# Biomechanics of Head Impacts in Soccer

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## SCHOLARLY ABSTRACT

An estimated 3.8 million sports-related concussions occur every year. Little research has been collected on soccer players, despite women's soccer having the third highest rate of concussion among all popular collegiate sports. The objective of this work was to evaluate multiple interventions that have been introduced to address the high rate of concussions in this population. Wearable head impact sensors were evaluated on their ability to accurately count and measure head impacts during a collegiate women's soccer season. Head impact exposure was quantified using video analysis of this season as well. Sensors were unable to accurately count impacts and reported nonsensical head acceleration measurements, indicating that data reported from head impact sensors should be interpreted with caution. The ability of soccer headgear to reduce linear and rotational head accelerations during common soccer impacts was examined in the laboratory. Ball-to-head and head-to-head impacts were performed at a range of speeds and impact orientations. Headgear resulted in small reductions during ball-to-head tests, which are not likely to be clinically relevant. In head-to-head tests, use of headgear on the struck head provided an overall 35% reduction in linear head acceleration, and a 53% reduction when another headgear was added to the striking head. The ten headgear tested varied greatly in performance. These data suggest that the use of protective headgear could reduce concussion incidence significantly in this population. Research presented in this thesis will inform soccer organizations on best practices for player safety with regard to head impacts.

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## GENERAL AUDIENCE ABSTRACT

Concussions in sports are an increasing concern for coaches, parents, and players. An estimated 3.8 million sports-related concussions occur every year. Little research has been collected on soccer players, despite women's soccer having the third highest rate of concussion among all popular collegiate sports. Various interventions have been proposed to address this high rate of concussion. The objective of this work was to evaluate some of these interventions. Wearable head impact sensors were evaluated on their ability to accurately count and measure all types of head impacts during a collegiate women's soccer season. The number and nature of head impacts was also gathered using video analysis. Results indicate that head impact sensors struggled with producing reliable data and thus we advise that data gathered using these types of sensors should be interpreted with caution. The ability of soccer headgear to reduce head impact severity during common soccer impacts was examined in the laboratory. Ball-to-head and head-to-head impacts were performed to evaluate headgear performance. Headgear resulted in only small reductions during ball-to-head tests. In head-to-head tests, use of headgear on the struck head provided an overall 35% reduction, and a 53% reduction when another headgear was added to the striking head. The ten headgear tested varied greatly in performance. These data suggest that the use of protective headgear could reduce concussion incidence significantly in this female soccer players. Research presented in this thesis will inform soccer organizations on best practices for player safety with regard to head impacts.

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# DEDICATION

I would like to dedicate this thesis to my mom, for her undying love and support, and my dad, for his constant guidance and inspiration. Thank you for always making me feel special and pushing me to strive for greatness in everything that I do. I love you.

~

I would like to first acknowledge my advisor, Dr. Steve Rowson. Thank you for encouraging me to grow as a student, researcher, and individual. Your mentorship and direction have meant so much to me, and I am so lucky to have had your guidance over the past two years. I would also like to thank my other committee members, Dr. Stefan Duma and Dr. Gunnar Brolinson.

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Last, but certainly not least, I would like to thank my friends back in Philadelphia and Richmond. I am so lucky to have forever friends in you all. Thank you for being such bright lights in my life.

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# CHAPTER 1: INTERVENTIONS ASSOCIATED WITH CONCUSSIONS IN SOCCER

## *Opening Remarks*

Little concussion research has been collected on soccer players, despite women's soccer having the third highest rate of concussion among all popular collegiate sports. The objective of this work was to evaluate multiple interventions that have been introduced to address the high rate of concussions in soccer players. Wearable head impact sensors have recently become available, and are being used to collect data in nonhelmeted sport populations such as soccer players. Soccer headgear has also become popular as a method of attenuating head accelerations during play. The research presented in this thesis addresses the relative accuracies of each of these interventions.

## *Wearable Head Impact Sensors*

For over a decade, football players have been instrumented with helmet-mounted accelerometers to record and measure head impacts. This provides valuable information regarding exposure and tolerance to head impacts in this population. Until recently, collecting this data for soccer players was not possible due to the lack of helmet use in this sport. Wearable head impact sensors have since become available, but have not yet been examined in their ability to accurately count and measure head impacts during regular play. The research presented in this thesis collects exposure data in women's soccer players through video analysis, and also compares these data with sensor output to evaluate wearable head impact sensor accuracy on the field. The exposure metrics gathered will contribute towards a better understanding concussion incidence in this population, and sensor evaluation will provide information for consumers regarding the efficacy of a new technology.

### *Soccer Headgear*

Many manufacturers are now creating headgear to attenuate head accelerations for soccer players. It has been shown that player-to-player contact, specifically head-to-head impact, during aerial challenges is the most likely activity to result in concussion. Soccer is unique in that it is the only sport that involves the intentional redirection of the ball with one's head, and therefore protection from headers is also of interest. The ability of currently available headgear to reduce concussion risk in common soccer head impact scenarios is presented in this study. These data have applications towards headgear design, performance regulation of headgear, and recommendations to athletes with regards to relative performance.

# CHAPTER 2: QUANTIFYING HEAD IMPACT EXPOSURE IN COLLEGIATE WOMEN'S SOCCER


## *Abstract*

Objective: The aim of this study was to quantify head impact exposure for a collegiate women's soccer team over the course of the 2014 season.

Design: Observational and prospective study.

Setting: Virginia Tech women's soccer games and practices.

Participants: Twenty-six collegiate level women's soccer players with a mean player age of 19 ± 1 years old.

Interventions: Participating players were instrumented with head impact sensors for biomechanical analysis. Video recordings of each event were used to manually verify each impact sustained.

Main Outcome Measurements: Head impact counts by player position and impact situation.

Results: The sensors collected data from a total of 17,865 accelerative events, 8,999 of which were classified as head impacts. Of these, a total of 1,703 impacts were positively identified (19% of total real impacts recorded by sensor), 90% of which were associated with heading the ball. The average number of impacts per player per practice or game was 1.86 ± 1.42. Exposure to head impact varied by player position.

Conclusions: Head impact exposure was quantified through two different methods, which illustrated the challenges associated with autonomously collecting acceleration data with head impact sensors. Users of head impact data must exercise caution when interpreting on-field head impact sensor data.

Key Words: concussion, gender, frequency, sensor, acceleration, football

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## *Introduction*

Over 265 million people worldwide are actively involved in the game of soccer and there is ongoing growth in the sport's popularity.[1] This consistent increase in the sport's prominence has led to higher awareness of potential injuries, including brain injuries. This is especially true considering the intentional redirection of the ball with one's head that is imperative in soccer, but not in any other sport. In a retrospective study, it was found that 62.7% of soccer players had suffered concussion symptoms in a given year.[2] Fuller et al. performed a case-control study of players sustaining head and neck injuries during FIFA tournaments and found the incidence of concussion in women to be 2.4 times higher than that in men.[3]

While the potential negative effects of repetitive concussions have been well studied, the effects of chronic exposure to subconcussive impacts are not yet understood. In order to investigate this, researchers have performed neurocognitive tests on soccer players and compared the results to those of control subjects.[2, 4-8] Some studies have noted deficits in attention, concentration, memory, and judgment in soccer players when compared with controls.[2, 4, 7] Electroencephalograph (EEG) tracings of the brain have also been shown to exhibit increased abnormalities in soccer players compared to nonsoccer players.[7] Other research, however, disputes these findings and criticizes the earlier studies for methodological flaws such as inadequate control groups and inability to account for subject variables including alcohol intake and previous concussion history.[5, 6, 8] Furthermore, it is unclear whether the observed neurological deficits are from repeated subconcussive head impacts or associated with a history of concussion. These studies highlight the need for further research on head impacts in soccer.

While there have been studies in the past that have examined women's soccer head impacts, most have been restricted to recreating impact situations in a laboratory setting in order to estimate head accelerations.[9-11] Naunheim et al. fitted elite level high school soccer players with a football helmet that was instrumented with a triaxial accelerometer.[12] Subjects were instructed to head a regulation size soccer ball kicked from a distance of approximately 30 yards while ball velocity was measured using a hand-held radar gun. Funk et al. subjected 20 volunteers to a soccer ball impact to the forehead to examine head and neck loading.[9] Participants were struck by a regulation adult size soccer ball traveling horizontally toward his/her forehead at four different speeds (5, 8.5, 10, and 11.5 m/s). The participants remained stationary and did not attempt to actively head the soccer ball. Bite blocks made from dental impressions of each subject were instrumented with two triaxial accelerometers and one angular rate sensor. Subjects were found able to tolerate head accelerations well over 20g and 2000 rad/$s^2$ without injury.

Hanlon and Bir have reported on-field head acceleration data collected from 24 girls' youth soccer players that were instrumented with a wireless head acceleration measurement system implanted into a soccer headband. The girls were then asked to participate in a single scrimmage and head acceleration measurements were recorded.[13] While providing valuable data about on-field head accelerations, this study only investigated scrimmages due to the challenge of getting players to wear non-required impact measurement equipment during competitive games.

Recent technological advancements in head impact sensors have now made more widespread on-field measurements possible. In order to better understand exposure to head impact in female soccer players, we aimed to quantify the head impact exposure in collegiate women's soccer through on-field study of head impacts. With this in mind, this work uses both video analysis and sensor-recorded measurements.

## *Materials and Methods*

A total of 26 players from the Virginia Tech Women's Soccer Team consented to participate in this IRB approved study. A total of 26 practices and 20 games were analyzed, however, not every player participated in each of these events. Three goalkeepers, nine midfielders, five forwards, and nine defensive players were instrumented. The subjects had an average height of 5'7" ± 2"and an average weight of 142.2 ± 14.7 lbs. Player age ranged from 18 to 22 with a mean age of 19 ± 1 years old. Forty two percent of instrumented players had experienced a soccer-related concussion at some point prior to the beginning of this study, while 15% had experienced two or more.

Head impact sensors (XPatch, X2 Biosystems, Seattle, WA) were placed on the mastoid process behind the ear of every participating player using an adhesive patch before each game and practice to record linear and rotational head accelerations (Figure 1). Each of these sensors consists of three linear accelerometers and three angular rate sensors. When any event exceeded an accelerometer reading of 10g, data acquisition was triggered and recorded 100ms of data (10ms before and 90ms after the event). Data were sampled at 1000 Hz and were downloaded from each sensor following each game and practice. Results collected were then compared to video recordings of each event in order to manually verify each impact sustained, as well as categorize the type of impact that occurred. All impacts were categorized according to defined impact types (Table 1). Two games were excluded from this process due to poor video quality.

**Figure 1. A head impact sensor containing three linear accelerometers and three angular rate sensors placed behind the ear of a player**

**Table 1. Impact types with descriptions.**

| Abbreviation | Meaning | Description |
|---|---|---|
| H | Header | Player intentionally headed the ball |
| UFH | Up for Header | Player attempted to head the ball but exact contact is unclear (i.e. player to player or ball to player) |
| Fall | Fell Down | Player fell down and head struck the ground |
| PPC | Player to Player Contact | Player's head was struck by another player's body |
| UBPC | Unintentional Ball to Player Contact | Player's head was unintentionally struck by the ball |

The head impact sensors used come equipped with a filter used to determine the validity of each impact recorded. The filter identifies each impact as either a real or false impact. For the purpose of this study, impacts were classified as shown in Table 2.

**Table 2. Classification of true positive, false positive, true negative, and false negative impacts for the purpose of this study.**

| | Visually identified | Sensor filter marked as real impact |
|---|---|---|
| True Positive (TP) | Yes | Yes |
| False Positive (FP) | No | Yes |
| True Negative (TN) | No | No |
| False Negative (FN) | Yes | No |

While this study used a 10g trigger to initiate data collection, we also investigated if a different trigger threshold would improve automated data collection in the field. Impact classifications were identified at various imposed linear acceleration thresholds and used to construct a positive predictive value (PPV) curve which indicated the percentage of positive recorded impacts that are true positive results (Equation 1). Furthermore, sensitivity and specificity were computed at each threshold. Receiver operating characteristic (ROC) curves were then generated to investigate the accuracy of the sensors. Separate curves were produced for the following three scenarios: games alone, practices alone, and games and practices combined.

$$\boldsymbol{PPV = \frac{True\ Positives}{True\ Positives + False\ Positives}} \qquad (1)$$

## *Results*

From a total 26 practices and 20 games, 1,703 impacts were identified through video analysis. Of these impacts, 89.6% were headers. The remaining impacts were a mixture of head to head, head to body, head to ground, and unintentional ball to head contact. The average number of impacts per player per practice was 1.69 ± 1.06 and the average number of impacts per player per game was 2.16 ± 2.77. The average number of impacts per player per practice or game was 1.86 ± 1.42. Impact totals varied by position (Figure 2, Figure 3). Total impact incidence breakdown can be seen in Table 3.

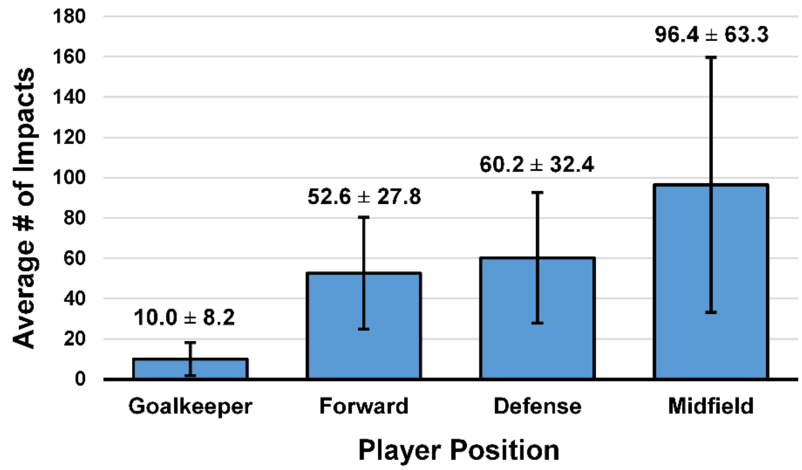


**Figure 2. Average number of impacts sustained per player over the course of a season divided into player positions. Midfielders sustained the highest average number of impacts per player while goalkeepers sustained the lowest. Forwards and defenders saw similar average number of impacts per player.**

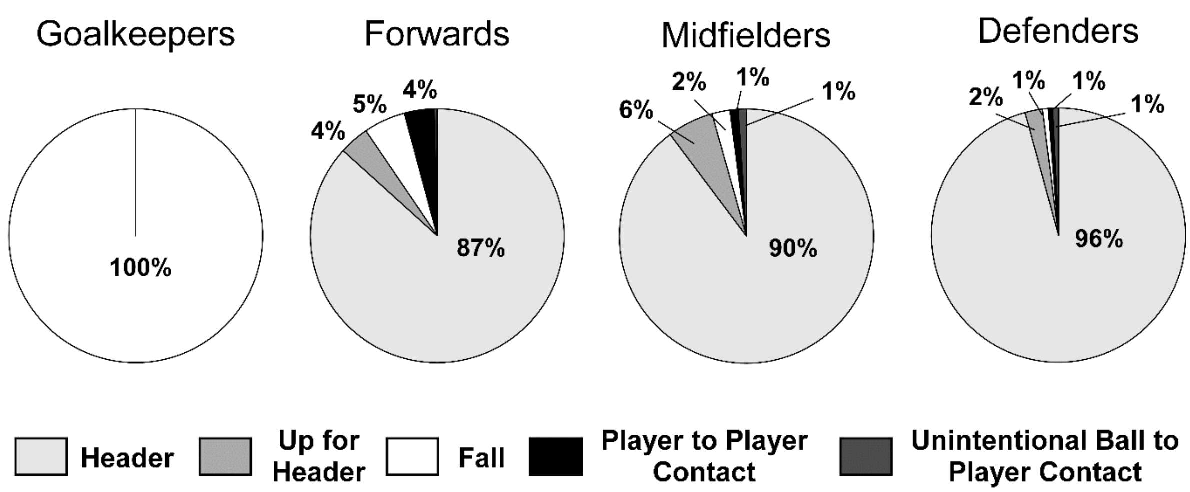


**Figure 3. Distribution of impact type by player position. Headers represented the majority of impacts for forwards, midfielders, and defenders. Goalkeepers only sustained head impacts from falling.**

**Table 3. Impact incidence breakdown across all players for all games and practices throughout the season (# G/P = # games/practices). The vast majority of impacts sustained throughout the season were headers, representing 90% of all 1,703 total impacts.**

| Player # | Position | # G/P | Total Impacts | Impacts/ Practice | Impacts /Game | H | UFH | Fall | PPC | UBPC |
|---|---|---|---|---|---|---|---|---|---|---|
| 1 | GK | 17 | 1 | 0.06 | 0.00 | 0 | 0 | 1 | 0 | 0 |
| 2 | D | 33 | 57 | 1.75 | 1.67 | 56 | 1 | 0 | 0 | 0 |
| 3 | M | 44 | 137 | 3.54 | 2.50 | 124 | 3 | 5 | 4 | 1 |
| 4 | F | 43 | 98 | 1.39 | 3.30 | 72 | 5 | 12 | 9 | 0 |
| 6 | D | 44 | 103 | 2.16 | 2.58 | 102 | 1 | 0 | 0 | 0 |
| 7 | M | 40 | 69 | 2.00 | 1.21 | 62 | 4 | 3 | 0 | 0 |
| 8 | D | 44 | 54 | 0.85 | 1.78 | 47 | 3 | 1 | 2 | 1 |
| 9 | M | 40 | 70 | 1.54 | 2.06 | 67 | 3 | 0 | 0 | 0 |
| 10 | D | 22 | 40 | 2.24 | 0.40 | 40 | 0 | 0 | 0 | 0 |
| 11 | D | 36 | 26 | 0.96 | 0.18 | 26 | 0 | 0 | 0 | 0 |
| 12 | F | 27 | 24 | 1.00 | 0.25 | 24 | 0 | 0 | 0 | 0 |
| 13 | F | 37 | 44 | 1.70 | 0.59 | 40 | 3 | 0 | 1 | 0 |
| 14 | D | 37 | 111 | 3.64 | 1.67 | 107 | 2 | 0 | 0 | 2 |
| 15 | M | 43 | 113 | 1.74 | 3.65 | 94 | 10 | 5 | 2 | 2 |
| 16 | M | 30 | 49 | 2.13 | 0.00 | 47 | 1 | 0 | 0 | 1 |
| 17 | F | 31 | 41 | 1.81 | 0.30 | 41 | 0 | 0 | 0 | 0 |
| 18 | M | 43 | 236 | 3.00 | 8.94 | 204 | 22 | 5 | 2 | 3 |
| 19 | D | 43 | 78 | 0.72 | 3.33 | 70 | 4 | 3 | 1 | 0 |
| 21 | M | 44 | 107 | 2.56 | 2.26 | 100 | 5 | 0 | 1 | 1 |
| 22 | M | 45 | 68 | 1.20 | 1.90 | 63 | 3 | 1 | 0 | 1 |
| 23 | F | 44 | 56 | 1.15 | 1.44 | 51 | 2 | 2 | 0 | 1 |
| 24 | D | 9 | 15 | 1.88 | 0.00 | 14 | 0 | 0 | 0 | 1 |
| 27 | M | 25 | 19 | 1.00 | 0.54 | 18 | 0 | 1 | 0 | 0 |
| 29 | D | 26 | 58 | 2.33 | 1.00 | 57 | 1 | 0 | 0 | 0 |
| 30 | GK | 38 | 12 | 0.21 | 0.50 | 0 | 0 | 12 | 0 | 0 |
| 31 | GK | 31 | 17 | 0.58 | 0.43 | 0 | 0 | 17 | 0 | 0 |
| **TOTAL:** | | **916** | **1703** | | | **1526** | **73** | **68** | **22** | **14** |
| **AVERAGE:** | | **35.2** | **65.5** | **1.69** | **2.16** | **58.7** | **2.8** | **2.6** | **0.8** | **0.5** |
| **STD. DEVIATION:** | | **9.6** | **50.0** | **1.06** | **2.77** | **45.0** | **4.6** | **4.5** | **1.9** | **0.8** |

Head impact exposure varied by position due to the different nature of play experienced by each individual. Midfielders saw the greatest average number of impacts (94 ± 63.3), followed by defenders (60.2 ± 32.4), forwards (52.6 ± 27.8), and goalkeepers (10.0 ± 8.2).

While goalkeepers exclusively experienced head impacts from diving/falling, all other positions primarily sustained head impacts associated with head to ball contact. Full impact mechanism breakdown by position can be seen in Figure 3. These findings are consistent with the observed nature of play: headers seemed to be fairly evenly distributed across all positions excluding goalkeepers, with defenders experiencing a slightly elevated percentage of headers expectedly due to their role of clearing the ball from the opposing team's side of the field.

A probability density function and cumulative distribution function characterizing head impacts per season between players were computed (Figure 4). There was a median of 56.5 total impacts per player and the interquartile range of the data was 63.5 impacts per player ($Q_1$=29.5, $Q_3$=93). A logistic fit was imposed on the data.

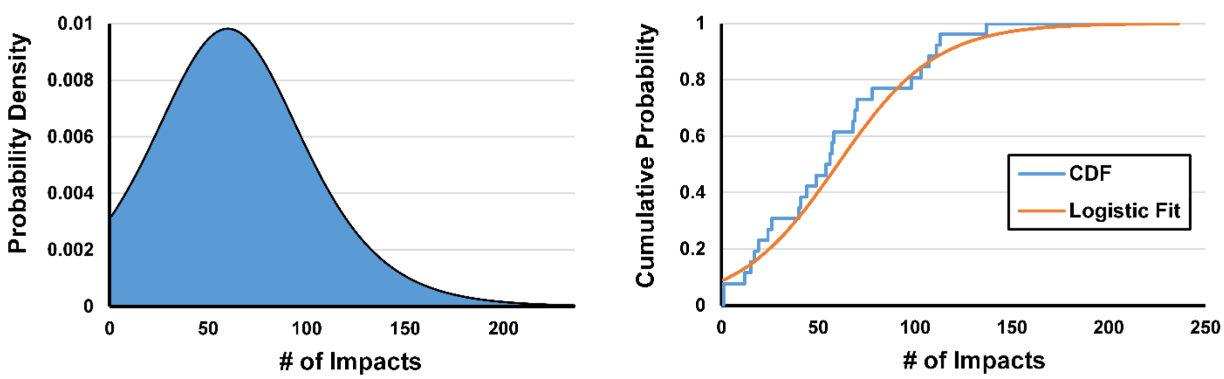


**Figure 4. Probability density function and cumulative distribution function for total impacts across all instrumented players throughout the season. A logistic distribution with μ = 60.01 AND σ = 25.45 was used to fit the data.**

Each impact recorded by the sensors was categorized according to the sensor performance classifications outlined in Table 2. A breakdown of these values for practices and games can be seen in Table 4. These impact classifications were used to construct a PPV curve (Figure 5). At a linear acceleration threshold of 10g, 16.3% of positive recorded impacts were true positive results. This value was maximized at 34g, where 65.8% of positive recorded impacts were true positive results. At this threshold there were 383 true positive impacts, 199 false positives, 15,963 true negatives, and 1,320 false negatives.

**Table 4. Sensor-recorded impact classifications for the duration of the season.**

| | True Positives | False Positives | True Negatives | False Negatives | Total Impacts Recorded |
|---|---|---|---|---|---|
| Practices Only | 780 | 3863 | 3664 | 208 | 8515 |
| Games Only | 683 | 3673 | 4962 | 32 | 9350 |
| **Practices and Games** | **1463** | **7536** | **8626** | **240** | **17865** |

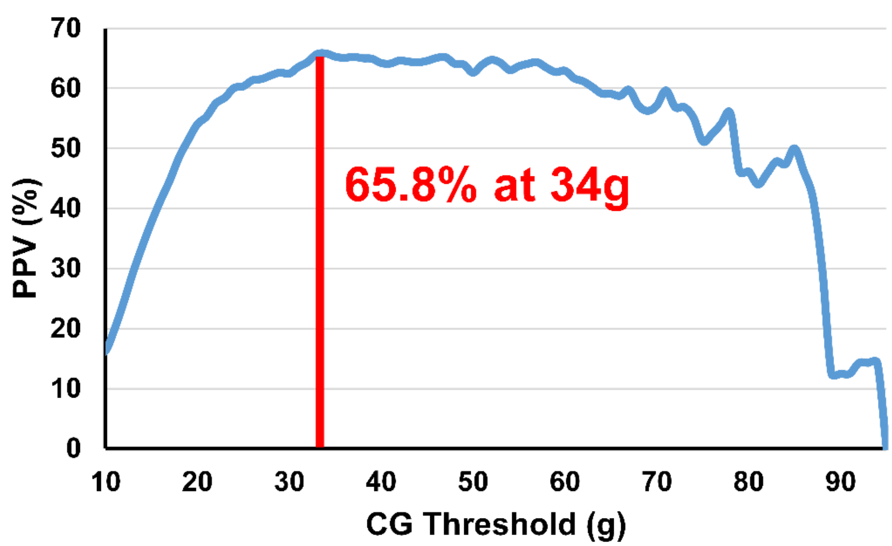


**Figure 5. Positive predictive value (PPV) of a head impact being correctly identified across a span of linear acceleration thresholds. PPV was maximized at a threshold of 34g, where 65.8% of positive recorded impacts were true positive results.**

The relationship between sensitivity and specificity for both practices and games, practices alone, and games alone was characterized with ROC curves (Figure 6). The area under the curve is 0.865 when combining practices and games, 0.797 for practices alone, and 0.938 for games alone. A case of random guessing would be indicated by an area of 0.500, while perfect classification would be indicated by an area of 1.000.

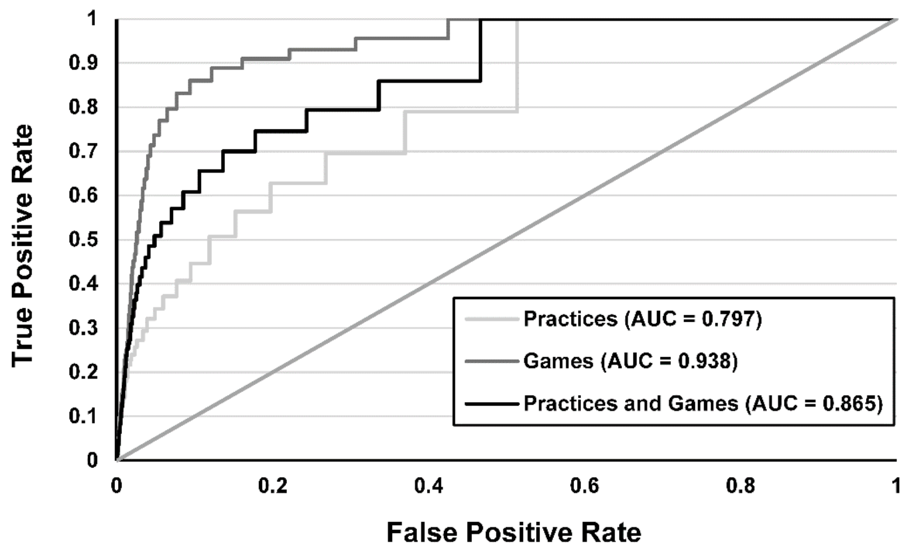


**Figure 6. ROC plot indicating sensor accuracy. An area under the curve of 1.0 would indicate perfect classification, while an area of 0.5 would indicate random guessing. Sensors are 18% more accurate with games than with practices.**

The linear acceleration values for all impacts ranged from 6g to 113g with an average of 25 ± 16g (Table 5). Rotational acceleration spanned from 380 rad/$s^2$ to 26,222 rad/$s^2$ and had an average of 5626 ± 4223 rad/$s^2$. Linear acceleration measurements for practices alone ranged from 6g to 113g with an average of 20 ± 13g. For games alone, measurements ranged from 7g to 94g with an average of 32 ± 18g. Rotational acceleration during practices ranged from 380 rad/$s^2$ to 26,222 rad/$s^2$ and averaged 4541 ± 3597 rad/$s^2$. For games, rotational acceleration ranged from 703 rad/$s^2$ to 24467 rad/$s^2$

with an average of 7126 ± 4555 rad/$s^2$. Caution must be exercised in interpreting these values, as they are much higher than one would expect and are likely prone to measurement error.

**Table 5. Impact measurement breakdown by type across all players for games and practices throughout the season.**

| | Linear Accel. (g) | | | | Rotational Accel. (rad/$s^2$) | | | |
|---|---|---|---|---|---|---|---|---|
| | Min | Max | Avg | Std Dev | Min | Max | Avg | Std Dev |
| H | 6 | 113 | 25 | 17 | 380 | 26222 | 5709 | 4281 |
| UFH | 7 | 89 | 27 | 17 | 699 | 17614 | 5785 | 3953 |
| Fall | 10 | 76 | 18 | 11 | 1042 | 13533 | 3507 | 2064 |
| PPC | 9 | 40 | 20 | 9 | 1330 | 13154 | 4177 | 3082 |
| UBPC | 10 | 84 | 35 | 22 | 3079 | 19017 | 8310 | 5039 |
| **All Impacts** | **6** | **113** | **25** | **16** | **380** | **26222** | **5626** | **4223** |

## *Discussion*

The sensor's algorithm reported a total number of impacts that was much greater than the number of impacts identified through video analysis (8,999 recorded compared to 1,703 confirmed). This difference illustrates some of the challenges associated with automated data collection using head impact sensors in sports. Not every accelerative event that a sensor identifies as being a head impact results from one. If the sensor alone was used to quantify head impact exposure, over 7,000 accelerative events not associated with impact would have been reported as head impacts resulting in inaccurate exposure counts. Jumping, running, and instances of players touching their sensors resulted in false positive accelerative events. The opposite also applies, in that a head impact can occur and the sensor does not recognize the accelerative event as being a head impact. When capturing head impact data with automated sensors in the field, it is vital that the accelerative events captured by the sensors are confirmed as actually occurring. This is true for all currently available head impact sensors, not just the sensor used in this study. For instance, we have used video analysis to confirm impacts throughout our football studies using the HIT System (Simbex, Lebanon, NH) for the past decade.[14-17]

While a threshold of 10g was used for this study, the positive predictive value (PPV) analysis enabled observation of sensor performance at higher thresholds. This analysis indicated that PPV would be maximized at 34g. However, this would still only result in 65.8% of positive recorded impacts being true positive results. Furthermore, previous research suggests that soccer headers typically only produce head accelerations less than 30g.[9, 10, 12, 13, 18-20] Therefore it would be impractical to use this optimal threshold for a soccer head impact study, seeing that the majority of the impacts would still be missed by the sensors. The ROC curves generated for these data actually look fairly good, which is mainly because the sensors did a good job of filtering out non-impact events. It should be noted, however, that this does not suggest that the sensor was good at identifying impact events.

Recorded head accelerations for confirmed impacts were also higher than expected, most notable for rotational acceleration. The kinetic parameters of linear and rotational head acceleration are commonly used to assess brain injury risk because they are thought to be indicative of the inertial response of the brain.[21, 22] Using a previously published risk curve which takes into account both linear and rotational head accelerations, it was estimated that 260 of the sensor's visually confirmed impacts were above 50% risk of concussion and 153 were above 90% risk.[21] Furthermore, some head acceleration values were in the range of events associated with severe brain injury, not just concussion.[23] Despite this, there were no head injuries (concussions or otherwise) identified over the course of the season. Given this context, it is unlikely that head acceleration was accurately measured for these confirmed head impacts. At first glance, one might think that acceleration magnitude must not be related to injury given that these values are so high. However, this is more a reflection of the high values being inaccurate. Over 60 years of biomechanical research has linked head kinematics to risk of brain injury.[24-27] It is unlikely that any player actually experienced the head accelerations measured by the sensor, which likely explains why no concussions were observed. Measurement error resulting from relative motion between the skin at the mastoid process and skull likely amplified computed head acceleration magnitudes. Modifying sensor placement to better couple the head impact sensor to skull motion will likely produce more accurate head acceleration data.

It should be noted that two games were excluded from this study due to poor video quality. In addition, we suspect that the sensor did not measure the biomechanics of the head impacts accurately and the results of this study should be interpreted accordingly. These errors are likely a result of relative motion between the skin and skull during head impact. The head impact sensors consist of hardware and software that can be upgraded to improve accuracy. Future iterations of the software and hardware could improve the accuracy of the sensor, and the version used in this study may not be the most up-to-date by the time of publication. Additional research will be necessary as on-field head impact sensors become more accurate.

This study has quantified head impact exposure for women's soccer players through video analysis of a collegiate women's soccer season. On average, players saw 65.5 ± 50.0 head impacts per season. Impact totals and types varied by position; which is consistent with the diverse nature of play experienced at different player positions. Headers represented the vast majority of total impacts sustained (89.6%). Furthermore, this study illustrated the challenges of using head impact sensors on the field, which are likely not specific to the sensor used. Data collected from head impact sensors should be interpreted with caution when used in an automated setting.

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# CHAPTER 3: BIOMECHANICAL PERFORMANCE OF HEADGEAR USED IN SOCCER


## *Abstract*

High incidence of concussion in soccer players has provoked increased efforts to reduce exposure to high magnitude head impacts in this population. Many manufacturers are now creating soccer headgear to attenuate head accelerations. However, no data are available that quantify their effectiveness. This study aimed to evaluate the ability of ten commercially available headgears to reduce concussion risk for soccer players in ball-to-head and head-to-head impacts. Ball-to-head impacts were recreated using a mounted soccer ball shooter to shoot balls at a NOCSAE headform, while head-to-head impacts were performed by accelerating one headform into another. Results from a total of 724 tests indicated that headgear did not meaningfully reduce head acceleration in ball-to-head impacts, but did result in large reductions in head acceleration during head-to-head impacts. Overall reductions in linear head accelerations compared to the control condition averaged about 35%. With the addition of another, matched headgear being worn on the striking headform, reductions increase to an average of about 53%. The ten headgear tested varied greatly in construction and composition, and thus exhibited diverse performance. Data will have applications towards headgear design, performance regulation of headgear, and recommendations to athletes with regards to relative performance.

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## *Introduction*

Sport-related concussions and their short- and long-term effects are a growing concern for athletes, coaches, and parents alike. Evidence suggests that these injuries can be detrimental to the mental health of athletes into the later years of life, causing deficits in memory, attention, concentration, and other functions.[1-11] A 2001 survey of retired professional football players found that players who self-reported concussions were at a greater risk for having depressive episodes later in life compared with those who self-reported no concussions.[10] Other researchers have examined motor cortex integrity in former collegiate athletes and reported that sports-related concussion incidence was positively correlated with long-term motor system dysfunctions.[11] A series of studies on professional soccer players reported that the number of concussions incurred was correlated to poorer results on many different neurocognitive tests.[1-3] These discoveries, among others, have inspired a movement to make sports safer for the brain by changing practice techniques, modifying game rules, and improving protective headgear.[12-14]

Soccer is well-known as the world's most popular and fastest-growing sport.[15] It is also the contact sport with the highest rate of concussion that does not require the use of protective headgear.[16] Recently, manufacturers have attempted to address this by creating various forms of protective headgear to dissipate head impact forces for soccer players. It has been shown that player-to-player contact, specifically head-to-head impact, during aerial challenges is the most likely activity to result in concussion.[17-21] Soccer is unique in that it is the only sport that includes the intentional redirection of the ball with one's head, therefore there is also concern associated with headers leading to neurocognitive changes.[15] Although headers typically do not generate the accelerations necessary to cause concussion, the effects of cumulative sub-concussive head impacts are not yet well-understood. Many studies have compared neurocognitive test performance between soccer players and control groups, with the majority reporting no significant differences.[22-27] There is a small subset of studies, however, which have reported inverse relationships between the number of ball impacts and certain neurocognitive performance scores in categories such as verbal learning, attention, strategic planning, and visual processing.[28-30]

Most currently available headgears claim to reduce the risk of concussion, however little research has been done to quantify the effectiveness of each headgear during various impact scenarios. In 2003, researchers propelled soccer balls at a standard magnesium headform and compared peak linear head accelerations with and without the use of protective headgear finding no measurable protection.[31] In 2005, Withnall et al. performed a combination of head-to-head dummy impacts, ball-to-head dummy impacts, and volunteer ball heading with and without the use of protective headgear.[32] This study also found that headgear provided no measurable protection during ball impacts. However, in head-to-head impacts, headgear was found to provide an overall 33% reduction in impact response. Both of these studies tested a very limited sample of headgear, and many of the models tested are no longer on the market today. This study aimed to evaluate the ability of ten commercially available headgears to reduce concussion risk for soccer players. Data will have applications towards headgear design, performance regulation of headgear, and recommendations to athletes with regards to relative performance.

## *Materials and Methods*

Efficacy of soccer headgear was evaluated during common impact scenarios identified by previous research. Individual headgear performance was evaluated in comparison to a control condition (bare headform) during ball-to-head and head-to-head impacts.

<u>Ball-to-Head Impacts</u>

A JUGS Soccer Machine (JUGS Sports, Tualatin, OR) was used to shoot official NCAA Wilson Forte Size 5 soccer balls at a medium NOCSAE headform mounted on a Hybrid III 50$^{th}$ percentile neck.[33, 34] A ball pressure of 10 psi and wheel pressures of 17 psi were maintained throughout testing. The soccer machine was modified to shoot balls in a horizontal path through an aluminum tube to ensure constant and repeatable impact conditions. A NOCSAE headform was used for this testing to provide a realistic fit between the headgear and the headform.[33] A custom adaptor plate was constructed to improve anatomic positioning of the neck relative to the head.[34] The head and neck assembly was contained in a cage and mounted onto a sliding mass intended to simulate the effective mass of the torso during impact (Biokinetics, Ottawa, Ontario, Canada) (Figure 1). The headform was instrumented with 3 linear accelerometers and 3 angular rate sensors to measure impact kinematics (Endevco 7264B-2000, Meggitt Sensing Systems, Irvine, CA)(ARS3 PRO-18K, DTS, Seal Beach, CA). Data were sampled at 20,000 Hz and filtered using a 4-pole Butterworth low pass filter with a cutoff frequency of 297 Hz (CFC 180) for accelerometer data and 256 Hz (CFC 155) for angular rate sensor data. Filters were chosen after preliminary testing. CFC 1000 was not used for linear acceleration due to noise (500 – 1000 Hz) associated with high rate impacts on the NOCSAE headform.

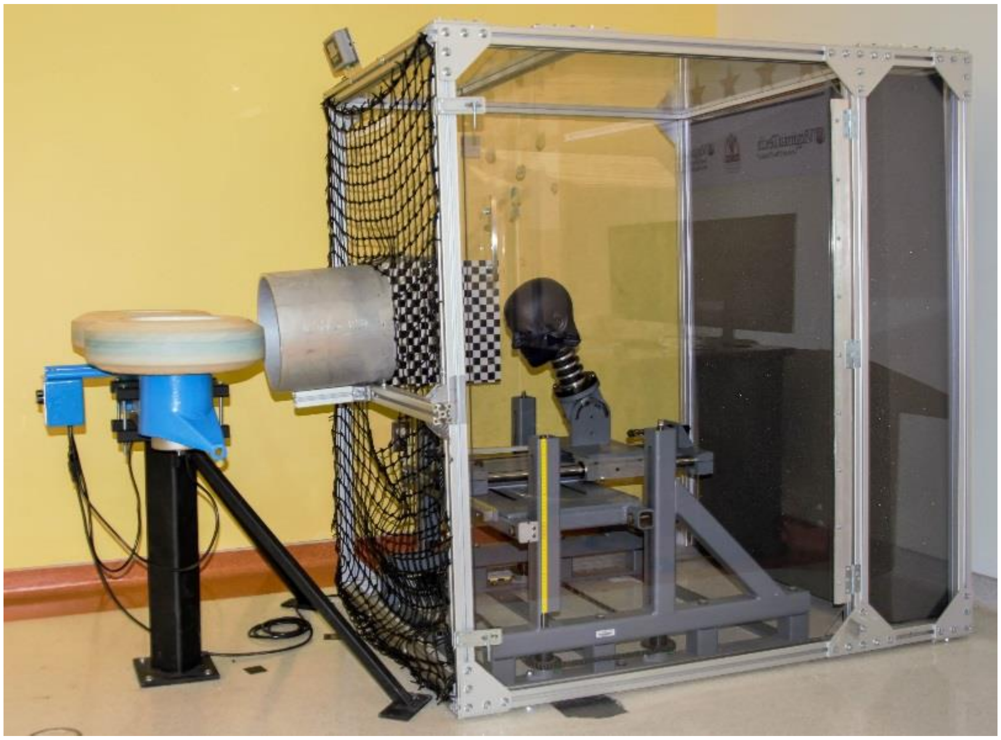
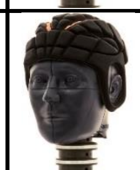


**Figure 1. A JUGS Soccer Shooter was mounted to shoot soccer balls horizontally towards a medium NOCSAE headform mounted on a Hybrid III 50th percentile neck.**

All headgear available to the public at the time of the study's onset were included in testing. Ten soccer headgears were purchased and tested along with a control condition (bare headform) (Table 1). Each headgear was fitted onto the headform and carefully positioned according to the instructions supplied by the manufacturer. The soccer ball is most commonly contacted by the player using the middle upper portion of the forehead.[35] To achieve this impact orientation, the headform was rotated forward 25 degrees about the y-axis. A total of 220 ball-to-head impact tests were performed; 10 trials for each of 2 ball velocities (15 m/s and 25 m/s) on each of 10 headgear as well as the control. A high speed video camera was used to calculate ball velocity for each trial using a frame rate of 2600 fps (Phantom v9.1, Vision Research, Inc., Wayne, New Jersey).

**Table 1. Ten commercially available soccer headgear were purchased for this study.**

| Headgear Name (Abbreviation) | Company | Company Location | | Headgear Name (Abbreviation) | Company | Company Location | |
|---|---|---|---|---|---|---|---|
| Don Joy Hat Trick (DJ) | DJO, LLC | Vista, CA |  | Full90 Select (F90S) | Full90 Sports, Inc. | San Diego, CA |  |
| ForceField Regular (FFR) | ForceField FF (NA) Ltd. | Great Neck, NY |  | Head Blast (HB) | Head Blast Soccer Band Company | St. Louis, MO |  |
| ForceField Ultra (FFU) | ForceField FF (NA) Ltd. | Great Neck, NY |  | Storelli ExoShield (STOR) | Storelli Sports | Brooklyn, NY |  |
| Full90 FN1 (FN1) | Full90 Sports, Inc. | San Diego, CA |  | Unequal Halo 10mm (UN10) | Unequal Technologies Company | Glen Mills, PA |  |
| Full90 Premier (F90P) | Full90 Sports, Inc. | San Diego, CA |  | Unequal Halo 6mm (UN6) | Unequal Technologies Company | Glen Mills, PA |  |

At each of the two ball speeds, peak linear and angular head accelerations were averaged over the 10 trials performed for each headgear to determine average peak head accelerations. Concussion risk was calculated using a bivariate risk function (Equation 1) where $a$ is peak linear acceleration, and $\alpha$ is peak rotational acceleration.[36] These values were compared using statistical software (JMP, Version 11, SAS Institute Inc., Cary, NC) to identify any significant differences from the control condition using Dunnett's method ($p < 0.05$).

$$Concussion\ Risk = \frac{1}{1+e^{-(-10.2+0.0433\cdot a+0.000873\cdot \alpha-0.000000920\cdot a\alpha)}} \tag{1}$$

<u>Head-to-Head Impacts</u>

To observe how these headgear perform in head impact conditions most commonly associated with concussion, a custom head-to-head impactor system was designed and built (Figure 2). The system consists of two head and neck assemblies each mounted onto sliding masses intended to simulate the effective mass of the torso during impact. Both torso masses are free to translate along their respective rail segments, one of which is extended to allow for about 70 inches of travel. An additional sliding platform is located behind one of the sliding torso masses, providing two functions: 1) accelerating the sliding torso mass, and 2) preventing extension of the neck during translation through its support bar attachment. The sliding platform is accelerated down the rail segment by its attachment to a cable and pulley system connected to a falling mass. The sliding torso mass is pushed by the platform, until the platform is stopped by the cable attachment 5 inches before the end of the rails. This allows the sliding torso mass free travel until impact. The impacted sliding torso mass was instrumented with 3 accelerometers and 3 angular rate sensors to measure impact kinematics. Data were sampled at 20,000 Hz and filtered using a 4-pole Butterworth low pass filter with a cutoff frequency of 1650 Hz (CFC 1000) for accelerometer data and 256 Hz (CFC 155) for angular rate sensor data.


**Figure 2. A custom impactor was designed and built to simulate head-to-head impacts in soccer. The system consists of two head and neck assemblies mounted onto sliding masses which are free to translate. One sliding mass is accelerated to a target velocity and impacts another, stationary sliding mass with free travel. The impacted mass is instrumented to measure impact kinematics.**

Two of the most prevalent soccer head impact scenarios described by Withnall et al. were investigated during this study: front boss of the striking head to the side of the struck head, and front boss of the striking head to the rear of the struck head.[32, 37] These conditions were replicated and described according to a global coordinate system (Table 2, Figure 2). All measurements are in relation to a “zero” condition where midsagittal plane and transverse plane markings on the NOCSAE headform were aligned with headforms facing each other. Both impact scenarios were performed under the following headgear conditions: a) bare head to bare head, b) bare head to headgear, and c) headgear to matched headgear. The same ten headgears used in the ball-to-head tests were used for this portion of the study. Each of the aforementioned conditions was repeated four times at three different impact velocities (2, 3, and 4 m/s), resulting in a total of 504 tests.

**Table 2. Two head-to-head impact locations were tested. These conditions are described for potential replication. All measurements are in relation to a "zero" condition where midsagittal plane and transverse plane markings on the NOCSAE headform were aligned with headforms facing each other. Directions are in relation to a global coordinate system (Figure ).**

| | Side Impact Location | | | | | | Back Impact Location | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| |  | | | | | |  | | | | | |
| | Struck Head | | | Striking Head | | | Struck Head | | | Striking Head | | |
| **Linear** | X | Y | Z | X | Y | Z | X | Y | Z | X | Y | Z |
| | 0 cm | +1 cm | +3.5 cm | -- | -- | -- | 0 cm | +3.8 cm | +5 cm | -- | -- | -- |
| **Rotation** | X | Y | Z | X | Y | Z | X | Y | Z | X | Y | Z |
| | 0° | 0° | +90° | -- | +20° | -50° | 0° | 0° | +180° | -- | +20° | -50° |

Peak linear and angular head accelerations were averaged over the four trials performed for each location/speed/headgear combination to determine average peak head accelerations. Concussion risk was calculated using a bivariate risk function (Equation 1). Values were compared using statistical software to identify any significant differences from the control condition using Dunnett's method ($p < 0.05$).

## *Results*

Ball-to-Head Impacts

Ball-to-head impacts were performed at two ball impact velocities: 15 and 25 m/s. High speed video analysis indicated that the average speed for the 15 m/s condition was 15.5 ± 0.3 m/s, and 25.1 ± 0.7 m/s for the 25 m/s condition. Overall linear and rotational head acceleration distributions with the use of protective headgear were generated (Figure 3).

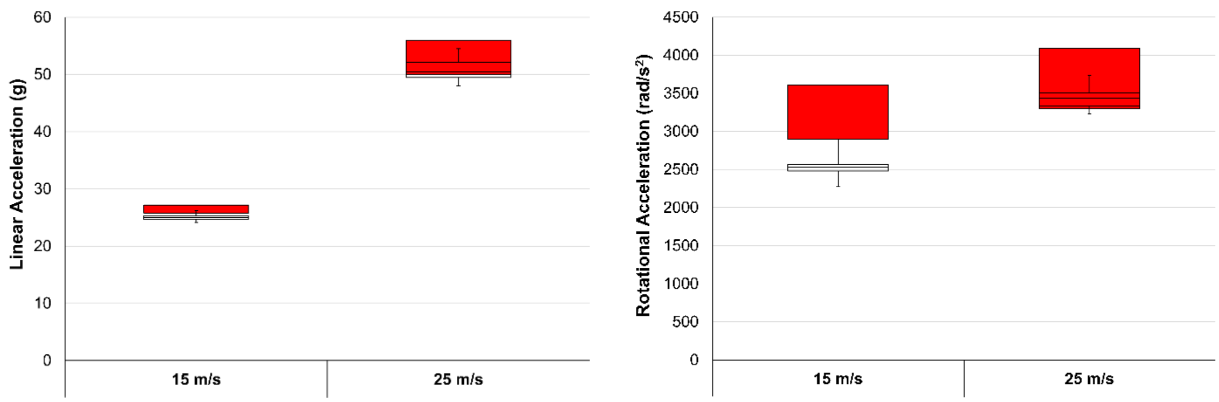



**Figure 3. Box plots denote the first, second, and third quartiles for head acceleration values with the use of protective headgear for all ball-to-head tests. Whiskers extend to minimum and maximum recorded values. Solid red boxes above the box plots indicate mean control accelerations ± one standard deviation. In some cases, the red control box overlaps the box plot. Small differences are observed between head accelerations with and without the use of protective headgear.**

At a ball velocity of 15 m/s, five of the ten headgear tested provided significant decreases in average peak linear head acceleration from the control condition (DonJoy, Full90 Premier, ForceField Ultra, Storelli ExoShield, and Unequal Halo 10mm). The greatest decrease was achieved with the use of the Unequal Halo 10mm headgear, which reduced average peak linear head acceleration by 2.4 g (Table 3). With respect to peak rotational acceleration, all headgears except the Full90 FN1 provided significant decreases from the control condition. The greatest decrease was achieved with the use of the Storelli

ExoShield, which reduced average peak rotational head acceleration by 982 rad/s$^2$. Concussion risk for all impacts was less than 1% (Figure 4, Equation 1).[36] Average concussion risk for all tests with the use of protective headgear conducted at the 15 m/s speed was 0.1 ± 0.0%, with all headgears providing a small, but significant decrease in concussion risk from the control.

When the ball velocity was increased to 25 m/s, three of the ten headgear tested provided significant decreases in average peak linear head acceleration from the control (DonJoy, Head Blast, and Storelli ExoShield) (Table 3). The greatest decrease was achieved with the use of the DonJoy, which reduced average peak linear head acceleration by 4.7 g. However, no headgears were able to significantly reduce peak rotational acceleration or concussion risk at this impact velocity. Average concussion risk for all tests conducted at the 25 m/s speed was 0.7 ± 0.3%. Concussion risk for all ball-to-head impacts was very low, with and without the use of protective headgear.

**Table 3. Average values ± one standard deviation for peak linear and rotational accelerations, as well as for concussion risk. At a ball velocity of 15 m/s, all headgear significantly reduced concussion risk from the control condition (bare headform). When the ball velocity was increased to 25 m/s, no headgear were able to significantly reduce concussion risk.**

| | 15 m/s | | | 25 m/s | | |
|---|---|---|---|---|---|---|
| **Headgear** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** |
| **CTRL** | **26.5 ± 0.7** | **3254 ± 359** | **0.2 ± 0.1** | **52.7 ± 2.7** | **3696 ± 399** | **0.8 ± 0.4** |
| DJ | 24.9 ± 1.3* | 2392 ± 274* | 0.1 ± 0.0* | 48.0 ± 4.0* | 3387 ± 199 | 0.5 ± 0.2 |
| F90P | 24.9 ± 1.1* | 2549 ± 206* | 0.1 ± 0.0* | 51.7 ± 3.2 | 3257 ± 652 | 0.5 ± 0.1 |
| F90S | 25.2 ± 1.2 | 2521 ± 178* | 0.1 ± 0.0* | 52.4 ± 2.4 | 3519 ± 508 | 0.8 ± 0.7 |
| FFR | 25.1 ± 1.0 | 2573 ± 293* | 0.1 ± 0.0* | 49.9 ± 0.9 | 3637 ± 495 | 0.7 ± 0.4 |
| FFU | 24.6 ± 1.2* | 2557 ± 290* | 0.1 ± 0.0* | 50.2 ± 1.8 | 3730 ± 648 | 0.8 ± 0.3 |
| FN1 | 26.1 ± 1.1 | 2898 ± 5223 | 0.1 ± 0.0* | 54.5 ± 2.6 | 3230 ± 360 | 0.7 ± 0.5 |
| HB | 26.2 ± 1.1 | 2475 ± 291* | 0.1 ± 0.0* | 49.4 ± 2.7* | 3313 ± 172 | 0.5 ± 0.2 |
| STOR | 24.4 ± 0.8* | 2272 ± 161* | 0.1 ± 0.0* | 49.1 ± 1.9* | 3465 ± 789 | 0.7 ± 0.4 |
| UN10 | 24.1 ± 1.2* | 2490 ± 257* | 0.1 ± 0.0* | 52.3 ± 2.3 | 3412 ± 251 | 0.6 ± 0.1 |
| UN6 | 25.3 ± 1.6 | 2633 ± 407* | 0.1 ± 0.0* | 50.7 ± 1.7 | 3485 ± 651 | 0.7 ± 0.4 |

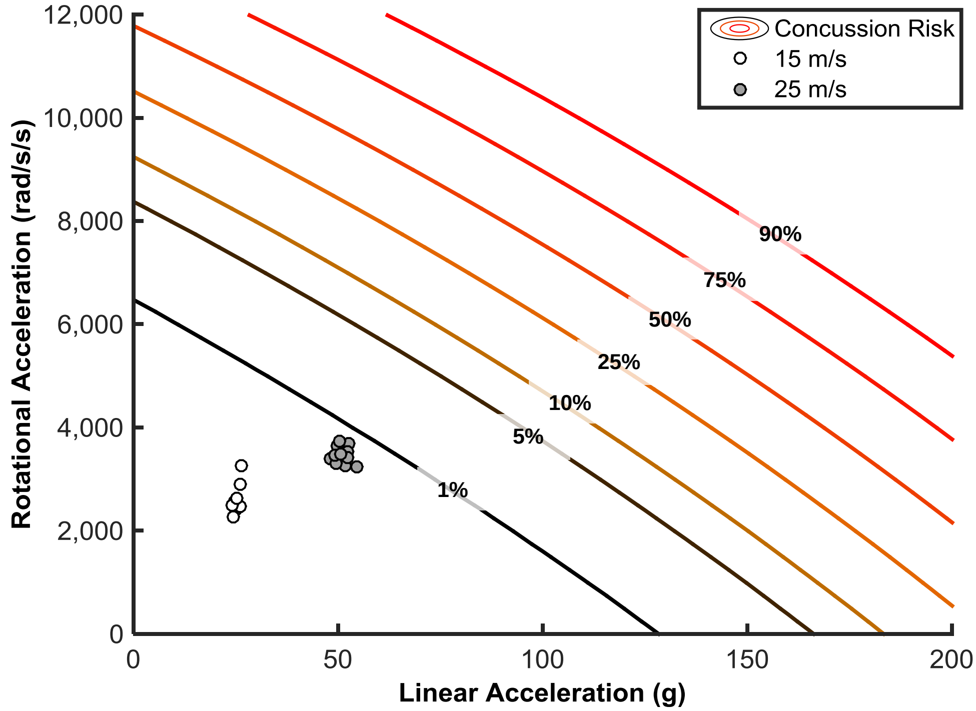



**Figure 4. Concussion risk for all ball-to-head impacts was below 1%, with or without the use of protective headgear.**

<u>Head-to-Head Impacts</u>

Head-to-head impact tests were performed at three target impact velocities: 2, 3, and 4 m/s. Five additional control tests were completed prior to the start of data collection to confirm that the desired impact velocities were accurate and repeatable. High speed video analysis of these tests indicated that the average velocity was 2.0 ± 0.0 m/s for the 2 m/s condition, 3.0 ± 0.0 m/s for the 3 m/s condition, 3.9 ± 0.0 m/s for the 4 m/s condition. As a whole, the head-to-head tests produced head accelerations ranging from well below 1% risk to well above 90% risk, thus providing information about headgear performance across the entire spectrum of potential head impacts in soccer (Figure 5). Linear and rotational acceleration were correlated for these impacts, decreasing proportionally between headgear. Two trends can be observed, each representative of a specific head impact location.

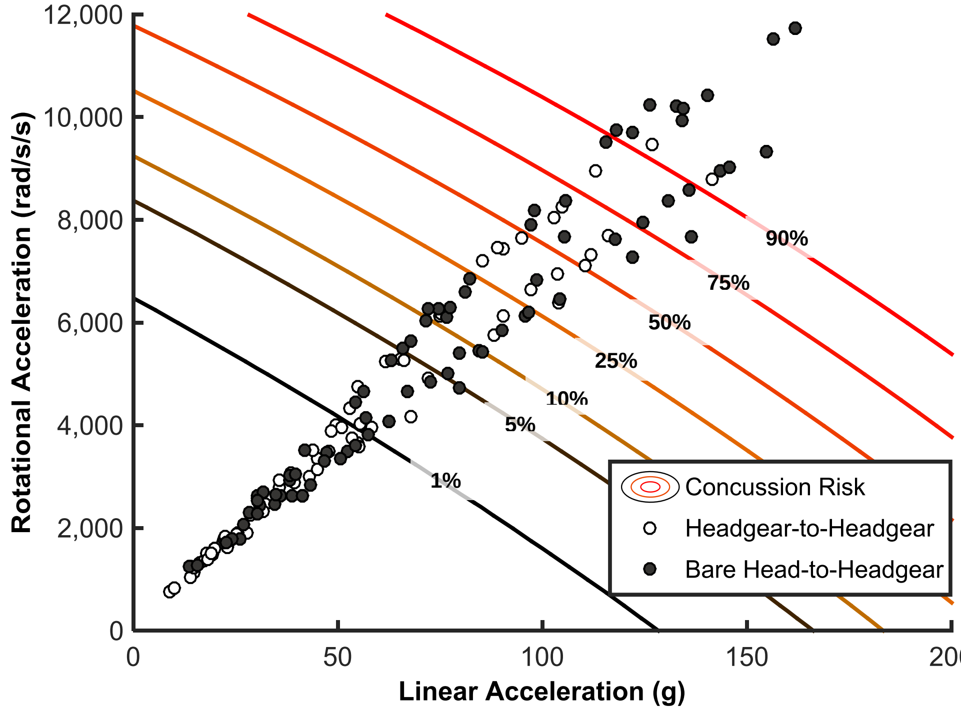



**Figure 5. Concussion risk for all head-to-head impacts ranged from less than 1% to greater than 90% risk.**

*Bare Head to Headgear*

Overall linear and rotational head acceleration distributions with the use of protective headgear were generated for each impact velocity and head orientation (Figure 6). At side location, all of the headgear tested were able to reduce peak linear and rotational head accelerations significantly from the control condition at the low (2 m/s) and (3 m/s) impact velocities. At the low impact velocity, the average reduction in linear acceleration across all headgear was 28 ± 9 g, whereas the average reduction in rotational acceleration was 2426 ± 792 rad/s$^2$. The greatest reductions in both linear and rotational head accelerations from the control condition were achieved using the Storelli

ExoShield (40 g and 3560 rad/s$^2$). At the medium impact velocity, the average reduction in linear acceleration across all headgear was 35 ± 16 g, whereas the average reduction in rotational acceleration was 2533 ± 1323 rad/s$^2$. The greatest reductions in both linear and rotational head accelerations from the control condition were again achieved using the Storelli ExoShield (66 g and 5312 rad/s$^2$). Concussion risk was also reduced significantly for all tests at the low and medium impact velocities. At the maximum velocity (4 m/s), the average reduction in linear acceleration across all headgear was 39 ± 21 g, and the average reduction in rotational acceleration was 2164 ± 1634 rad/s$^2$. greatest reductions in linear and rotational acceleration were achieved using the Storelli ExoShield (81 g and 5129 rad/s$^2$). Only the ForceField Regular failed to significantly reduce peak linear or rotational head accelerations from the control at this impact

In addition, five of the headgear tested did not significantly reduce concussion risk in the 4 m/s tests (Full90 Select, ForceField Regular, ForceField Ultra, Full90 FN1, and Head Blast). The Storelli ExoShield provided the greatest reduction in concussion risk at all impact velocities tested for this location (

Table 4).

At the back location, all of the headgear tested were able to reduce peak linear and rotational head accelerations significantly from the control condition at the lowest impact velocity. The average reduction in linear acceleration across all headgear was 29 ± 10 and the average reduction in rotational acceleration was 1734 ± 615 rad/s$^2$. The reductions in linear and rotational acceleration from the control were achieved using the Unequal Halo 10mm (47 g and 2807 rad/s$^2$). Concussion risk was also reduced significantly for these tests. At the medium impact velocity, the average reduction in acceleration across all headgear was 36 ± 21 g, whereas the average reduction in rotational acceleration was 1926 ± 1177 rad/s$^2$. The greatest reductions in linear and rotational accelerations from the control were achieved using the Unequal Halo 10mm (74 g and 4185 rad/s$^2$). The Head Blast failed to reduce rotational acceleration significantly at the medium impact velocity, although concussion risk was still reduced. When the impact velocity was increased to 4 m/s, the average reduction in acceleration across all headgear increased to 40 ± 28 g, and the average reduction in rotational acceleration was 2164 ± 1634 rad/s$^2$. The greatest reductions in linear and rotational acceleration were achieved using the Unequal Halo 10mm (97 g and 5512 rad/s$^2$). At the 4 m/s impact velocity, three headgear were unable to reduce rotational acceleration significantly (Full90 Select, ForceField Regular, and Head Blast), and five headgear were unable to reduce concussion risk significantly (Full90 Premier, Full90 Select, ForceField Regular, Full90 FN1, and Head Blast). The Unequal Halo 10mm provided the greatest reduction in concussion risk at all impact velocities tested for this location (

Table 4).

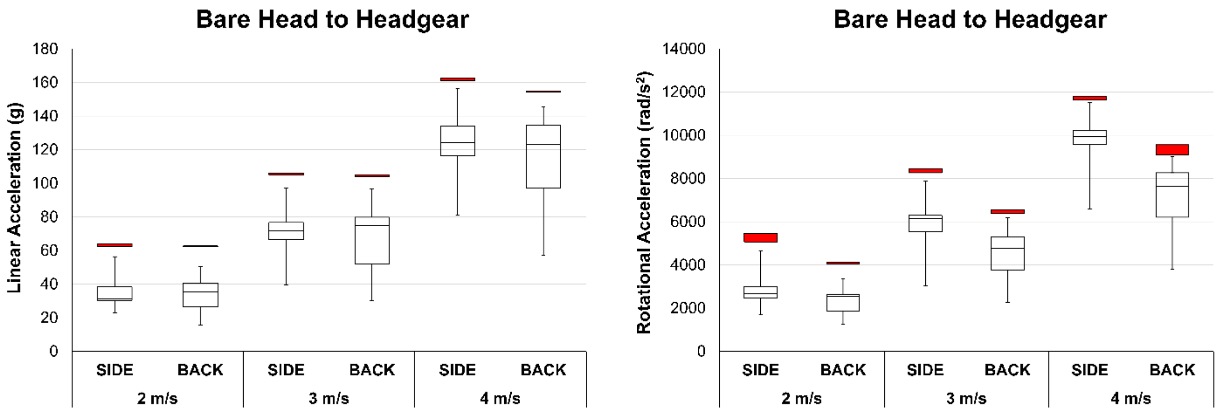



**Figure 6. Box plots denote the first, second, and third quartiles for head acceleration values with the use of protective headgear for all bare head to headgear tests. Whiskers extend to minimum and maximum recorded values. Solid red boxes above the box plots indicate mean control accelerations ± one standard deviation. Headgear were able to reduce linear and rotational accelerations significantly from the control condition.**

**Table 4. Almost all headgear were able to reduce linear and rotational head accelerations, and consequently concussion risk, significantly in bare head to headgear tests. Use of the Storelli ExoShield resulted in the greatest reduction in concussion risk at the side location, while the ForceField Regular provided the smallest reduction. At the back location, the Unequal Halo 10mm provided the greatest reduction in concussion risk, whereas the Head Blast provided the smallest reduction.**

| Side Location | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| Headgear | 2 m/s | | | 3 m/s | | | 4 m/s | | |
| | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) |
| **CTRL** | **63.1 ± 0.9** | **5266 ± 198** | **4.1 ± 0.7** | **105.6 ± 0.5** | **8360 ± 94** | **70.2 ± 1.7** | **161.8 ± 0.9** | **11728 ± 89** | **99.5 ± 0.0** |
| DJ | 30.4 ± 2.3* | 2631 ± 150* | 0.1 ± 0.0* | 66.0 ± 1.8* | 5496 ± 171 | 5.4 ± 1.0* | 115.5 ± 3.4* | 9522 ± 267* | 88.8 ± 3.3* |
| F90P | 30.8 ± 1.9* | 2434 ± 128* | 0.1 ± 0.0* | 68.0 ± 1.7* | 5642 ± 117 | 6.4 ± 0.9* | 122.0 ± 2.6* | 9700 ± 244* | 92.0 ± 2.1* |
| F90S | 31.9 ± 2.6* | 2692 ± 242* | 0.1 ± 0.0* | 74.6 ± 1.4* | 6273 ± 286 | 13.0 ± 3.0* | 132.7 ± 1.5* | 10220 ± 135* | 96.1 ± 0.5 |
| FFR | 56.4 ± 1.3* | 4647 ± 208* | 1.9 ± 0.4* | 97.2 ± 1.8* | 7892 ± 73 | 54.8 ± 2.1* | 156.4 ± 1.4 | 11518 ± 129 | 99.3 ± 0.1 |
| FFU | 38.2 ± 4.1* | 2926 ± 261* | 0.2 ± 0.1* | 82.1 ± 2.9* | 6856 ± 199 | 23.8 ± 4.4* | 140.4 ± 2.5* | 10420 ± 107* | 97.4 ± 0.4 |
| FN1 | 30.4 ± 3.6* | 2543 ± 433* | 0.1 ± 0.1* | 72.0 ± 2.9* | 6278 ± 278 | 12.0 ± 3.2* | 126.2 ± 3.4* | 10227 ± 323* | 95.1 ± 1.8 |
| HB | 38.3 ± 1.4* | 3017 ± 70* | 0.2 ± 0.0* | 77.4 ± 2.0* | 6292 ± 104 | 14.2 ± 1.9* | 134.3 ± 2.8* | 10172 ± 145* | 96.2 ± 0.8 |
| STOR | 22.8 ± 0.8* | 1707 ± 71* | 0.0 ± 0.0* | 39.6 ± 1.6* | 3048 ± 208 | 0.3 ± 0.1* | 81.0 ± 4.5* | 6599 ± 289* | 19.9 ± 6.0* |
| UN10 | 28.5 ± 1.5* | 2302 ± 172* | 0.1 ± 0.0* | 54.2 ± 2.2* | 4447 ± 81 | 1.5 ± 0.2* | 98.1 ± 3.2* | 8174 ± 267* | 60.8 ± 7.8* |
| UN6 | 41.8 ± 0.8* | 3507 ± 178* | 0.4 ± 0.1* | 71.6 ± 0.9* | 6041 ± 204 | 9.8 ± 1.5* | 118.1 ± 0.9* | 9741 ± 195* | 91.3 ± 1.4* |
| **Back Location** | | | | | | | | | |
| Headgear | 2 m/s | | | 3 m/s | | | 4 m/s | | |
| | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) | Peak Linear Accel. (g) | Peak Rotational Accel. (rad/$s^2$) | Concussion Risk (%) |
| **CTRL** | **62.5 ± 0.2** | **4077 ± 34** | **1.5 ± 0.0** | **104.3 ± 0.5** | **6463 ± 86** | **34.1 ± 1.7** | **154.6 ± 0.5** | **9322 ± 241** | **96.4 ± 0.7** |
| DJ | 38.7 ± 3.0* | 2622 ± 214* | 0.2 ± 0.7* | 72.7 ± 4.1* | 4839 ± 340* | 4.3 ± 1.6* | 122.0 ± 10.6* | 7274 ± 1556* | 63.7 ± 32.7* |
| F90P | 35.9 ± 2.8* | 2620 ± 239* | 0.2 ± 0.0* | 76.8 ± 3.5* | 5016 ± 255* | 5.6 ± 1.6* | 124.5 ± 3.3* | 7954 ± 289* | 76.9 ± 6.3 |
| F90S | 34.5 ± 2.9* | 2465 ± 221* | 0.1 ± 0.0* | 79.8 ± 2.1* | 5409 ± 168* | 8.3 ± 1.5* | 130.8 ± 1.8* | 8374 ± 173 | 85.3 ± 2.3 |
| FFR | 43.3 ± 2.7* | 2846 ± 111* | 0.3 ± 0.0* | 84.6 ± 4.0* | 5449 ± 321* | 10.4 ± 3.7* | 135.7 ± 2.7* | 8574 ± 348 | 88.6 ± 3.9 |
| FFU | 27.0 ± 3.3* | 2055 ± 295* | 0.1 ± 0.0* | 67.1 ± 4.3* | 4669 ± 345* | 3.1 ± 1.2* | 117.8 ± 4.1* | 7620 ± 183* | 67.2 ± 6.2* |
| FN1 | 41.3 ± 1.8* | 2630 ± 154* | 0.2 ± 0.0* | 79.6 ± 1.5* | 4734 ± 163* | 4.9 ± 0.7* | 136.3 ± 1.8* | 7669 ± 234* | 80.5 ± 3.4 |
| HB | 50.5 ± 0.3* | 3356 ± 133* | 0.5 ± 0.1* | 96.6 ± 0.7* | 6195 ± 96 | 23.9 ± 1.7* | 145.7 ± 1.4* | 9029 ± 42 | 94.2 ± 0.1 |
| STOR | 23.9 ± 0.9* | 1770 ± 100* | 0.0 ± 0.0* | 47.2 ± 1.0* | 3477 ± 144* | 0.5 ± 0.1* | 90.3 ± 4.0* | 5848 ± 241* | 16.2 ± 4.2* |
| UN10 | 15.8 ± 0.4* | 1270 ± 47* | 0.0 ± 0.0* | 30.3 ± 1.0* | 2277 ± 89* | 0.1 ± 0.0* | 57.4 ± 2.4* | 3810 ± 170* | 1.0 ± 0.2* |
| UN6 | 26.2 ± 0.6* | 1787 ± 155* | 0.1 ± 0.0* | 46.6 ± 1.5* | 3304 ± 191* | 0.4 ± 0.1* | 85.3 ± 2.1* | 5428 ± 44* | 10.1 ± 0.9* |

*Headgear to Matched Headgear*

Overall linear and rotational head acceleration distributions with the use of protective headgear were generated for each impact velocity and head orientation (Figure). At both locations and all three impact velocities, all headgear tested were able to reduce peak linear and rotational head accelerations significantly from the control condition. At the side location, the average reductions in linear acceleration across all headgear were 38 ± 7 g at the low impact velocity, 54 ± 14 g at the medium impact velocity, and 72 ± 24 g at the high impact velocity. The average reductions in rotational acceleration across all headgear were 3290 ± 512 rad/$s^2$ at the low impact velocity, 4157 ± 1200 rad/$s^2$ at the medium impact velocity, and 4521 ± 1700 rad/$s^2$ at the high impact velocity. At the low impact velocity, the greatest reductions in linear and rotational acceleration from the control were achieved using the Full90 FN1 (48 g and 4030 rad/$s^2$). However, at the medium and high impact velocities, the greatest reductions in linear and rotational acceleration were achieved using the Storelli ExoShield (74 g and 6043 rad/$s^2$ at the medium impact velocity, and 113 g and 7847 rad/$s^2$ at the high impact velocity). Every headgear tested significantly reduced concussion risk from the control condition at the low and medium impact velocities. At the highest impact velocity, the ForceField Regular failed to significantly reduce concussion risk (Table 5).

At the back location, the average reductions in linear acceleration across all headgear were 43 ± 9 g at the low impact velocity, 56 ± 17 g at the medium impact velocity, and 62 ± 29 g at the high impact velocity. The average reductions in rotational acceleration across all headgear were 2544 ± 552 rad/$s^2$ at the low impact velocity, 3067 ± 1081 rad/$s^2$ at the medium impact velocity, and 3288 ± 1785 rad/$s^2$ at the high impact velocity. At the low impact velocity, the greatest reduction in linear acceleration was achieved using the Unequal Halo 10mm (49 g), while the greatest reduction in rotational acceleration was achieved using the ForceField Ultra (3030 rad/$s^2$). At the medium and high impact velocities, the greatest reductions in both linear and rotational accelerations were achieved using the Unequal Halo 10mm (79 g and 4549 rad/$s^2$ for the medium impact velocity, and 110 g and 6180 rad/$s^2$ for the high impact velocity). Every headgear tested significantly reduced concussion risk from the control condition at the low and medium impact velocities. At the highest impact velocity, the Head Blast failed to significantly reduce concussion risk (Table 5).

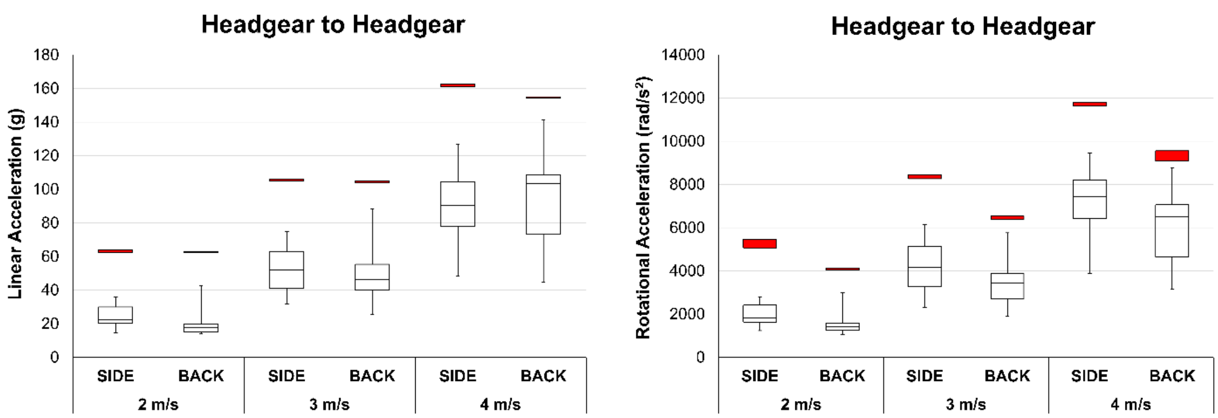



**Figure 7. Box plots denote the first, second, and third quartiles for head acceleration values with the use of protective headgear for all headgear to headgear tests. Whiskers extend to minimum and maximum recorded values. Solid red boxes above the box plots indicate mean control accelerations ± one standard deviation. Headgear were able to reduce linear and rotational accelerations significantly from the control condition, even more-so than in the bare head to headgear tests.**

**Table 5. Almost all headgear were able to reduce linear and rotational head accelerations, and consequently concussion risk, significantly in headgear to matched headgear tests. Use of the Storelli ExoShield resulted in the greatest reduction in concussion risk at the side location, while the ForceField Regular provided the smallest reduction. At the back location, the Unequal Halo 10mm provided the greatest reduction in concussion risk, whereas the Head Blast provided the smallest reduction.**

| **Side Location** | | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|
| **Headgear** | **2 m/s** | | | **3 m/s** | | | **4 m/s** | | |
| | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** |
| **CTRL** | **63.1 ± 0.9** | **5266 ± 198** | **4.1 ± 0.7** | **105.6 ± 0.5** | **8360 ± 94** | **70.2 ± 1.7** | **161.8 ± 0.9** | **11728 ± 89** | **99.5 ± 0.0** |
| DJ | 21.7 ± 1.3* | 1725 ± 169* | 0.0 ± 0.0* | 52.9 ± 2.6* | 4321 ± 273* | 0.3 ± 0.4* | 90.5 ± 2.1* | 7434 ± 430* | 40.2 ± 9.6* |
| F90P | 19.9 ± 1.0* | 1609 ± 128* | 0.0 ± 0.0* | 49.6 ± 1.8* | 3994 ± 205* | 0.9 ± 0.2* | 89.0 ± 3.5* | 7452 ± 382* | 39.2 ± 10.1* |
| F90S | 22.2 ± 2.0* | 1778 ± 144* | 0.0 ± 0.0* | 50.8 ± 4.9* | 3950 ± 355* | 0.9 ± 0.4* | 102.9 ± 4.2* | 8042 ± 287* | 62.3 ± 8.7* |
| FFR | 35.7 ± 3.8* | 2779 ± 282* | 0.2 ± 0.1* | 75.0 ± 3.9* | 6134 ± 120* | 11.9 ± 2.4* | 126.7 ± 2.7* | 9456 ± 74* | 91.9 ± 1.1 |
| FFU | 28.5 ± 2.3* | 2263 ± 85* | 0.1 ± 0.0* | 65.4 ± 2.3* | 5298 ± 214* | 4.6 ± 1.0* | 113.1 ± 3.8* | 8956 ± 251* | 82.6 ± 5.0* |
| FN1 | 14.7 ± 1.3* | 1236 ± 97* | 0.0 ± 0.0* | 35.7 ± 2.5* | 2925 ± 197* | 0.2 ± 0.0* | 75.3 ± 4.0* | 6165 ± 405* | 12.7 ± 4.5* |
| HB | 32.3 ± 1.8* | 2462 ± 173* | 0.1 ± 0.0* | 66.2 ± 1.9* | 5274 ± 228* | 4.6 ± 1.1* | 104.8 ± 3.4* | 8250 ± 88* | 67.7 ± 3.9* |
| STOR | 19.0 ± 0.6* | 1498 ± 98* | 0.0 ± 0.0* | 31.9 ± 1.0* | 2317 ± 122* | 0.1 ± 0.0* | 48.4 ± 2.3* | 3882 ± 364* | 0.8 ± 0.3* |
| UN10 | 22.6 ± 0.4* | 1831 ± 46* | 0.0 ± 0.0* | 38.4 ± 0.2* | 3071 ± 96* | 0.3 ± 0.0* | 61.8 ± 0.7* | 5243 ± 44* | 3.7 ± 0.2* |
| UN6 | 30.4 ± 0.4* | 2580 ± 147* | 0.1 ± 0.0* | 54.8 ± 0.3* | 4740 ± 79* | 1.9 ± 0.1* | 85.2 ± 0.7* | 7198 ± 140* | 31.3 ± 2.8* |
| **Back Location** | | | | | | | | | |
| **Headgear** | **2 m/s** | | | **3 m/s** | | | **4 m/s** | | |
| | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** | **Peak Linear Accel. (g)** | **Peak Rotational Accel. (rad/s$^2$)** | **Concussion Risk (%)** |
| **CTRL** | **62.5 ± 0.2** | **4077 ± 34** | **1.5 ± 0.0** | **104.3 ± 0.5** | **6463 ± 86** | **34.1 ± 1.7** | **154.6 ± 0.5** | **9322 ± 241** | **96.4 ± 0.7** |
| DJ | 16.3 ± 1.5* | 1319 ± 121* | 0.0 ± 0.0* | 47.7 ± 4.1* | 3500 ± 275* | 0.6 ± 0.2* | 103.6 ± 5.5* | 6940 ± 320* | 42.5 ± 10.6* |
| F90P | 14.8 ± 0.9* | 1131 ± 105* | 0.0 ± 0.0* | 44.9 ± 2.9* | 3344 ± 249* | 0.4 ± 0.1* | 97.2 ± 5.7* | 6637 ± 193* | 31.6 ± 7.4* |
| F90S | 17.4 ± 1.4* | 1367 ± 119* | 0.0 ± 0.0* | 58.2 ± 3.2* | 3955 ± 215* | 1.2 ± 0.3* | 110.5 ± 3.3* | 7101 ± 231* | 51.5 ± 7.2* |
| FFR | 19.4 ± 2.5* | 1483 ± 275* | 0.0 ± 0.0* | 55.6 ± 4.2* | 3993 ± 279* | 1.1 ± 0.4* | 112.0 ± 4.8* | 7310 ± 244* | 56.7 ± 8.7* |
| FFU | 14.1 ± 1.6* | 1046 ± 123* | 0.0 ± 0.0* | 39.4 ± 4.2* | 2893 ± 263* | 0.2 ± 0.1* | 90.5 ± 5.5* | 6127 ± 367* | 20.0 ± 7.1* |
| FN1 | 19.9 ± 1.4* | 1600 ± 92* | 0.0 ± 0.0* | 54.7 ± 3.1* | 3655 ± 67* | 0.8 ± 0.1* | 103.9 ± 2.3* | 6377 ± 322* | 32.6 ± 6.7* |
| HB | 42.9 ± 1.5* | 3005 ± 208* | 0.3 ± 0.1* | 88.2 ± 1.3* | 5759 ± 82* | 13.9 ± 0.2* | 141.4 ± 1.9* | 8782 ± 96* | 92.0 ± 1.0 |
| STOR | 17.8 ± 0.6* | 1496 ± 32* | 0.0 ± 0.0* | 31.2 ± 1.7* | 2313 ± 82* | 0.1 ± 0.0* | 53.5 ± 3.5* | 3757 ± 147* | 0.8 ± 0.2* |
| UN10 | 13.9 ± 0.4* | 1243 ± 100* | 0.0 ± 0.0* | 25.6 ± 0.8* | 1914 ± 131* | 0.1 ± 0.0* | 44.9 ± 1.0* | 3142 ± 129* | 0.4 ± 0.1* |
| UN6 | 23.1 ± 0.4* | 1627 ± 178* | 0.0 ± 0.0* | 41.2 ± 0.2* | 2636 ± 60* | 0.2 ± 0.0* | 67.8 ± 0.8* | 4163 ± 131* | 2.0 ± 0.3* |

## *Discussion*

A total of 724 tests were conducted to evaluate the ability of ten headgear to reduce head accelerations in common soccer head impacts. From these tests, it was determined that headgear did not provide substantial protection in ball-to-head impacts, but did result in large reductions in head acceleration during head-to-head impacts. The ten headgear tested varied greatly in construction and composition, and thus exhibited diverse performance.

### Ball-to-Head Impacts

Concussion risk for all tests in ball-to-head impacts were below 1%. This finding is consistent with current literature, which also indicates that heading a soccer ball is a very low-risk activity. From the few studies that have measured head accelerations during active heading of a soccer ball, peak linear accelerations range from 12 to 55 g with an average of 27 ± 12 g.[31, 38-40] Peak rotational accelerations range from 732 to 3003 rad/s$^2$ with an average of 1966 ± 879 rad/s$^2$.[38] These values are comparable to those measured in the current study, in which peak linear accelerations ranged from 24 to 55 g with an average of 38 ± 13 g, and peak rotational accelerations ranged from 2272 to 3730 rad/s$^2$ with an average of 3034 ± 493 rad/s$^2$. For context, the average concussion in football occurs around 100g and 5000 rad/s$^2$.[41, 42]

From a mechanical standpoint, the ball is a low mass, highly compliant object. This results in the ball deforming greatly upon impact and managing most of the energy transfer.

Therefore, when the player strikes the ball, resulting accelerations are low and acute injury from this activity is unlikely. The greatest reductions in linear and rotational head accelerations that were achieved during all ball-to-head impacts were 4.7g and 982 $rad/s^2$. Although these were statistically different from the bare headform head accelerations, the reduction of already minimal risk (<1%) by 0.1 – 0.3 % is not likely to be clinically relevant.

Head-to-Head Impacts

Head-to-head impacts are generally associated with higher concussion risk than ball-to-head impacts. Withnall et al. tested head-to-head impacts with similar impact conditions to those found in this study, at impact velocities of 2, 3, 4, and 5 m/s.[32, 37] Three soccer headgear were examined in the aforementioned study, one of which is no longer available for purchase. Withnall et al. recorded average peak linear accelerations with and without protective headgear ranging from 10 to 169 g with an average of 80 ± 49 g. Rotational head accelerations were not reported. The current study had comparable findings with respect to peak linear head accelerations, which ranged from 14 to 162 g with an average of 68 ± 40 g.

Most headgear tested in this study were able to provide a substantial and meaningful reduction in peak linear and rotational head accelerations from the control condition at all impact velocities and orientations. In the Withnall et al. study, it was found that across all headgear, overall reductions in linear head accelerations averaged about 33%. This is in close agreement with the present study, which finds that overall reductions averaged about 35%. With the addition of another, matched headgear being worn on the striking headform, reductions increase to an average of about 53%.

There were substantial differences in performance across the ten headgears tested (Table 6). These differences are most likely due to differences in headgear padding composition, thickness, and coverage. In general, thicker padding led to greater reductions in head accelerations and concussion risk. An overall metric was developed to compare headgear performance based on the six impact scenarios conducted in the bare head to headgear tests (one 2 m/s side impact, one 2 m/s back impact, one 3 m/s side impact, one 3 m/s back impact, one 4 m/s side impact, and one 4 m/s back impact). If a player experienced each of these impacts once, we could create a theoretical prediction of concussion incidence for that player by adding risks together to summarize overall performance.[36, 43]

**Table 6. The Storelli ExoShield resulted in the smallest overall incidence metric, while the ForceField Regular resulted in the largest (yet still less than the control condition).**

| Headgear | Overall Metric |
|---|---|
| **CTRL** | **3.06** |
| STOR | 0.37 |
| UN10 | 0.64 |
| UN6 | 1.12 |
| DJ | 1.63 |
| F90P | 1.81 |
| FFU | 1.92 |
| FN1 | 1.93 |
| F90S | 2.03 |
| HB | 2.29 |
| FFR | 2.55 |

If a player were to experience the six previously mentioned head impacts without wearing any protective headgear, that player would theoretically sustain about three concussions, on average. The ForceField Regular and the Head Blast provided small reductions in this theoretical incidence, while the top performing headgear, the Storelli ExoShield, was able to reduce the metric by a factor of 10. The Unequal Halo 10mm also provided very substantial reductions. To illustrate the variable performance in headgear, two extreme cases were examined in greater detail (Figure 8).

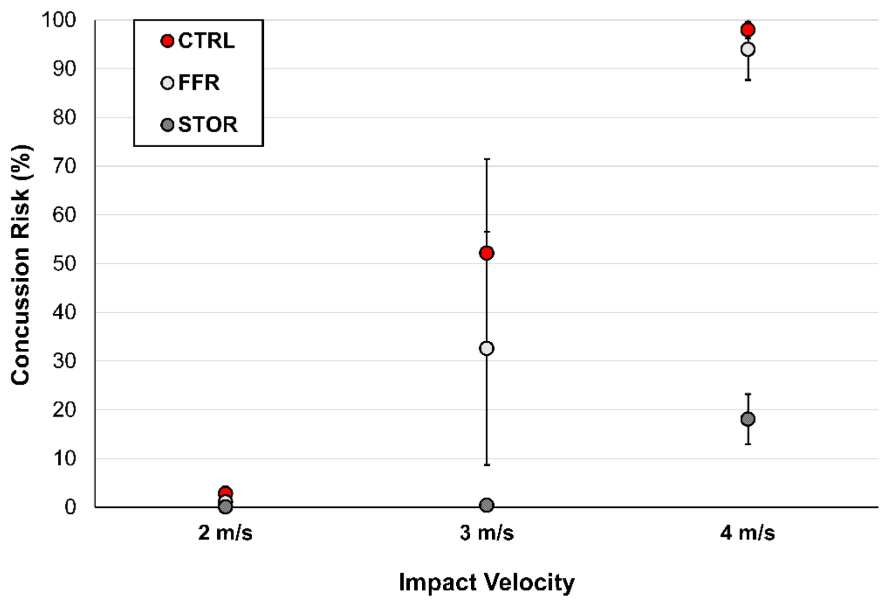



**Figure 8. Two individual headgear performances with respect to concussion risk compared to control (bare headform) performance. Mean risk ± one standard deviation is shown for each condition. Results for both impact orientations (side and back) were combined. The Storelli ExoShield resulted in the smallest overall incidence metric, while the ForceField Regular resulted in the largest (least different from control).**

This study was limited in several ways, the first of which is its restriction to laboratory recreations as opposed to on-field research. Several institutions have been instrumenting football players with head impact sensors to collect head acceleration data for the past 13 years.[44] These measurements are taken using helmet-mounted accelerometer arrays, making them unsuitable to measure impacts in unhelmeted sports such as soccer. Other head impact sensors have been developed for unhelmeted sports, however it has been shown that these devices struggle with both impact counting and acceleration measurements.[45] Future iterations of both software and hardware are needed before these sensors can be used reliably. This study is also limited due to the finite amount of impact locations examined. Furthermore, headgear were carefully positioned to ensure direct contact with the padding in an effort to evaluate the design of each headgear. In reality, any point on the head could be impacted during soccer play and soccer headgear would ideally provide protection in all impact scenarios, not just those tested in this study. Finally, the calculation of risk in this study was performed using a previously published risk equation developed to model concussion risk in football players (Equation 1). However, soccer players may not exhibit the same tolerance due to population-based differences.

This study comprehensively tested and evaluated currently available soccer headgear. Results indicate that the use of protective headgear could greatly reduce concussion incidence for athletes exposed to high level player-to-player contact, specifically head-to-head contact. If soccer leagues were to implement the use of protective headgear on a large scale, we could potentially see huge reductions in concussion incidence rates in this population.

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# CHAPTER 4: CLOSING REMARKS

## *Research Summary*

The research presented in this thesis evaluates multiple interventions that have been introduced to address the high rate of concussion in soccer players. By utilizing video analysis, head impact exposure for a collegiate women's soccer team was quantified and wearable head impact sensor performance was investigated. Exposure data will provide insight into understanding concussions in an understudied population. Sensor performance analysis illustrated the challenges associated with autonomously collecting head acceleration data with head impact sensors, and will inform consumers and researchers on the capabilities of currently available technology.

This study evaluated the ability of all currently available soccer headgear to reduce linear and rotational head accelerations through ball-to-head and head-to-head laboratory testing. The data presented will have applications towards headgear design, performance regulation of headgear, and recommendations to athletes with regards to relative performance. Results indicate that, while reductions in head accelerations in ball-to-head impacts were small, the adoption of protective headgear on in head-to-head impacts could reduce concussion risk significantly for soccer players. As a whole, the research presented in this thesis will inform soccer organizations on best practices for player safety.

### *Publication Outline*

The research presented in this thesis is intended to be published in several journals. Table 1 displays the publication destination for each research topic.

**Table 1. Publication plan for research presented in this thesis.**

| Chapter | Title | Journal | Status |
|---|---|---|---|
| 2 | Quantifying Head Impact Exposure in Collegiate Women's Soccer | Clinical Journal of Sports Medicine | Published |
| 3 | Biomechanical Performance of Headgear Used in Soccer | Annals of Biomedical Engineering | Preparing for Submission |