such they recommended a difference of at least 10 to 15 units, to enable the baseball to roll along the wheels instead, allowing for a smoother delivery and increased accuracy. These settings resulted in spin rates approximately between 500 and 1000 RPMs at the speeds tested (60 & 84 mph). Referenced to the pitches seen in professional baseball, these values would fall below that of a typical curveball (1800 RPM) (Adair, 1990). Although it is expected that the ball would have an inherent rotation after contact with the bat, this value is not readily quantified and therefore remains unknown. The baseballs used were the official Major League Baseballs manufactured by Rawlings (*St. Louis, MO)* model RO-MLB, as specified by NOCSAE.

Mask performance was evaluated through the response of a surrogate headform. The head was attached to the neck and mounted to a 6.8 kg sliding base that was able to translate along the rails of a linear bearing table with low friction (Rowson, et al., 2011). To collect head kinematic data, a Hybrid III anthropomorphic test dummy (ATD) 50th percentile male head was instrumented with 9 accelerometers (Endevco 7264B-2000, *San Juan Capistrano, CA*) in a 3-2-2-2 configuration (Padgaonkar et al., 1975) so that linear and angular head acceleration could be calculated. Data acquisition was done using a TDAS Pro (DTS, *Seal Beach, CA*) system sampling at 20000 samples / second. The experimental setup is shown in Figure 6.
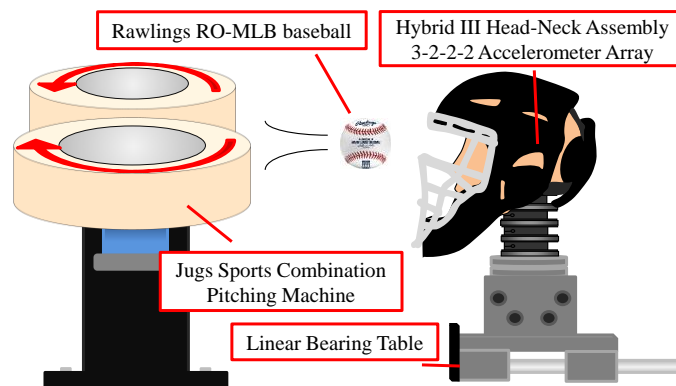
Figure 6: Pitching machine and the head and neck assembly on a linear bearing table.

A total of 7 locations were investigated (Figure 7) (Chapter 1). The impact points suggested fall in line with important landmarks on the mask construction, and are referenced to the tip of the nose of the Hybrid III. The forehead location is targeted 3 ⅝ inches above the tip of the nose; while the eyebrow location is targeted 2 ½ inches above the tip of the nose. The nose position is targeted at the tip of the nose. Finally, the chin position is targeted below the mouth, 2 ⅛ inches from the tip of the nose. The three targets on the lateral edge of the face are 3 inches away from their corresponding mid-sagittal locations. This separation distance represents the approximate diameter of a baseball so as to avoid impacting the same section of the mask structure twice.

3 in.

X X Forehead

X X Eyebrow

3 ⅝ in.

2 ½ in.

X X Nose

2 ⅛ in.

X Chin
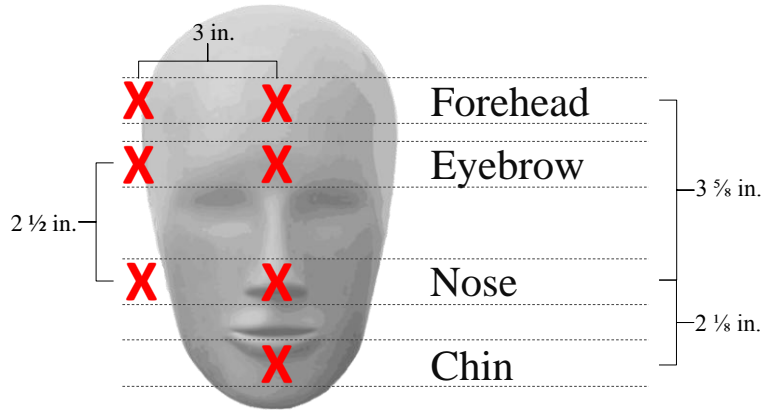
Figure 7: An illustration of the 7 test locations relative to the Hybrid III head.

The testing order went from the chin up to the forehead along the mid-sagittal plane of the face, and then from the forehead down to the nose on the lateral locations. The height of the linear bearing table was adjusted independently from the pitching machine by the use of blocks cut to the spacing distances, ensuring that each location was perpendicular to the flight of the ball. The horizontal spacing was set by sliding the bearing table laterally along guide slots.

The catcher and umpire masks tested were of the 1-piece "hockey-style" and 2-piece "traditional style". The 2-piece masks included the use of a skull cap (Rawlings – CCBCH) to replicate the conditions as a catcher would wear. It was found that while catchers use a skull cap for the mask, umpires do not normally wear one. Although umpire-specific masks were included in the evaluation, all masks were evaluated with a skull cap to maintain consistency. A total of 6 mask models (detailed in Table 3) were tested at 60 mph at each of the 7 locations. The speed of 60 mph was chosen to avoid severe damage or deformation to the masks throughout the tests. Because each mask was tested on 7 times, deformation or damage caused at one location could affect the response at other locations. NOCSAE certification tests the faceguards at 70 mph (NOCSAE, 2009a), therefore a lower value of 60 mph was chosen to minimize this confounding effect while still enabling a realistic response. After analysis of the performance of the masks at these 7 locations from 60 mph impacts, the eyebrow and chin targets along the mid-sagittal plane were selected for further testing at 84 mph on new masks of the same models listed in Table 3.

Table 3: Description of the masks used for testing. Sizing was chosen to best fit the Hybrid III head size.

Mask #	Make	Model	Type	Size
1	Rawlings	LWMX	Traditional-Style	Standard
2	Wilson	WTA3008	Traditional-Style	Standard
3	All-Star	FM25LMX	Traditional-Style	Standard
4	Easton	Stealth Speed	Hockey-Style	7 1/8 – 7 7/8
5	Wilson	WTA5520	Hockey-Style	7 – 7 5/8
6	All-Star	MVP2500	Hockey-Style	7 – 7 3/4

Mask sizing was determined by manufacturer recommendations for the circumference of the Hybrid III head (22.5 inches) and equivalent hat size (7 1/8 to 7 1/4). Positioning of the mask on the head also followed manufacturer instructions. The masks were fit to enable the line of sight of the dummy to "see" out the middle of the vision gap of the cage. This corresponded with aligning the top of the hockey mask forehead padding approximately ¼ inch above the top of the eye cavity (eyebrow). The chin padding was adjusted as per instructions to rest below the bottom lip. For the traditional-style masks, the skull cap was first fit snug to eyebrow line. The mask itself was placed over the skull cap into a realistic position enabling the dummy line of sight out of the middle of the vision gap. Snugness of fit was determined through a vigorous shake test. If the mask moved relative to the head throughout the shaking, straps were tightened until there was no relative movement. Once the desired fit was obtained for a mask, photographs were taken for comparison to ensure each repeated trial was positioned in the same way.

Data was processed according to SAE J211, and filtered using channel frequency class (CFC) 1000. The desired metrics calculated were peak resultant linear acceleration, peak resultant angular acceleration, severity index (SI) (Gadd, 1966), and Head Injury Criterion (HIC) (Versace, 1971) using custom written MATLAB programming. To compare the response at each location, peak linear and peak angular accelerations were tested by ANOVA, and a t-test performed.

Results

60 mph tests

The average baseball velocity was 60.6 mph with a standard deviation of 0.54, resulting in a coefficient of variation of 0.89%, indicating high repeatability.

The effect of position on peak linear acceleration was only found to be statistically significant for the eyebrow and chin locations compared to the lateral forehead. The highest linear acceleration of 28 g occurred at the lateral nose position, while the lowest value of 9 g occurred at the lateral forehead position. Figure 8 displays the results for linear acceleration and Table 4 details the statistical results. This lack of statistical significance was also found for the SI and HIC injury metric values (not shown). The SI values varied from $1 - 5$ while HIC values varied from $1 - 4$.
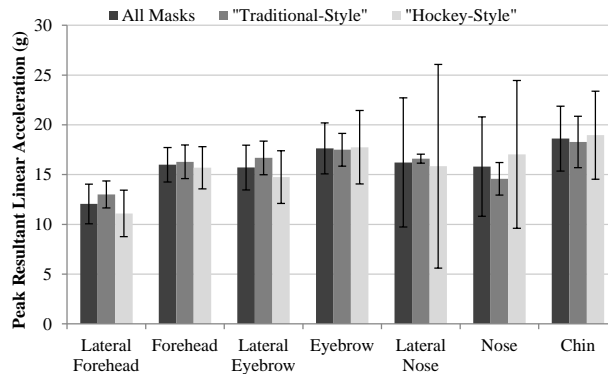
Figure 8: Average peak resultant linear acceleration at 60 mph impact speeds.

Table 4: P-value comparison between locations for linear acceleration (p < 0.05).

	Lateral Forehead	Forehead	Lateral Eyebrow	Eyebrow	Lateral Nose	Nose
Forehead	0.0755					
Lateral Eyebrow	0.0879	0.9396				
Eyebrow	**0.0165***	0.4966	0.4504			
Lateral Nose	0.0646	0.9396	0.8795	0.5453		
Nose	0.0755	1.0000	0.9396	0.4966	0.9396	
Chin	**0.0043***	0.2301	0.2029	0.5965	0.2599	0.2301

Impacts to the chin and eyebrow level induced the greatest angular accelerations of the head. The highest angular acceleration of 2591 rad/s^2 occurred at the eyebrow position, while the lowest of 769 rad/s^2 occurred at the lateral nose position. The chin and eyebrow locations were significantly greater than all locations except the lateral eyebrow. The lateral eyebrow location was statistically significant to all remaining locations except the nose (Table 5). Figure 9 presents the results of the rotational kinematics for 60 mph impacts.
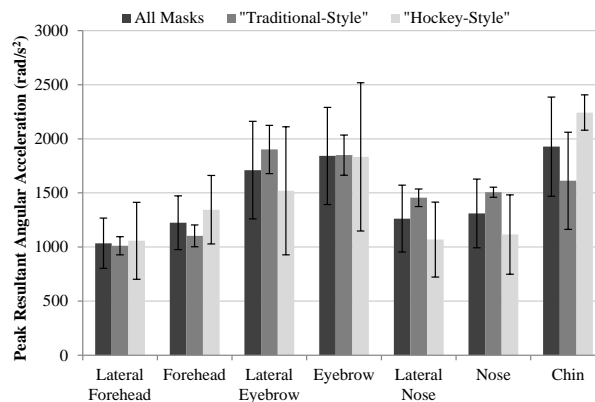
Figure 9: Average peak resultant angular acceleration at 60 mph impact speeds.

Table 5: P-value comparison between locations for angular acceleration.

	Lateral Forehead	Forehead	Lateral Eyebrow	Eyebrow	Lateral Nose	Nose
Forehead	0.3737					
Lateral Eyebrow	**0.0028***	**0.0265***				
Eyebrow	**0.0050***	**0.0058***	0.5371			
Lateral Nose	0.2864	0.8569	**0.0399***	**0.0092***		
Nose	0.1973	0.6820	0.0653	**0.0162***	0.8182	
Chin	**0.0020***	**0.0020***	0.3088	0.6848	**0.0032***	**0.0059***

84 mph tests

The average velocity was 83.9 mph +/- 0.59, resulting in a coefficient of variation of 0.70%, indicating high repeatability.

From the 60 mph test, the eyebrow and chin locations were chosen for further testing at 84 mph. Comparisons between locations were not statistically significant. The greatest linear acceleration was 42 g at the chin position, and the smallest was 26 g at the eyebrow position. The greatest angular acceleration was 5266 rad/s^2 at the chin position, and the smallest was 1974 rad/s^2 at the eyebrow position. Figure 10 and Figure 11 display the results of the 84 mph impacts graphically. Table 6 tabulates the range of values for peak linear and peak angular acceleration.
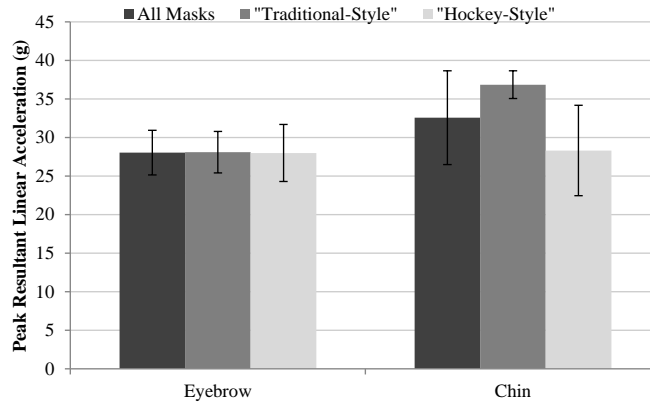
Figure 10: Average peak resultant linear acceleration at 84 mph impact speeds.
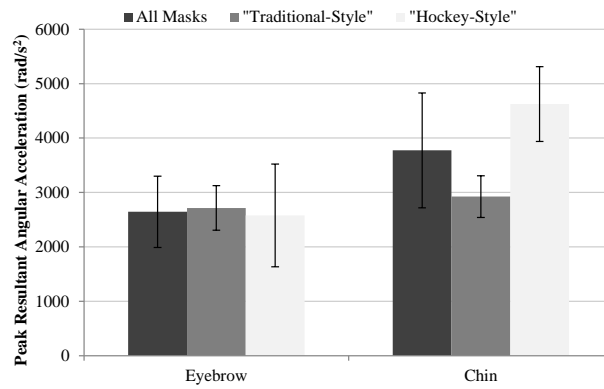
Figure 11: Average peak resultant angular acceleration at 84 mph impact speeds.

Table 6: Summary of average peak linear and angular accelerations.

Impact Speed (mph)	Number of Locations	Average Peak Linear Acceleration (g)	Average Peak Angular Acceleration (rad/s^2)
60	7	16 +/- 3.9	1473 +/- 468
84	2	30 +/- 5.1	3210 +/- 1024

Discussion

In an evaluation of 7 different impact locations, impacts to the eyebrow and chin along the mid-sagittal plane of the head resulted in the greatest angular accelerations for impacts at 60 mph. Linear acceleration, SI and HIC were not statistically significant among the locations. Furthermore, no significant difference in response was exhibited between the eyebrow and chin locations at 84 mph.

The difference in angular acceleration among locations is described by the positioning of the impact locations and the rotation point of the head (occipital condyle pin). Figure 12 illustrates the relative positioning between impact locations. The increased angular acceleration from impacts to the chin and eyebrow relative to those at the nose is expected, because of the distance between these locations and the point of rotation. Whereas the eyebrow and chin are vertically above/below the occipital condyle pin, the nose roughly falls in line with it.
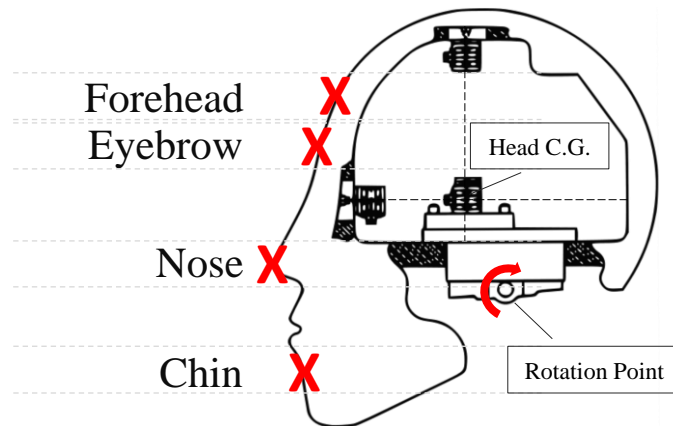
Figure 12: Schematic of the Hybrid III dummy head and occipital condyle pin point in relation to the impact locations.

It also can be expected that the forehead locations would induce high angular accelerations as well, given their distance from the point of rotation. However, high angular accelerations were not seen. At the forehead targets, the shape of the masks begins to contour posteriorly, providing for a surface which the projected baseball can deflect off. As such, less energy is transferred into the head, therefore resulting in smaller angular acceleration values.

Peak linear resultant acceleration ranged from 26 to 42 g for the 84 mph tests. The linear acceleration values fall well below accepted injury thresholds found in the literature. In the reconstruction of 25 concussive impacts from National Football League (NFL) game video, Pellman (2003) found that average linear acceleration for concussive impacts was 98 +/- 28 g, with none falling below 48 g (Pellman et al., 2003). Logistic regressions performed on this data predicted a 50% risk of concussion at 79-82 g. (King, et al., 2003; Zhang et al., 2004). Using the parameters provided by Pellman our greatest value of 42 g resulted in a 9% risk. However, as this data set has been judged to be overly conservative because of its bias towards concussive impacts; less conservative estimates suggest a 10% risk of concussion at 165 g (Funk et al., 2007). Although various thresholds have been presented, the linear accelerations seen in these impacts do not reach near the published thresholds, suggesting the influence of linear acceleration on the concussions experienced by catchers and umpires is relatively small.

Peak angular acceleration ranged from 1974 to 5266 rad/s^2 for the 84 mph tests. Although on the lower side of published values, angular acceleration is in a better agreement with thresholds provided in the literature than linear acceleration. The Pellman reconstruction average angular acceleration was 6432 +/- 1813 rad/s^2, however values of 2615, 3476, 4727 and 4381 rad/s^2 were reported for concussive causing impacts (Pellman, et al., 2003). The range for these four impacts agrees with the range of angular accelerations seen in this study. Logistic regression on this data suggested a 50% risk of concussion at 5757 – 5900 rad/s^2 with a 25% risk of concussion at 4384 to 4600 rad/s^2, (Zhang, et al., 2004) (King, et al., 2003). Using the Pellman parameters on the values obtained in this study (1974 – 5266 rad/s^2), risk was calculated to fall between 4 and 45%. It is likely however, that this curve overestimates risk (Funk, et al., 2007).

The duration of these impacts was approximately 4 – 5ms. This falls closer in range with that seen for head impacts to vehicle structures in crashes (< 6 ms) than with those for which the angular acceleration criterion was developed (average 14 – 15 ms) (Pellman, et al., 2003). As such, caution may be needed when evaluating head accelerations by injury criterion determined from football impacts.

There are several important limitations of this study that need to be highlighted for consideration. Although the 60 mph speed was expected to be low enough to avoid damage to masks, some deformation was seen, particularly on the hockey style helmets. Such damage was slight and localized to the impact area and not visually found to extend to other impact areas. Even so, it is possible that the response of the masks could have been altered due to existing damage from previous trials. However, the spacing of the impacts was large enough to where it is believed this influence was small and did not affect the performance of the masks. At 84 mph, more substantial localized deformation to the masks was seen. Because the impact locations were located on two separate sections of the mask, the deformation at one site was not expected to influence the response at the other.

When evaluating the response of the Hybrid III at the chin location, questions of biofidelity must be addressed. The obvious difference between the Hybrid III and a human chin is the lack of an articulated tempromandibular joint (TMJ). It is expected that the ability of the jaw to translate when subjected to an impact at the chin could alter the response of the head. This is an inherent limitation of all testing with ATDs, and is not specific to this test setting. However, industry accepted NOCSAE standards suggest the use of a linear bearing table and a headform without an articulated TMJ to certify masks for sale and use in all levels of baseball. Although NOCSAE does not use the Hybrid III, its use in this study was necessary to determine angular kinematics through a 3-2-2-2 accelerometer array. Furthermore, the Hybrid III has commonly been used to assess the potential for injury in sports (Pellman, et al., 2003; Rowson et al., 2008; Shain, et al., 2010).

The sample size of the masks is also a potential limitation to this study. The six masks chosen were all common models offered for sale by popular manufacturers. Furthermore, both traditional and hockey style masks were represented within the sample pool. As such, it is believed that the sample size accurately portrays the wide variety of masks available to consumers.

Through these series of tests, the chin and center-eyebrow locations resulted in the greatest head accelerations. Engineering analyses can be used to evaluate and influence product design with

this information (Rowson et al., 2010; Rowson, et al., 2008). Future work can provide insight as to how mask design and construction can be tailored to performance under these conditions, to increase the protection and safety provided to the wearer. When pooled with advances in understandings of human tolerance to head impacts, such work can help to improve injury prevention (Cormier et al., 2011; Greenwald, et al., 2008).

Conclusions

To determine impact locations that elicited the most severe response in a Hybrid III head when impacted by a baseball, catcher and umpire masks were tested at 7 different locations. Testing at 60 mph and 84 mph provided data to support that impacts to the center-eyebrow and chin locations were the most severe. Peak linear accelerations of 26 to 42 g at 84 mph were found to be lower than suggested injury thresholds. Peak rotational accelerations of 1974 to 5266 rad/s^2 were found to be comparable with the lower limits of human tolerance for concussion with respect to rotational kinematics, and coincided with values for impacts of known concussions. Knowledge of these impact locations can be used to test and evaluate masks to facilitate a comparison between manufacturers, mask types and styles.

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Chapter 4 - A Biomechanical Analysis of Baseball Catcher and Umpire Masks

Abstract

Although at lower rates than contact sports, concussions occur in baseball. One mode of this injury is through an impact by a fouled ball directly to the mask of a catcher or umpire. Various types and styles of masks are available from manufacturers that are designed to protect a catcher or umpire in this type of situation. In this study, 26 masks were impacted with baseballs at 84 mph to the center eyebrow and chin regions of the face. Head kinematics were measured with a Hybrid III head and neck mounted on a linear bearing table. Results indicated that peak resultant linear acceleration were below tolerance thresholds. However, peak angular accelerations were on the lower limit of thresholds, and well within ranges of values known to cause concussions. When evaluating the difference between mask types, no significance was seen between 1-piece hockey-style masks and 2-piece traditional-style masks. Titanium masks were shown to have significantly higher linear accelerations than the same masks with a steel cage. The outcome of this study has applications to mask design, construction and certification. By evaluating masks under the same conditions in which injury is experienced, a true comparison of mask performance can be made.

Introduction

It has been suggested that approximately 1.6 to 3.8 million sports-related Traumatic Brain Injuries (TBIs) occur across the United States every year (Langlois et al., 2006). The Center for Disease Control and Prevention reports that as many as 75% of TBIs per year fall into the classification of Mild Traumatic Brain Injuries (MTBIs) (Centers for Disease Control and Prevention (CDC), 2003). MTBI garners the most frequent attention in contact sports such as football, hockey or boxing, but also can occur in non-contact sports such as baseball. The National Electronic Injury Surveillance System (NEISS) estimates that head injuries in baseball comprised 18.5% of all sports related head injuries in 1995 (Thurman et al., 1998). The long-term impact of these concussions can vary from a short loss of playing time, to retirement from the game. Furthermore, a history of concussive injuries may increase a person's risk of additional concussions when compared to those with no prior concussions (Collins, 2002).

One mechanism for head injury in baseball is through impact of a ball to the head of a catcher or umpire from a fouled ball. Although few studies have focused specifically on the occurrence of head injury in baseball at the Major League level (MLB) (McFarland, 1998), work has been published on injury data for collegiate teams. Dick et al. (2007) showed that injuries to catchers account for approximately 7.5% of all injuries in baseball, and 9.3% come from batted balls (Dick et al., 2007).

Currently, little information is available in the peer-reviewed literature as to the performance of catcher and umpire masks in baseball. Shain et al. (2010) have provided the most relevant work so far, detailing the ability of catcher's masks to attenuate head accelerations (Shain et al., 2010). In previous chapters, work has been done to understand the locations and speeds that correlate with the conditions experienced by catchers and umpires in concussive impacts during games.
The National Operating Committee on Standards for Athletic Equipment (NOCSAE) establishes standards of performance for catcher masks, but does not evaluate or compare them directly. For catcher (and umpire) masks, the evaluation standard is comprised of drop tower tests for the helmet and projectile impacts for the faceguard. The drop tower tests occur at various orientations from a 36 in drop height onto a specified anvil. To pass the tests the Severity Index (SI) (derived from linear acceleration) must not measure above 1200. Projectile tests for the

helmet also occur at various locations at a speed of 55 mph with a softball and 60 mph with a baseball, and the same SI threshold is used. Faceguards are evaluated with impacts from both baseballs and softballs at 70 mph. Locations are determined by landmarks on the faceguard structure. Pass/fail criterion are based on the interaction of the faceguard with areas of the face designated as limited or no contact, and not related to any acceleration metrics (NOCSAE, 2009a).

Various types and styles of masks are available on the market, all with advertised advantages. Masks fall into two categories: 2-piece traditional-style and 1-piece hockey-style. Traditional-style masks are worn in two pieces, with the mask pulling over a backwards helmet or skull cap on the top of the head. Hockey-style masks are a single unit similar to the type of masks worn by hockey goalies, in which the mask and protective helmet are attached. On top of these two styles, caging material is also variable. Most masks are constructed of steel alloy, while some masks will utilize titanium. These titanium masks are considerably higher in price and advertised to provide the same strength and durability at a reduced weight. Athletes interested in the use of a helmet have little information at hand to assist in selecting the correct helmet for their protective needs.

Evaluation of currently available masks and a presentation of the performance results will help aid athletes and coaches in mask selection and purchase. Furthermore, classification of systematic differences between types and styles can lead to improved design and construction by all manufacturers. The objectives of this study were to evaluate a large sample of masks under the conditions that may be most likely associated with injury (chapter 2), to provide a relative comparison between manufacturer models, and to uncover any systematic trends across mask types and styles.

Methods

Baseballs (RO-MLB, Rawlings, *St. Louis, MO*) were propelled towards masks at a target speed of 84 mph using a pneumatic-wheel, electric-motor driven combination baseball/softball pitching machine (Jugs Sports, *Tualatin, OR*). The machine was able to hit a targeted velocity within 3%,

and maintain an accuracy within a ¼" radius circle as per NOCSAE standard requirements (NOCSAE, 2009b). Velocity measurements were taken over the final 4 inches before impact from high speed video shot at 3802 fps (Phantom V9 Vision Research, *Wayne, NJ*). Although some spin is inherent to this type of machine, the spin rates of 500 – 1000 RPM were not expected to have any influence on response.

Masks were fitted to a 50th percentile male Hybrid III anthropomorphic test dummy (ATD) head to quantify performance. The dummy head and neck were mounted to a 6.8 kg sliding base capable of translation with low friction along the rails of a linear bearing table (Rowson et al., 2011). Cut to length blocks were available to adjust the height of the linear bearing table to maintain an orthogonal relationship between the incoming baseball and the coronal plane of the head. The head was equipped with a 3-2-2-2 accelerometer array (Endevco 7264B-2000, *San Juan Capistrano, CA*) to measure linear acceleration and enable angular acceleration calculation (Padgaonkar et al., 1975). Data acquisition was done using TDAS Pro (DTS *Seal Beach, CA*) sampling at 20000 samples/sec. Post-test processing was performed in MATLAB (Mathworks *Natick, MA*). Data were filtered at channel frequency class (CFC) 1000, as specified in SAE J211.

Table 7 lists the masks that were evaluated in this study. Mask sizing was selected based on the circumference of the Hybrid III head (22.5 inches). Fitting was done according to manufacturer specifications. To ensure a consistent fit across brands, various landmarks on the Hybrid III face were used. The traditional-style masks were fit with a skull cap (Wilson WTA3121, 7 – 7 5/8, mass 315 +/- 5g) pressed down snug to the head, with the front of the cap aligned with a line drawn along the top of the eye cavity (eyebrow). Four skull caps were paired so that repeated trials with the same mask used the same skull cap. Hockey-style masks were fit with the forehead padding approximately ¼ inch above this eyebrow line. Once in position, the chin pad was adjusted to line up just below the lips. Throughout the fitting, masks were centered on the face and aligned in a realistic position to allow the Hybrid III to "see" out of the middle of the vision gap in the cage. Photographs were taken of the fit of each mask to ensure consistency within each set of two masks.

Table 7: A description of all the masks used in this study (listed alphabetically). All masses are listed in grams with a tolerance of +/- 5 grams. In some instances, only one mask of a model was tested.

Manufacturer	Model	Style	Cage	Mass 1	Mass 2	Size
All-Star	FM25LMX	Traditional	Steel	590	-	Standard
All-Star	FM25LUC	Traditional	Steel	580	585	Standard
All-Star	FM25TI	Traditional	Titanium	530	535	Standard
All-Star	MVP 2500	Hockey	Steel	1280	-	7 – 7 3/4
All-Star	MVP 4000 SL	Hockey	Steel	1270	1280	7 – 7 3/4
All-Star	MVP 4000 TI	Hockey	Titanium	1195	1175	7 – 7 3/4
Diamond	DFM-iX3 Ump	Traditional	Steel	465	465	Standard
Diamond	Edge iX3	Hockey	Steel	985	985	7 1/8 – 7 1/2
Diamond	Edge Ump	Hockey	Steel	1110	1110	7 1/8 – 7 1/2
Diamond	Pro Ultra-Light	Traditional	Steel	455	460	Standard
Easton	Stealth Speed	Hockey	Steel	1305	-	7 1/8 – 7 7/8
Louisville Slugger	TPXCH	Hockey	Steel	1130	1135	7 1/8 - 8
Louisville Slugger	TPXFM	Traditional	Steel	800	820	Standard
Mizuno	Classic G2	Traditional	Steel	810	790	Standard
Mizuno	Samurai G3	Hockey	Steel	1100	1125	7 – 7 5/8
Nike	Pro Gold	Hockey	Steel	1125	1135	6 3/8 – 7 3/8
Rawlings	HLCH1 Coolflo	Hockey	Steel	1020	1025	7 1/8 – 7 3/4
Rawlings	LWMX	Traditional	Steel	790	-	Standard
Under Armour	Pro Catcher's Face Mask	Traditional	Steel	565	575	Standard
Under Armour	Pro Catcher's Helmet	Hockey	Steel	1215	1215	7 – 7 3/4
Wilson	A 3077 MLB	Traditional	Steel	725	710	Standard
Wilson	A 3077 Titanium MLB	Traditional	Titanium	615	620	Standard
Wilson	Dyna-Lite A 3008	Traditional	Steel	710	-	Standard
Wilson	Shock FX 2.0 A 5520	Hockey	Steel	1205	1205	7 – 7 5/8
Wilson	Shock FX 2.0 A 5520	Hockey	Steel	1195	-	7 – 7 5/8
Wilson	Titanium Shock FX A 5590	Hockey	Titanium	1085	1085	7 – 7 5/8

Of the 26 masks, 20 mask models had two separate masks tested, while 6 models only had one trial. Two locations shown to produce the greatest accelerations were tested on each mask: the center-eyebrow location (2.5 inches above the tip of the nose), and the chin location (2.125 inches below the tip of the nose). Both locations fell along the mid-sagittal plane of the Hybrid III head. Testing speed was targeted at 84 mph, with a tolerance of +/- 3%. Although no interaction between locations was expected, the test was order varied for each repeated trial to minimize potential sequential effects from deformation at one location affecting the response at the other.

Statistical analysis was used to determine significance between the desired groupings for comparison. To evaluate traditional-style and hockey-style masks, a t-test was performed ($p < 0.05$). To evaluate the steel and titanium masks, a matched pair t-test was performed ($p < 0.05$). The matched pair was used because the mask padding structure was the same between the two

cage styles. Therefore the steel and titanium data were paired, and could be analyzed as such. The individual masks were compared with ANOVA, and then significance was determined with a Fisher's least significant difference (LSD) ($p < 0.05$).

Results

Table 8 details the average kinematic ranges for the various groupings of masks. Figure 13 shows the results of the masks by linear and angular accelerations. Each average value is calculated from all tests performed on that mask type. A small subset of masks was only evaluated with one impact at each location. The raw linear and angular acceleration values are provided in Table 9. Figure 14 shows the comparison between average grouping values. In comparing titanium and steel masks $p = 0.037$, indicating statistically higher linear accelerations for the titanium masks when compared to their paired steel counterparts. No statistical significance was seen for any other groups in angular or linear accelerations.

In additional to linear and angular kinematics, common injury metrics such as Gadd Severity Index (SI) and Head Injury Criterion (HIC) were calculated (Gadd, 1966; Versace, 1971). For all masks, SI values ranged from $3 - 26$ and HIC 15 values ranged from $2 - 21$. The average duration of the impacts was $4 - 5$ ms. Angular velocity values ranged between 7 and 18 rad/sec. The average ball velocity of 85.2 +/- 0.82 mph was within the desired target of $83 - 86$ mph. Furthermore, the range of speeds fell within the specified +/- 3% of the average ($83.6 - 87.0$ mph). The low coefficient of variation (0.97%) indicates very good repeatability between trials.

Table 8: Summary of peak linear and peak angular acceleration ranges.

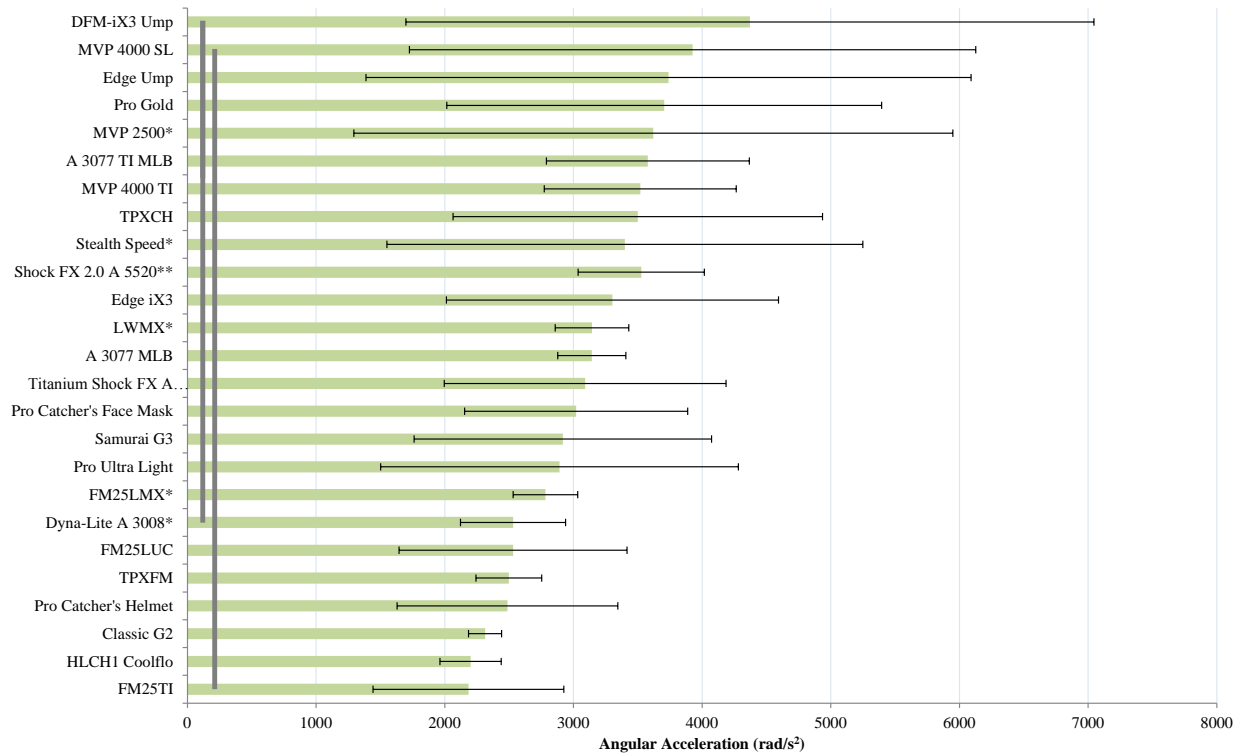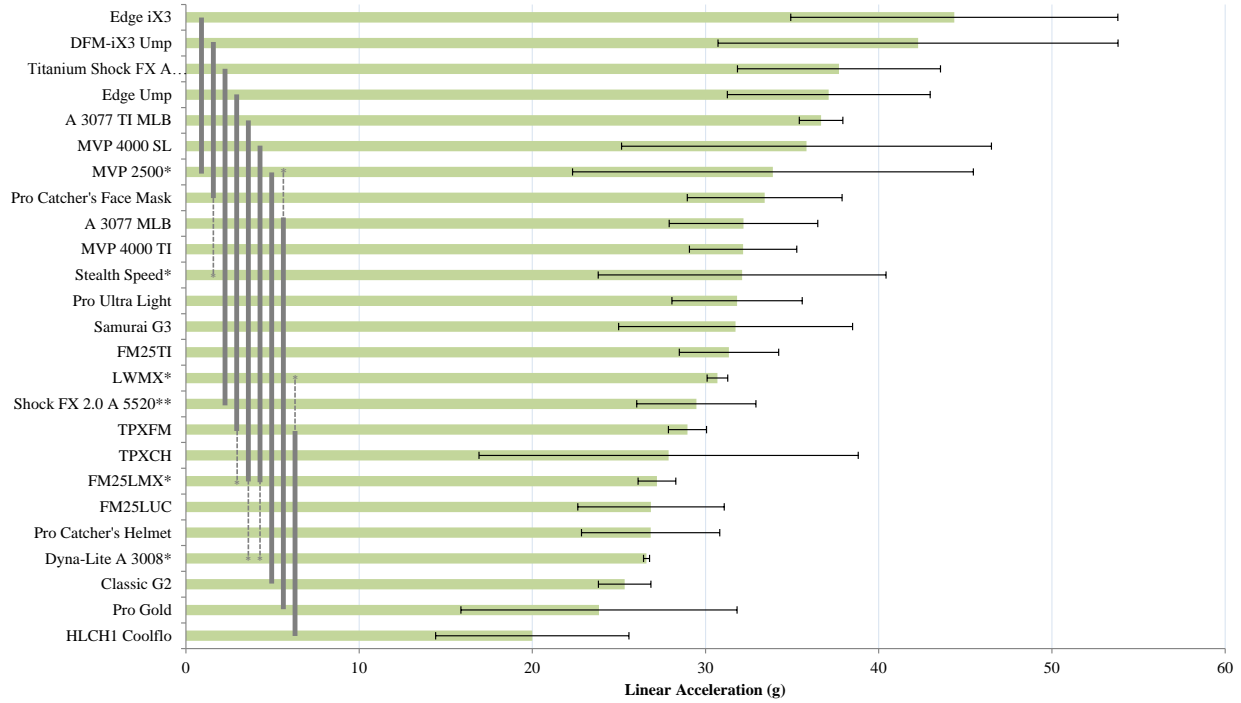|	Peak Linear Range (g)	Peak Angular Range (rad/s^2)
All Masks	32 (12 – 53)	3103 (1394 – 7237)
Hockey	32 (12 – 53)	3231 (1482 – 7237)
Traditional	32 (23 – 53)	2984 (1394 – 6997)
Steel	31 (23 – 51)	3187 (1583 – 7058)
Titanium	34 (29 – 44)	3093 (1394 – 4481)

Figure 13: Mask rankings by average peak linear and peak angular accelerations. Any masks connected by vertical bars were not significantly different. The dashed vertical lines connect * to their vertical bars. In the model names, * indicates only one mask used to calculate the average, ** indicates three masks.

Table 9: Shows the raw linear (g) and angular (rad/s^2) accelerations grouped by mask and location.

| Model | Center-Eyebrow | | | | Chin | | | |
| | Mask 1 | | Mask 2 | | Mask 1 | | Mask 2 | |
	Linear	Angular	Linear	Angular	Linear	Angular	Linear	Angular
FM25LMX	26	2960	-	-	28	2605	-	-
FM25LUC	23	1969	23	1583	30	3333	31	3237
FM25TI	30	1394	32	1792	29	2479	35	3070
MVP 2500	26	1974	-	-	42	5266	-	-
MVP 4000 SL	30	2274	26	2518	51	7058	36	3851
MVP 4000 TI	31	2918	30	2941	37	4481	30	3736
DFM-iX3 Ump	34	2204	31	1933	51	6351	53	6997
Edge iX3	31	1644	44	2973	48	4014	53	4579
Edge Ump	34	2267	34	2466	46	7237	35	2983
Pro Ultra-Light	30	1446	28	1996	36	4346	34	3777
Stealth Speed	26	2092	-	-	38	4707	-	-
TPXCH	18	2093	19	2430	37	4708	38	4767
TPXFM	28	2632	28	2201	29	2775	30	2384
Classic G2	24	2383	24	2453	27	2172	26	2244
Samurai G3	24	1703	29	2323	36	3297	39	4346
Pro Gold	12	1482	29	4048	30	5577	25	3714
HLCH1 Coolflo	24	1863	26	2424	16	2244	15	2269
LWMX	31	2942	-	-	30	3347	-	-
Pro Catcher's Face Mask	33	2496	30	2698	40	4315	31	2572
Pro Catcher's Helmet	25	2179	28	2205	23	1819	32	3747
A 3077 MLB	35	2917	36	2928	28	3451	29	3276
A 3077 Titanium MLB	35	2818	37	2979	38	4193	37	4323
Dyna-Lite A 3008	26	2242	-	-	27	2819	-	-
Shock FX 2.0 A 5520	28	2546	32	3551	25	2958	32	3549
Shock FX 2.0 A 5520	32	3665	-	-	31	3899	-	-
Titanium Shock FX A 5590	33	2203	33	2108	44	3814	41	4239
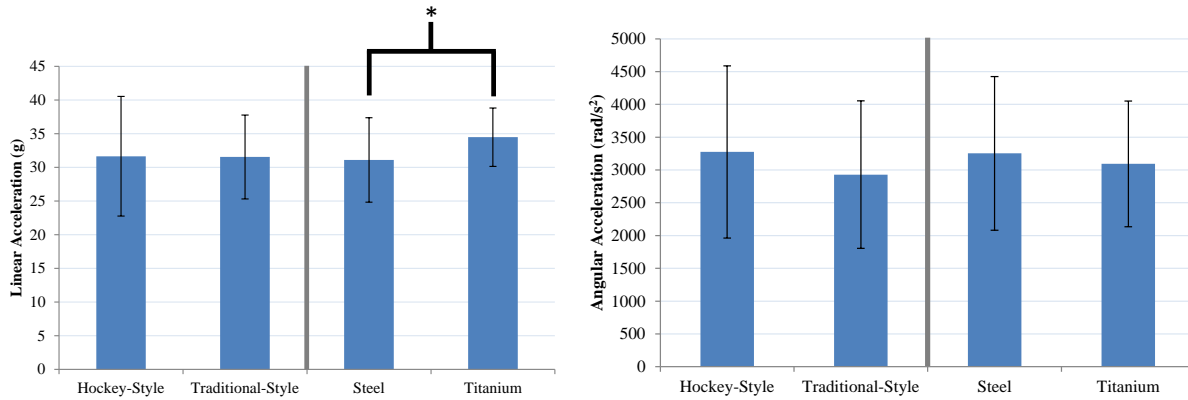
Figure 14: Group comparisons between linear and angular accelerations of hockey-style and traditional-style masks, and steel and titanium masks.

Figure 15 shows a comparison between the chin and center eyebrow locations for all impacts. In comparing the peak linear acceleration, p = 0.0005, indicating strong statistical significance. In comparing the peak angular acceleration, p < 0.0001, again indicating strong statistical significance.
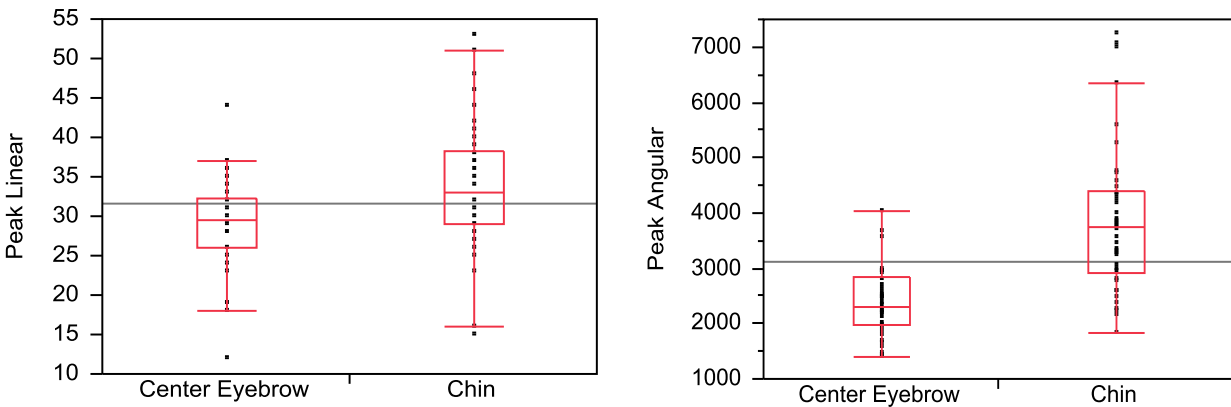

Figure 15: Performance comparison between the chin and center eyebrow locations for linear and angular acceleration for all impacts. Both comparisons were statistically significant.

Discussion

Peak resultant linear and peak resultant angular accelerations were the only metrics analyzed in this study. Although SI and HIC values were reported, they were omitted from analysis because of their relatively low values compared to previously reported injury thresholds (Pellman et al., 2003). Across all 26 masks tested, there was no significant difference between hockey-style and traditional-style masks in terms of resultant peak linear or angular acceleration when subjected to an 85 mph frontal impact with a baseball. Any differences are more dependent on manufacturer

model than they are on mask style. However, it is important to consider that impacts do not always occur in the coronal plane of the face. Catchers and umpires can also be hit in various other parts of the head from broken bats, bat backswings, and errant foul balls. Additionally, catchers experience collisions with base runners attempting to cross home plate. The frontal impact is only one facet of protection that needs to be considered when purchasing a mask.

Despite advertisements claiming that titanium masks are stronger and provide better protection, little evidence was discovered in this study to support such claims. In fact linear acceleration was deemed to be significantly higher in titanium cages. One explanation for the increased linear acceleration response may be related to the use of a stronger material for the cage construction. With the titanium deforming less on impact compared to the steel, more energy is transferred to the head than into the bending of the material, therefore increasing the head acceleration. Additionally, with titanium masks being 50 – 120 grams lighter than their steel counterparts, the reduced mass could also contribute to the increased response. Although the benefits of a lighter mask include less head supported mass and the ability to remove the mask quicker, one must look at the true benefits the reduction of 50 – 120 grams provide compared to their possible drawbacks. Furthermore, titanium masks are considerably more expensive than steel masks, and therefore may not be economical for all.

It is important to keep in mind the magnitude of the responses when looking at the comparisons between mask models, styles and types. Average linear acceleration values were still well below suggested injury thresholds for risk of MTBI (King et al., 2003; Pellman, et al., 2003). Additionally, angular acceleration values were on the low end of injury thresholds published in the literature (King, et al., 2003; Pellman, et al., 2003). However, peak values of 53 g and 7237 rad/s^2 fall within range of reported concussions values. Guskiwicz et al. (2007) recorded concussions in football players at linear accelerations as low as 60 g, and various concussions at angular accelerations between 1000 and 9000 rad/s^2 (Guskiewicz et al., 2007). Broglio et al. (2010) recorded concussions at angular accelerations between 5500 and 9500 rad/s^2 (Broglio et al., 2010). Furthermore, although various levels of statistical significance were seen between masks, the nominal values indicate different responses. With a 50% increase over the range of

linear accelerations, and substantial differences in angular acceleration values, mask design can play a role in attenuating impact energy.

There are several limitations to acknowledge when considering the results of this study. The Hybrid III is known to have limited biofidelity; for example the rigid jaw fixture of the Hybrid III does not portray the hinged tempromandibular joint seen in humans. This mobility in the jaw could influence the response of the mask to impact. However, it is the gold standard for automotive testing and frequently utilized in applications outside the automotive industry, such as sports injury research (Pellman, et al., 2003; Rowson et al., 2008). Due to the neck response and ability to measure rotational kinematics it is the best available device for use today.

Dummy response at the chin was consistently higher in terms of linear and angular response when compared to the center-eyebrow. Although it can be contributed in part to the lack of biofidelity in the jaw, the fit of the mask also affected the chin response. Although the sizing of masks was selected correctly based on the head circumference, there was a noticeable amount of correction necessary to the chin pad of hockey-style masks to enable the padding to line up on the face correctly. The length of the Hybrid III face relative to its head circumference may not be the same proportion to which manufacturers design their masks. As such, responses at the chin may not truly represent the responses seen in humans. However, this was consistent across all masks, and the overall ratings were comprised of an average between the response at the center-eyebrow and chin.

One facet of MTBI applicable to impact scenarios that is not well understood is the effect of cumulative impacts (Guskiewicz et al., 2003). Furthermore, the exposure of catchers and umpires to these impacts during a typical season has not been studied. Future work could explore the frequency of impacts to the mask from a fouled ball to understand how often catchers and umpires experience impacts of these magnitudes. This information could be used to understand how cumulative impacts at low magnitudes may affect the incidence of concussion. Additionally, in situ devices such as the Head Impact Telemetry System (HITS) (Simbex *Lebanon, NH*) could be adapted to the masks worn by catchers and umpires to gain a further understanding of the kinematic response of the head (Duma & Rowson, 2009, 2011; Rowson et al., 2009). With this

further understanding of injury thresholds and exposure, an analytical evaluation of these results could lead to improved product design and performance (Rowson et al., 2010; Rowson, et al., 2008).

Conclusions

The data put forth in this study will contribute substantially towards the design, construction and sale of catcher and umpire masks to the general public. Results have shown that there is no significant difference between the performance of traditional-style and hockey-style masks when subjected to frontal impacts at 85 mph. Additionally titanium masks, while lighter than their steel counterparts, may actually increase the response when subjected to the same impact severity. However, care must be taken when interpreting these results as one to one comparisons between mask models. Although mask response fell below injury thresholds, values approached those associated with the lowest recorded concussions. As such, the differences between masks may not correlate to considerable reductions in risk of MTBI for catchers and umpires. Nevertheless, these data are a first attempt at comparing the equipment currently available on the market, and may provide insight into design and construction of masks, leading to increased protection and safety.

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Chapter 5 – Concluding Remarks

Research Summary

The research put forth in this thesis examines the etiology of impact related concussions from foul balls seen in catchers and umpires in baseball. By utilizing available video and observational technologies, concussive events in Major League Baseball were analyzed to determine the locations and speeds with which impacts to the mask occur. An understanding of these impact conditions led to an evaluation narrowing down the targets to only the most severe. With this knowledge in hand, a large sample of commonly available masks was tested under real-world conditions simulating concussive impacts. The results of this work will help clarify common questions surrounding the purchase of catcher equipment, such as whether newer hockey-style 1-piece masks perform better or worse than their traditional-style 2-piece counterparts, and whether or not lighter titanium caged masks provide increased protection to justify their increased cost as opposed to steel caged masks of the same type.

Although the mechanisms of concussive injuries from impact scenarios are still not well understood, this series of works presents the most comprehensive investigation into the evaluation of catcher and umpire mask performance to date. It is anticipated that results from these studies will help to spur industry wide changes in mask design and construction, ultimately leading to a safer product for use at all levels of play. Furthermore, the processes through which these evaluations were conducted have applications beyond the sport of baseball. By applying these techniques to other sport scenarios in which concussive impacts occur, considerable progress can be made towards improving equipment safety, specifically protective gear for the head and face

Publication Outline

The research completed and presented in this thesis is proposed to be published in several peer-reviewed journals. Table 10 presents the destination journal for each article.

Table 10: Outline of publication details for the 3 chapters presented in this thesis.

Chapter	Title	Journal
2	Analysis of Concussions in Major League Baseball Catcher and Umpires from Baseball Impacts	Clinical Journal of Sport Medicine
3	The Effect of Baseball Impact Location on Head Accelerations: Experiments with Catcher and Umpire Masks	Sports Biomechanics
4	A Biomechanical Analysis of Catcher and Umpire Masks in Baseball	Journal of Sports Sciences