

## **Head Acceleration Measurements during Head Impact in Pediatric Populations**

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*This paper has not been screened for accuracy nor refereed by any body of scientific peers and should not be referenced in the open literature.*

### **ABSTRACT**

*The long-term objective of this study is to quantify biomechanical thresholds for concussion in pediatric populations. Knowledge of the biomechanics associated with pediatric head injury is lacking. The majority of research on the biomechanics of head injury has focused on skull fracture and traumatic brain injury (TBI) in an adult population through the utilization of cadaver and animal experiments. While these studies have advanced the head injury biomechanics field, they offer limited application to a pediatric population. There are size, shape, and developmental changes that occur in the human skull and brain during aging that likely result in biomechanical tolerance differences between humans of varying ages. Efforts to determine biomechanical tolerance thresholds for head injury in pediatric populations have been limited to using inertial scaling laws. Determinations of these thresholds involve no data derived from the representative population for which they were developed for. In this study, the helmets of youth football players, aged 7 to 18 years old, have been instrumented with accelerometer arrays that measure head acceleration each time their helmets are impacted during play. A total of 112 players under the age of 18 years have been instrumented and 31,557 head impacts have been recorded. Of these players, 8 sustained clinically diagnosed concussions. This paper presents a summary of these data as a function of age. Data will be collected for several more years in an effort to compile a comprehensive biomechanical dataset characterizing pediatric concussion. These data will have applications towards pediatric-specific injury metrics, computational modeling, and advanced dummy design.*

## INTRODUCTION

An estimated 1.6 to 3.8 million sports-related concussions occur annually in the United States (Langlois et al., 2006). This high incidence of concussion in sports provides scientists with the unique opportunity to collect biomechanical data to characterize mild traumatic brain injury (MTBI), or concussion. Due to football resulting in the highest incidence of concussion of any sport, researchers have been instrumenting the helmets of football players to record head acceleration data for every head impact players experience during play since 2003 (Duma et al., 2011). The archetypal thought of such research is: by observing a population that is at high risk for sustaining concussion, data characterizing concussion can be obtained in a natural and ethically sound manner. These data have applications in developing brain injury prediction tools with applications in automotive, military, and sports environments.

In 2003, researchers began instrumenting the helmets of Virginia Tech collegiate football players with commercially available accelerometer arrays to collect head acceleration data during head impacts of varying magnitudes from living subjects (Duma et al., 2005). Over the next decade, researchers from around the country adopted this technology to further our understanding of the biomechanical response to head impact in humans (Broglia et al., 2009, Broglia et al., 2012, Crisco et al., 2011, Crisco et al., 2012, Duma et al., 2009, Guskiewicz et al., 2007, Guskiewicz et al., 2011, Mihalik et al., 2007, Rowson et al., 2009, Schnebel et al., 2007). A comprehensive dataset biomechanically characterizing head impact and concussion in adults was compiled from these works consisting of over 1.5 million head impacts and 105 concussions (Beckwith et al., 2013, Beckwith et al., 2013). These data have greatly advanced the current understanding of the biomechanics of concussion in adults and have been used to develop concussion risk functions from the head kinematics resulting from impact (Rowson et al., 2011, Rowson et al., 2012, Rowson et al., 2013, Takhounts et al., 2011). Furthermore, these data have been used to help validate injury prediction tools in finite element head models (Takhounts et al., 2008).

While these studies have advanced the head injury biomechanics field, they offer limited application to a pediatric population. There are size, shape, and developmental changes that occur in the human skull and brain during aging that likely result in biomechanical tolerance and physiologic differences between humans of varying ages (Danelson et al., 2008, Meehan et al., 2011). Efforts to determine biomechanical tolerance thresholds for head injury in pediatric populations have been limited simple inertial scaling laws in the automotive environment (Mertz et al., 2003, Ommaya et al., 1967). Furthermore, these efforts investigated more severe head injuries, offering little insight towards concussion biomechanics. The long-term objective of this study is to quantify biomechanical thresholds for concussion in pediatric populations. A multi-year effort emulating the head acceleration measurements collected from a collegiate football population is being performed in a population under 18 years old. These data will have applications towards pediatric-specific injury metrics, computational modeling, and advanced dummy design. This paper summarizes the first year of data collection.

## METHODS

Head impact data were collected from 112 football players, age 7 to 18 years, on 6 football teams in Virginia and North Carolina instrumented with the in-helmet accelerometer arrays during the 2012 football season. The 6 football teams consisted a 7 to 8 year old team, two 9 to 11 year old teams, a 10 to 12 year old team, a middle school football team (12-14 year old), and a high school football team (14 to 18 year old). For the purpose of reporting the results, players will be group into 4 age categories: 7 to 8 years, 9 to 12 years, 12 to 14 years, and 14 to 18 years. Further description of the subjects is presented in Table 1. Players were monitored during each of the teams' games and contact practices. Approval for this study was given by the Virginia Tech and Wake Forest University Institutional Review Boards (IRBs). Each player provided assent and their parent/guardian gave written informed consent for participation in the study.

Helmets were instrumented with one of two accelerometer arrays that were used in parallel: the commercially available Head Impact Telemetry (HIT) System (Simbex, Lebanon, NH) or a custom 6 degree of freedom (6DOF) head acceleration measurement device (Beckwith et al., 2012, Rowson et al., 2009, Rowson et al., 2011). A total of 107 players were instrumented with the HIT System, which consists of 6 accelerometers mounted normal to the surface of the head. Each accelerometer is mounted on an elastic base to ensure that head acceleration is measured rather than helmet shell vibrations (Manoogian et al., 2006). During play, any time a single accelerometer experiences an acceleration greater than 14.4 g, data acquisition is automatically triggered. Data were

collected with a sampling rate of 1000 Hz for a total of 40 ms, which included 8 ms of pre-trigger data. Once data collection was complete, data were wirelessly transmitted to a computer on the sideline. The accelerations measured by individual accelerometers were then input into an algorithm that computes resultant linear acceleration at the center of gravity of the head and estimates peak rotational acceleration (Crisco et al., 2004, Rowson et al., 2012). Resultant linear head accelerations less than 10 g were removed from the dataset. These accelerations can be associated with non-impact events, such as jumping or running, and were not considered clinically relevant in the context of the study. The HIT System has previously been validated and been shown to have error for individual measurements of 6.3% +/- 15.7% for linear acceleration and -1.2% +/- 31.7% for rotational acceleration (Beckwith et al., 2012). In distribution analyses of large datasets, the variance of this error greatly diminishes.

Table 1. Descriptive statistics of instrumented players by age grouping.

Age Group	Player Mass (kg)	Player Age (years)	Number of Players
7 to 8 years	32.5 +/- 8.1	7.8 +/- 0.4	12
9 to 12 years	44.2 +/- 7.2	11.0 +/- 1.1	50
12 to 14 years	55.0 +/- 10.0	13.0 +/- 0.8	10
14 to 18 years	89.0 +/- 16.3	17.1	40

A total of 12 players were instrumented with the 6DOF head acceleration measurement device that consists of 12 accelerometers mounted in the crown of the helmet. Accelerometers are oriented tangential to the surface of the head and positioned around the crown in orthogonal pairs (Rowson et al., 2011). Data are collected from the accelerometers in the same way as the HIT System. The novel sensor design allows the calculation of linear and rotational acceleration about each axis of the head (Chu et al., 2006, Rowson et al., 2011). The 6DOF head acceleration measurement device has been previously validated and show to have error for individual measurements of 1% +/- 18% for linear acceleration and 3% +/- 24% for rotational acceleration (Rowson et al., 2011). As with the HIT System, the variance of this error greatly diminishes in distribution analyses of large datasets.

The two accelerometer arrays were used concurrently to increase the sample size of the study. Ideally, the helmets of all players would be instrumented with custom 6DOF head acceleration measurement devices because it provides the full three dimensional impact response of linear and rotational acceleration throughout time. However, only a small portion of players were instrumented with the 6DOF head acceleration measurement device due to the high cost (\$10,000 per helmet) and limited quantities. In previous studies, both the HIT System and 6DOF head acceleration measurement device have been validated and shown to produce the same response through laboratory testing and on-field data collection (Beckwith et al., 2012, Rowson et al., 2011, Rowson et al., 2012). Minimal variability between the two systems is anticipated.

Impacts were verified with video footage to remove any impacts collected while the players were not wearing their helmets. Descriptions of the distributions of impacts collected are presented as a function of age. Impact location was generalized into 1 of 4 categories (front, side, top, of back of the helmet) based on the acceleration vectors from the accelerometers (Greenwald et al., 2008). Furthermore, head accelerations associated with concussions are presented on an individual basis. The following protocol was used to identify and diagnose concussions during this study: coaches, parents, and players were informed of concussion signs and symptoms at the beginning of the season. All parties were urged to seek a physician if a player exhibited or reported any symptoms of a concussion. Concussions were only recorded if they were formally diagnosed by a physician.

## RESULTS

A total of 31,557 impacts were collected from the 112 instrumented football players. These players sustained a total of 8 concussions. Data are summarized below as a function of age due to the variability in impact distributions at each level of play (Table 2).

Table 2. Summary of head impacts collected as a categorized by age group.

	<b>7 to 8 Years</b>	<b>9 to 12 Years</b>	<b>12 to 14 Years</b>	<b>14 to 18 Years</b>
<b>Impacts</b>	3059	11,978	2098	16,502
<b>Median (g)</b>	16	19	21	22
<b>95<sup>th</sup> Percentile (g)</b>	37	46	61	58
<b>Concussions</b>	0	4	2	2

### **7 to 8 Years**

A total of 3059 impacts were collected from the 19 instrumented players between the ages of 7 and 8 years. The average instrumented player participated in 16 +/- 6 sessions, including 10 +/- practices and + +/- 2 games. Players sustained an average of 161 +/- 111 head impacts throughout data collection. The overall distributions of linear and rotational head accelerations were heavily right-skewed with a large number of low magnitude impacts. Linear accelerations ranged from 10 g to 111 g, with a median value of 16 g and a 95<sup>th</sup> percentile value of 37 g. Rotational accelerations ranged from 3 rad/s<sup>2</sup> to 7694 rad/s<sup>2</sup>, with a median value of 621 rad/s<sup>2</sup> and a 95<sup>th</sup> percentile value of 2016 rad/s<sup>2</sup>. Impacts to the front of the helmet were most common, accounting for 42% of the total number of impacts. Impacts to the back, side, and top of the helmet accounted for 24%, 23%, and 11%, respectively. Impacts to the top of helmet were more likely to result in greater linear accelerations, while impacts to the front of the helmet were more likely to result in greater rotational accelerations. A total of 125 impacts were associated with peak linear accelerations of 40 g or greater. A total of 11 impacts were associated with peak linear accelerations greater than 80 g. No diagnosed concussions were sustained by instrumented players in this age group.

### **9 to 12 Years**

A total of 11,978 impacts were collected from the 50 instrumented players between the ages of 9 and 12 years. The average player participated in 21.8 +/- 5.7 sessions and sustained an average of 240 +/- 147 impacts. The overall distributions of linear and rotational head accelerations were heavily right-skewed, being weighted towards low magnitude impacts. Linear accelerations ranged from 10 g to 126 g, with median value of 19 g and a 95<sup>th</sup> percentile value of 46 g. Rotational acceleration values ranged from 4 rad/s<sup>2</sup> to 5838 rad/s<sup>2</sup>, with a median value of 890 rad/s<sup>2</sup> and a 95<sup>th</sup> percentile value of 2081 rad/s<sup>2</sup>. Impacts to the front of the helmet were most common, representing for 41% of the total number of impacts. Impacts to the back, side, and top of the helmet accounted for 25%, 23%, and 11%, respectively. Impacts to the top of helmet were more likely to result in greater linear accelerations, while impacts to the front of the helmet were more likely to result in greater rotational accelerations. A total of 961 impacts were associated with peak linear accelerations of 40 g or greater. A total of 36 impacts were associated with peak linear accelerations greater than 80 g. Three of the instrumented players sustained diagnosed concussions.

### **12 to 14 Years**

A total of 2098 impacts were collected from the 10 instrumented players between the ages of 12 and 14 years. The average player participated in 19 +/- 7 practices and 4 +/- 1 games, while sustaining an average of 210 +/- 162 head impacts. These numbers are lower than the previous age group due the players not being instrumented for the full season. The overall distributions of linear and rotational accelerations were heavily right-skewed, similar to the impact distributions of previous age groups. Linear accelerations ranged from 10 g to 150 g, with a median value of 21 g and a 95<sup>th</sup> percentile value of 61 g. Rotational accelerations ranged from 4 rad/s<sup>2</sup> to 9019 rad/s<sup>2</sup>, with a median value of 898 rad/s<sup>2</sup> and a 95<sup>th</sup> percentile value of 2571 rad/s<sup>2</sup>. Impacts to the front of the helmet were the most common, accounting for 39% of all impacts. Impacts to the rear of the helmet accounted for 33% of all impacts. The sides and top of the helmet were impacted least frequently, each accounting for 14% of all impacts. Impacts to the top of the helmet most frequently exhibited the greatest linear acceleration magnitudes. A total of 329 impacts greater than 40 g and 48 impacts greater than 80 g were collected. Two instrumented players sustained diagnosed concussions during data collection.

## 14 to 18 Years

A total of 16,502 impacts were collected from the 40 instrumented high school football players during 47 sessions that included 33 practices and 14 games. The overall distributions of linear and rotational acceleration were highly right-skewed, being weighted more towards low magnitude impacts. Linear accelerations ranged from 10 g to 152 g, with a median value of 22 g and a 95<sup>th</sup> percentile value of 58 g. Rotational accelerations ranged from 3 rad/s<sup>2</sup> to 7701 rad/s<sup>2</sup>, with a median value of 973 rad/s<sup>2</sup> and a 95<sup>th</sup> percentile value of 2481 rad/s<sup>2</sup>. Impacts to the front of the helmet were most common (45%), followed by the back (22%) and top (15%) of the helmet. Impacts to the top of helmet were more likely to result in greater linear accelerations, while impacts to the back of the helmet were more likely to result in greater rotational accelerations. There were a total of 1650 impacts greater than 45.5 g and 165 impacts greater than 86.7 g collected. Two instrumented players sustained diagnosed concussions during data collection.

## Concussive Impacts

Players under 14 years old sustained 6 concussions. A total of 4 concussions were sustained by instrumented players in the 9 to 12 year old group. Concussion 1 was associated with an impact to the front of the helmet that resulted in head accelerations of 58 g and 4548 rad/s<sup>2</sup>. Concussion 2 was associated with an impact to the back of the helmet that resulted in head accelerations of 64 g and 2830 rad/s<sup>2</sup>. Concussion 3 was associated with an impact to the side of the helmet that resulted in head accelerations of 26 g and 1552 rad/s<sup>2</sup>. Concussion 4 was not recorded due to battery failure on the day of concussion.

An additional two concussions were sustained by instrumented players in the 12 to 14 year old group. Concussion 5 was associated with an impact to the top of the helmet that resulted in head accelerations of 87 g and 377 rad/s<sup>2</sup>. Concussion 6 was associated with an impact to the front of the helmet that resulted in head accelerations of 67 g and 3373 rad/s<sup>2</sup>.

## CONCLUSIONS

These data represent some of the first head impact data collected directly from pediatric human subjects at potentially injurious severities. A total of 8 concussions were sustained by instrumented players, including 6 concussions from subjects under the age of 14 years. The impacts associated with concussion were among the highest magnitude impacts experienced by concussed players for the entire season. These data provide valuable insight to the tolerance to head impact of children. With more years of data collection, a comprehensive dataset that biomechanically characterizes pediatric concussion as a function of age can be compiled. This will allow pediatric-specific risk analyses and injury prevention interventions in the automotive and sports environments.

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