Published in final edited form as:
Proc Inst Mech Eng H. 2020 December ; 234(12): 1472–1483. doi:10.1177/0954411920947850.

Concussion and the severity of head impacts in mixed martial arts

Stephen Tiernan[1], **Aidan Meagher**[1], **David O'Sullivan**[2], **Eoin O'Keeffe**[3], **Eoin Kelly**[4], **Eugene Wallace**[4], **Colin P Doherty**[4,5], **Matthew Campbell**[3], **Yuzhe Liu**[6], **August G Domel**[6]

[1]Technological University Dublin, Dublin, Ireland

[2]Smurfit Institute of Genetics, Trinity College Dublin, Dublin, Ireland

[3]Division of Sports Science, Pusan National University, Busan, Republic of Korea

[4]Department of Neurology, Health Care Centre, Hospital 5, St James's Hospital, Dublin, Ireland

[5]Academic Unit of Neurology, Biomedical Sciences Institute, Trinity College Dublin, Dublin, Ireland

[6]Department of Bioengineering, Stanford University, Stanford, CA, USA

Abstract

Concern about the consequences of head impacts in US football has motivated researchers to investigate and develop instrumentation to measure the severity of these impacts. However, the severity of head impacts in unhelmeted sports is largely unknown as miniaturised sensor technology has only recently made it possible to measure these impacts in vivo. The objective of this study was to measure the linear and angular head accelerations in impacts in mixed martial arts, and correlate these with concussive injuries. Thirteen mixed martial arts fighters were fitted with the Stanford instrumented mouthguard (MiG2.0) participated in this study. The mouthguard recorded linear acceleration and angular velocity in 6 degrees of freedom. Angular acceleration was calculated by differentiation. All events were video recorded, time stamped and reported impacts confirmed. A total of 451 verified head impacts above 10g were recorded during 19 sparring events (n = 298) and 11 competitive events (n = 153). The average resultant linear acceleration was 38.0624.3g while the average resultant angular acceleration was 256761739 rad/s2. The competitive bouts resulted in five concussions being diagnosed by a medical doctor. The average resultant acceleration (of the impact with the highest angular acceleration) in these bouts was 86.7618.7g and 756163438 rad/s2. The average maximum Head Impact Power was 20.6kW in the case of concussion and 7.15kW for the uninjured athletes. In conclusion, the study recorded novel data for sub-concussive and concussive impacts. Events that resulted in a concussion had an average maximum angular acceleration that was 24.7% higher and an average maximum Head Impact Power that was 189% higher than events where there was no injury. The findings are significant in understanding the human tolerance to short-duration, high linear and angular accelerations.

Corresponding author: David O'Sullivan, Division of Sports Science, Pusan National University, Busan 46241, Republic of Korea. davidosullivan@pusan.ac.kr.

Declaration of conflicting interests
The author(s) declared no potential conflicts of interest with respect to the research, authorship and/or publication of this article.

Keywords

Concussion; mixed martial arts; mild traumatic brain injury; linear acceleration; angular acceleration

Introduction

Concussion, or mild traumatic brain injury (mTBI), is very prevalent in sport with between 1.6 and 3.8 million sports-related concussions in the United States each year.[1] The diagnosis of concussion is particularly difficult with many studies reporting that approximately 50% of concussions go unreported.[2] Generally, concussion can be classified as an injury to the brain resulting from blunt trauma or acceleration/deceleration of the head and neck, with one or more of the following symptoms attributable to the head injury during the post-traumatic surveillance period: transient confusion, dysfunction of memory, headache, dizziness, irritability, fatigue or poor concentration.[3] Typically, contusions are associated with linear accelerations while diffuse axonal injuries are associated with angular accelerations.[4] The role of predisposing factors in determining an individual's susceptibility to concussion – such as age, sex, concussion history and genetic characteristics – is still unknown.[5]

Historically, efforts have focused on using linear acceleration to indicate head injury; this has changed in the last decade to include angular velocity and angular acceleration. It has been possible to capture this acceleration data in US football, due to the space available within the helmet to mount sensors. The Head Impact Telemetry System (HITS) (Simbex, Lebanon, NH, USA) was developed in 2002 and uses an array of accelerometers to measure linear and angular accelerations. A validation study, using a medium-sized US football helmet on a Hybrid III anthropomorphic dummy head, found that HITS overestimated peak linear acceleration (PLA) by 0.9% and underestimated peak angular acceleration (PAA) by 6.1%.[6] However, Jadischke et al. determined that a large helmet should be used with a Hybrid III head-form and this increased the angular acceleration error. In Jadischke's study, 85.7% of impacts with the large helmet had an angular acceleration error greater than 15%.[7] A study of the HITS data by Rowson et al.[8] found that concussive impacts had an average PLA of $104 \pm 30g$ and PAA of 4726 ± 1931 rad/s^2; the data included 300,977 impacts and 57 concussions. Despite the possible errors in HITS, it provides the only large head impact acceleration data set. Head accelerations that have resulted in a concussion have been investigated by many authors using different techniques. Most of the studies in Table 1 agree on an approximate threshold for concussion of $100g$ for linear acceleration but the reported angular acceleration threshold varies from 4300 to 7229 rad/s^2; this may in part be due to the techniques used to determine angular acceleration.

The alternative to measuring head accelerations in vivo is to recreate the impact using video data in a laboratory or using kinematic software. Several video angles are required for this to be successful and it is a difficult, time-consuming and error prone task.[20] A study by McIntosh et al.[10] of unhelmeted impacts in Australian football used video data to reconstruct 40 head impacts (13 uninjured and 27 concussion cases). The mean peak linear and angular acceleration for concussive injuries were $103.4g$ and 7951 rad/s^2 and for no injury were

59g and 4300 rad/s^2. McIntosh's study also found that 60% of concussive cases had a greater proportion of impacts to the temporal area of the head than non-concussive. The study concluded that there was a 75% probability of a concussion from a PAA of 2296 rad/s^2 in the coronal plane, and a 75% probability of concussion from a resultant PLA in excess of 88.5g. However, a brain simulation study by Zhang et al. used brain tissue criteria to determine that 66g and 4600 rad/s^2, 82g and 5900 rad/s^2 and 106g and 7900 rad/s^2 corresponded to a 25%, 50% and 80% probability of concussion.[21]

The relationship between impact severity and duration has been investigated since John Strapp's work in the 1940s. In 1964, Gurdjian and Roberts[22] published the Wayne State Tolerance Curve (WSTC) shown in Figure 1. This curve was developed from a variety of experiments on animals, cadavers and human volunteers. In 1975, the US National Highway Safety Administration adopted the Head Injury Criteria (HIC), which is a formula to fit the WSTC. HIC values in excess of 1000 are used to predict moderate or serious injury with probable concussion with or without skull fracture.[23]

The HIC and WSTC incorporate linear acceleration and duration but do not include angular acceleration. In 2000, Newman and Shewchenko[24] proposed a new kinematic criterion termed the Head Impact Power (HIP), shown in equation (1). This criterion has the advantage of incorporating linear and angular acceleration and duration

$$\mathrm{HIP} = \left[m a_x \int a_x dt + m a_y \int a_y dt + m a_z \int a_z dt + I_{xx} \propto_x \int \propto_x dt + I_{yy} \propto_y \int \propto_y dt + I_{zz} \propto_z \int \propto_z t \right] \tag{1}$$

where m is the mass of the head, and I_{xx}, I_{yy} and I_{zz} are the moments of inertia of the head around the x-, y- and z-axes, respectively.

Newman and Shewchenko[24] recreated 12 US football impacts, which included 24 players and nine concussions. They determined that there was a 5%, 50% and 95% probability of concussion from HIP values of 4.7, 12.79 and 20.88 kW, respectively.[24]

Following experimental work with animals, Adams et al.[25] proposed that concussion is primarily due to angular accelerations. This has been corroborated by statistical studies of the HITS database.[2,8] A simulation study by Post et al.[26] found that linear acceleration primarily affected the brain's strain response for short-duration events (<15 ms) but as the duration increases, angular acceleration becomes the dominant contributor to brain strain. It is difficult to determine a threshold for PAA as it depends on impact location, direction and duration.[20,27] A kinematic study by Hoshizaki et al.[28] found that the risk of head injury was a function of both the magnitude and duration of an impact; the study determined that a PAA of 5 krad/s^2 over 25 ms had a similar risk of injury as a PAA of 50 krad/s^2 impact over 2 ms.

Head impact sensors

Few studies have measured the severity of head impacts in vivo in unhelmeted sports due to the lack of suitable instrumentation. To address this need, a number of new wireless sensors have been developed, including instrumented mouthguards, headbands and skin patches. A skin patch sensor (X-Patch) developed by X2 Biosystems (Seattle, WA, USA) contained three single axis accelerometers and three gyroscopes. Due to the ease of application, the cost and unobtrusive nature of the device, it has been used in studies of women's soccer,[29] rugby[30] and US football unhelmeted practice.[31] These studies did not include any concussions. One rugby study that included six concussions used the X-Patch to determine the number of head impacts that occurred but did not report the magnitude of the head acceleration.[32] The accuracy of the X-Patch has been investigated and the linear acceleration was found to be in reasonable agreement with a reference sensor but the angular acceleration was underestimated by up to 25%.[33] Sim-G is a headband sensor developed by Triax Ltd. A validation study of this sensor found that due to the movement of the sensors relative to the skull, PLA errors could be up to 370%.[34]

Instrumented mouthguards have been in development for approximately 50 years. Most of the early devices were large, protruded from the mouth and were hard wired back to a fixed station. The instrumented mouthguard has been shown to be the most accurate method of measuring head accelerations in vivo.[35] This is primarily due to the degree of coupling between the head and the mouthguard. X2 Biosystems developed an instrumented mouthguard that was used by King et al.[36] in 2013 to measure head impacts in junior rugby but no concussions were recorded in this study. Bartsch and Samorezov[37] at Cleveland Clinic (USA) have also developed an instrumented mouthguard with reported errors of 3% PLA and 17% PAA. There have not yet been any known published studies which have used this device to record concussive injuries. An instrumented mouthguard has also been developed by CAMLab at Stanford University,[38] and this mouthguard is used in the study reported in this article. The accuracy of this mouthguard was investigated by the CAMLab group by fitting it to an anthropomorphic test head fitted with a US football helmet.[38] The helmet was impacted with a spring loaded horizontal impactor. The normalised root mean error was determined to be 9.9% ± 4.4% for linear acceleration and 9.7% ± 7.0% for angular acceleration. The CAMLab study did not include any direct impacts to the mouthguard; hence, this study does not include any such impacts as the mouthguard has not been validated for these. The coupling of the mouthguard to the skull was compared to a skin patch sensor and a head band sensor by Wu et al.[35] Sensor coupling was quantified by measuring the displacement of the sensor relative to an ear canal reference sensor while heading a soccer ball. The mouthguard error was < 1 mm while the skin patch and head band sensors displaced by up to 4 and 13 mm with reference to the ear canal sensor. The group at Stanford have used the mouthguard to measure head impacts in US football; this is the only known in vivo measurement of head accelerations that resulted in a concussive event by a device other than the HITS system.[39] Two concussions were reported, one case involved a loss of consciousness and the other was self-reported. The loss of consciousness injury had a PLA of 106*g* and a PAA of 12,900 rad/s^2, and the duration of the linear resultant acceleration was approximately 35 ms.

Mixed martial arts

Mixed martial arts (MMA) is a competitive, full-contact sport that involves an amalgamation of elements drawn from boxing, wrestling, karate, taekwondo, jujitsu, muay thai, judo and kickboxing.[40] The fighters wear 110–170 g gloves and do not wear head protection.

Competitive bouts consist of three rounds of 5 min each, provided the referee does not stop the fight. Fights are ended by the referee either due to a knockout (KO), where there is a loss of consciousness, or more commonly a technical knockout (TKO), where a fighter appears unable to defend themselves. The fight may also end by submission. A 10-year review of injuries in MMA found that head trauma was the most common reason for match stoppages (28.3%).[41] Another study of 844 MMA fights found that 12.7% of matches were stopped due to KOs and that it took the referee an average of 3.5 s after the KO to stop the match. During this time, the fighter suffered on average an additional 2.6 strikes to the head.[40] This study also found that 19.1% of matches were stopped due to a TKO, in the 30 s preceding the TKO stoppage the fighter suffered an average of 17.1 strikes to the head. A third of MMA fighters have reported suffering a TKO and 15% have suffered from a KO, having participated in MMA for an average of 5.8 years.[42] A 1-year study of 13 MMA fighters found cortical thinning and reduced memory and processing speed when compared to controls (n = 14).[43]

Method

Thirteen adult MMA fighters took part in this study, 12 professional or semi-professional and one amateur. Fighters took part in both sparring and competitive events, as shown in Table 2, and none of the events included two of the participants competing against each other. The fighters were fitted with the Stanford instrumented mouthguard (MiG2.0) and ethical approval was granted by the Institute of Technology Tallaght Ethics Committee REC-STF1-201819. To ensure that the mouthguard was a tight fit, a dental impression was taken and two mouthguards were manufactured for each fighter: one fitted with sensors and a ‘dummy’ one which had the look and feel of the instrumented one. The mouthguards were manufactured by OPRO, England, a leading gum shield manufacturer. The fit of the mouthguards was checked and each fighter was given the ‘dummy’ mouthguard for training; this ensured that the fighters were familiar with the mouthguard and that the instrumented mouthguard would not become worn or damaged.

The instrumented mouthguard used in this study was developed by the CAMLab research group, Stanford University and is shown in Figure 2.

The mouthguard has a triaxial accelerometer to measure linear acceleration and a triaxial gyroscope to measure angular rate. The sensors, processor and battery are completely sealed in three layers of ethylene vinyl acetate in a dental moulded mouthguard. Data are downloaded from the device post event via Bluetooth.[44] In this study, impacts were recorded when linear accelerations exceeded the $10g$ threshold established in previously published studies.[45,46] The acquisition window was 50 ms pre-trigger and 150 ms post-trigger. Linear acceleration and angular velocity were sampled at 1000 Hz, and all data were filtered using a fourth-order Butterworth low-pass filter with a cut-off frequency of 300 Hz.[47] Angular acceleration was estimated using a five-point stencil derivative of the measured angular velocity.[39] The accelerations were transformed to the centre of gravity using equation (2)[36] and the offsets for a 50th percentile human head (−0.07764, 0, 0.07207)[35]

$$\overrightarrow{a_{CG}} = \overrightarrow{a_s} + \overrightarrow{\propto}\left(\overrightarrow{r_s}\right) + \overrightarrow{\boldsymbol{\omega}}\left(\overrightarrow{\boldsymbol{\omega}} x \overrightarrow{r_s}\right) \quad (2)$$

where a_{CG} is the head linear acceleration at the centre of gravity, a_s is the linear acceleration at the mouthguard sensor, α is the head angular acceleration, $\boldsymbol{\omega}$ is the head angular velocity, and r_s is the vector position of the mouthguard sensor to the centre of gravity of the head.

Data capture and analysis

Each event was recorded by two cameras placed at different angles around the arena and the video was recorded at 60 frames per second. In addition, TV coverage was available for the competitive events. The time on the mouthguard data was aligned with the video time line and the video was examined frame by frame by two researchers using the Kinovea video analysis software. The video data were used to confirm that a head impact had occurred and that the direction of the impact conformed to the direction indicated by the mouthguard. To define the impact direction, the head was divided into eight equal transverse sectors as shown in Figure 3.

If an impact could not be confirmed, it was removed from the analysis. In addition, if it was found from the video that an impact was directly to the mouthguard and thus sensors, it was removed as it may produce a sharp spike in the acceleration data. This method was used to remove 47 impacts which were suspected to have been direct impacts to the mouthguard.

Reported linear and angular accelerations are calculated peak resultants. The duration reported for the impacts is the time interval over which the acceleration first exceeded a predetermined threshold, an example of how this calculation was performed is shown in Figure 4. A threshold of 10*g*, as established in other studies,[45,46] was utilised for linear acceleration. A threshold for the calculation of angular acceleration duration is not specified by other researchers; therefore, 500 rad/s^2 was used as this was greater than any spurious data. This approach allowed for a consistent and repeatable method to carry out a comparative analysis of the impact durations.

The MMA athletes were medically examined before the study commenced, immediately after competitive bouts and again approximately 48 h after the competitive events. The medical examinations were conducted by an emergency medicine doctor. Prior to the events, the examination included a physical examination and the recording of the participant’s medical history. After the events, the athletes had a physical examination and were checked for any concussion symptoms such as persistent headaches, visual disturbance and imbalance. If a concussion was suspected, the athlete was examined using the latest version of the Sports Concussion Assessment Tool (SCAT). It should be noted that the concussed fighters received between 4 and 26 head impacts during their bouts. It is not possible to identify which impact caused their injury.

To determine the severity of the impacts, the HIP was computed using the method developed by Newman and Shewchenko[24] as shown in equation (1). The HIP is calculated over the 200 ms capture time of each impact and the maximum value is reported.

To investigate the relationship between peak resultant linear and peak resultant angular acceleration, a linear regression analysis was performed, using Minitab, LLC, for each impact site. Pearson's correlation coefficient and adjusted R^2 values were calculated to determine if a linear relationship existed and if so the strength of that relationship.

Results

During this study, data were recorded during sparring sessions and competitive events. All fighters participated in sparring sessions and nine in competitive events. Above 10*g*, 298 confirmed head impacts were recorded during 19 sparring sessions, resulting in an average of 15.74 head impacts per sparring session. The average PLA for all impacts sustained during the sparring sessions was 32.0 ± 17.2*g* while the PAA was 2149 ± 14,285 rad/s^2. The median accelerations for the sparring sessions had a PLA of 28.4*g* and a PAA of 1701 rad/s^2. No injuries were occurred during the sparring sessions. Figure 4 shows an example of the mouthguard data recorded during a typical sparring head impact.

Eleven competitive events were studied at which 153 confirmed head impacts above 10*g* were recorded, resulting in an average of 13.9 head impacts per event. The median PLA for the competitive events was 36.8*g* and the PAA was 2956 rad/s^2. The average PLA for all impacts sustained during competitive events was 46.5 ± 29.9*g* and the average PAA was 3355 ± 1912 rad/s^2. Five of the competitive events resulted in the fighter sustaining a concussion.

Histograms of the linear and angular accelerations of the impacts from both sparring and competitive events are shown in Figure 5. As expected, these are skewed to the left demonstrating that the majority of impacts are below 50*g* (77.5%) and 4000 rad/s^2 (74.4%).

The impacts with the highest PAA from each competitive bout were selected as angular acceleration has been correlated with concussion,[2,8,25] and these are presented in Table 3. HIP values were calculated for all impacts. The average of the maximum values from each event that resulted in a concussion was 20.6 kW. The maximum HIP value for both sparring and competitive events at which there was no head injury was averaged and found to be 7.15 kW.

Averages and standard deviations were calculated for impacts with the highest PAA recorded at each competitive bout and sparring session, and these are presented in Table 4. The competitive events were divided into injury and no injury categories, and no injury occurred at the sparring sessions.

In competitive events, the average PLA was 30.3% higher, and the PAA was 28.5% higher in the cases of concussion. In concussive cases in competitive events, the average PLA was 69% higher and the PAA was 49.6% higher than the sparring sessions. The impact with the highest PAA in each competitive and sparring event was analysed and the duration of the linear acceleration versus the duration of the angular acceleration of that event is plotted in Figure 6. These high angular acceleration impacts were from impacts with the gloved fist as opposed to body impacts. From this figure, it is clear that the majority of impacts that resulted in a concussion had longer durations than those that did not result in an injury. Four impacts that resulted in a concussion had an angular acceleration duration ⩾ 20 ms. Figure 7 shows a plot of the angular acceleration versus the linear acceleration, the concussed and uninjured fighters are indicated by red diamonds and blue squares, respectively.

The impact direction shown in Figure 8 was determined from the azimuthal and polar angles of the resultant linear acceleration vector and verified by video analysis. It was found that 57.5% of the impacts in sparring and competitive events were to the front of the head (including front, front left and front right 135°) with 33.9% of these being to the front left of the head (45° quadrant).

The relationship between linear and angular acceleration was also investigated as shown in Table 5. Using a statistical regression model, the best fit line was found to be from impacts to the back right of the head ($n = 21$), where R^2 adjusted was 0.70. This was followed by impacts to the front left of the head ($n = 156$), where R^2 adjusted was 0.61. No relationship was evident between the linear and angular accelerations for impacts from other directions as R^2 adjusted was less than 0.6.

Discussion

This study measured head impacts in vivo during 19 sparring MMA sessions and 11 competitive MMA bouts. Five of the fighters were diagnosed with a concussion following their competitive bouts. Four of the concussions were diagnosed immediately after the event while the other one was diagnosed during the 48-h follow-up medical examination. There are few studies where in vivo head accelerations have been measured in unhelmeted sports and the studies that do exist are of non-injurious impacts.[35,36,48] The majority of published head acceleration data have been acquired through laboratory tests, kinematic reconstructions or in-helmet sensors as shown in Table 1.

The impacts experienced by the fighters were complex three-dimensional (3D) waves (Figure 4). The average PLA of the concussive events was 86.7*g* which was lower than Australian rugby at 103.4*g*[10] and US football at 105*g*.[11,19] The average PAA of the concussive events was 7561 rad/s^2 which was similar to Australian football at 7951 rad/s^2[10] and higher than the range published for US football, 4726 rad/s^2[8] to 7230 rad/s^2.[19] In a study by Viano et al.[49] where boxers punched a Hybrid III head (including neck and upper torso), the PAA ranged from 3181 to 9306 rad/s^2 depending on the type of punch. Punches in this study had a lower PLA than US football but higher PAA due to the higher moment applied to the head, this is in agreement with other studies that analysed punches to the head.[49,50]

The prediction of concussion based on acceleration alone is not reliable,[11] and impact duration is also an important factor.[26,51] Punches in MMA (using light gloves and no head protection) result in large angular accelerations of short duration. The risk of brain injury is dependent on both the magnitude of the accelerations and their duration, as demonstrated by the WSTC.[52] This study found that the impact durations were considerably longer in the cases of concussion. PLA and PAA duration was on average 87.0% and 72.1% longer in the cases of concussion when compared to the uninjured fighters. Deck and Willinger[51] reported the duration of linear acceleration in pedestrian head impacts to be 9–14.5 ms, these impacts were unhelmeted and the duration is comparable to the average of 13.4 ms found in this study. This contrasts with longer linear acceleration durations in US football due to the level of padding in the helmets and the body armour worn by the players.[53] The duration of an impact is dependent on a number of factors including the compliance of the impact surface, the impact direction and the energy absorption of any head protection, if worn. In a concussive impact recorded by the Stanford mouthguard in US football, the duration was approximately 35 ms,[39] which is considerably longer than the 17.4 ms average duration of the concussive impacts in this study, demonstrating that the dynamics of unhelmeted and helmeted impacts are different.

Four of the mTBI's reported in this study were to fighters who sustained impacts with a linear acceleration duration $\geqslant$ 11 ms and an angular acceleration duration $\geqslant$ 20 ms (Table 2 and Figure 6). This study suggests that repeated impacts within a short time are also a factor in concussion; Fighter 3 in Bout 1 was concussed and sustained four short-duration impacts within 3 s, the most severe of which had a PLA of 94.2g and a PAA of 4100 rad/s^2. This is different to the normal second impact syndrome reported in sport in which repeated impacts may be over days or weeks.[54] This study recorded some interesting data on subconcussive impacts. There were eight non-injurious impacts recorded in competitive and sparring events whose PAA exceeded 6000 rad/s^2, and it is hypothesised that this is due to the duration of the angular acceleration being less than 20 ms.

Some researchers have proposed that linear and angular acceleration are correlated,[8,16] but this was not apparent in this study. Impacts to the front and back right had the highest R^2 adjusted value (0.46 and 0.70) and Pearson's correlation coefficient (0.69 and 0.85, respectively). It is thought that the relationship between PLA and PAA depends on the style of fighting of both opponents.

The impact with the highest HIP value did not correspond to the impact with the maximum angular acceleration in three of the five cases of concussion. The average HIP value for the concussed cases was 20.6 and that for the uninjured fighters was 7.15. Newman et al. determined that a HIP of 12.79 and 20.88 corresponded to a probability of concussion of 50% and 95%, respectively. Thus, the average HIP value of concussed athletes in this study compares with the 95% probability determined by Newman et al.[18] Marjoux et al.[55] determined a much higher HIP value of 24 kW for a 50% risk of injury but their study included severe neurological injuries and head fractures.

The direction of the impact was investigated and found that 54.9% of sparring impacts and 36.4% of impacts in competition were sustained to the front left section of the head. This is consistent with the majority of fighters being right-handed and hook style or jaw punches. Four of the five impacts that resulted in concussion were to the front or front left of the head. Impact location is significant as impacts in the temporal region have been shown to be more likely to result in a concussion.[56,20]

Conclusion

This study measured head linear and angular accelerations in vivo in MMA during both sparring sessions and competitive bouts. It is one of very few studies to record in vivo concussive and subconcussive impacts in an unhelmeted sport. No injuries resulted from the sparring sessions despite three impacts which had PAAs in excess of 6000 rad/s^2. The 11 competitive bouts studied resulted in five concussions being diagnosed either immediately after the event or in a 48-h checkup. The average PAA differed by 24.7%, between concussed and uninjured fighters, but the duration of the linear and angular acceleration was considerable longer in the cases of concussion: 87% for linear acceleration and 52.5% for angular acceleration.

The impacts in MMA are of a shorter duration than those experienced in US football due to the light gloves worn by the fighters and the lack of head protection gear. The human tolerance to repeated relatively short severe impacts is unknown, but the data in this study are important to help understand the magnitude and variation of impacts that can cause an injury.

Limitations

The number of fighters and events in this study was limited; a greater number of impacts are required to improve the robustness of these findings, as well as further validation of the mouthguard in MMA style impacts such as direct strikes. The duration was calculated based on 10g and 500 rad/s^2 thresholds; this is not directly comparable to other studies as they have not specified their methods of calculation. Impacts that could not be video verified and also impacts that appeared to be direct hits to the mouthguard were removed; this may have resulted in some valid data not being included. The concussed fighters received multiple impacts during their bouts; therefore, it is not possible to identify which impact caused the injury.

Funding

The author(s) disclosed receipt of the following financial support for the research, authorship and/or publication of this article: This work was, in part, supported by the Financial Supporting Project of Long-term Overseas Dispatch of PNU's Tenure-track Faculty.

References

1. Langlois JA, Rutland-Brown W and Wald MM. The epidemiology and impact of traumatic brain injury: a brief overview. J Head Trauma Rehabil 2006; 21: 375–378. [PubMed: 16983222]
2. Elliott MR, Margulies SS, Maltese MR, et al. Accounting for sampling variability, injury under-reporting, and sensor error in concussion injury risk curves. J Biomech 2015; 48: 3059–3065. [PubMed: 26296855]
3. McCrory P, Meeuwisse W, Dvorak J, et al. Consensus statement on concussion in sport: the 5th international conference on concussion in sport held in Berlin, October 2016. Br J Sports Med 2017; 51: 699.
4. Hoshizaki B, Andrew P, Marshall K, et al. The relationship between head impact characteristics and brain trauma. J Neurol Neurophysiol 2014; 5: 1–8.
5. Mcgrew CA. Sports-related concussion V genetic factors. Am Coll Sport Med 2019; 18: 20–22.

6. Beckwith JG, Greenwald RM and Chu JJ. Measuring head kinematics in football: correlation between the head impact telemetry system and hybrid III headform. Ann Biomed Eng 2012; 40: 237–248. [PubMed: 21994068]
7. Jadischke R, Viano D, Dau N, et al. On the accuracy of the head impact telemetry (hit) system used in football helmets. J Biomech 2013; 46: 2310–2315. [PubMed: 23891566]
8. Rowson S, Duma SM, Beckwith JG, et al. Rotational head kinematics in football impacts: an injury risk function for concussion. Ann Biomed Eng 2012; 40: 1–13. [PubMed: 22012081]
9. Wilcox BJ, Beckwith JG, Greenwald RM, et al. Biomechanics of head impacts associated with diagnosed concussion in female collegiate ice hockey players. J Biomech 2015; 48: 2201–2204. [PubMed: 25913243]
10. McIntosh AS, Patton DA, Fréchède B, et al. The biomechanics of concussion in unhelmeted football players in Australia: a case–control study. BMJ Open 2014; 4: e005078.
11. Rowson S and Duma SM. Brain injury prediction: assessing the combined probability of concussion using linear and rotational head acceleration. Ann Biomed Eng 2013; 41: 873–882. [PubMed: 23299827]
12. Reed N, Taha T, Keightley M, et al. Measurement of head impacts in youth ice hockey players. Int J Sports Med 2010; 31: 826–833. [PubMed: 20830655]
13. Stojsih S, Boitano M, Wilhelm M, et al. A prospective study of punch biomechanics and cognitive function for amateur boxers. Br J Sports Med 2010; 44: 725–730. [PubMed: 19019907]
14. Guskiewicz KM, Mihalik JP, Shankar V, et al. Measurement of head impacts in collegiate football players. Neurosurgery 2007; 61: 1244–1253. [PubMed: 18162904]
15. Viano D, Casson I, Pellman E, et al. Concussion in professional football: brain responses by finite element analysis: Part 9. Neurosurgery 2005; 57: 891–915. [PubMed: 16284560]
16. Pellman E, Viano D, Tucker AM, et al. Concussion in professional football: reconstruction of game impacts and injuries. Neurosurgery 2003; 53: 796. [PubMed: 14560730]
17. Duma SM, Manoogian SJ, Bussone WR, et al. Analysis of real-time head accelerations in collegiate football players. Clin J Sport Med 2005; 15: 3–8. [PubMed: 15654184]
18. Newman J, Barr C, Beusenberg M, et al. A new biomechanical assessment of mild traumatic brain injury, part 2: results and conclusions. In: International IRCOBI conference on the biomechanics of impact, Montpellier, 20–22 September 2000, pp.223–233. International Research Council on the Biomechanics of Impacts.
19. Broglio SP, Schnebel B and Sosnoff J. The biomechanical properties of concussions in high school football. Med Sci Sports Exerc 2010; 42: 2064–2071. [PubMed: 20351593]
20. Pellman E, Viano D, Tucker AM, et al. Concussion in professional football: location and direction of helmet impacts – Part 2. Neurosurgery 2003; 53: 1328–1341. [PubMed: 14633299]
21. Zhang L, Yang K and King A. A proposed injury threshold for mild traumatic brain injury. J Biomech Eng 2004; 126: 226–236. [PubMed: 15179853]
22. Gurdjian E and Roberts V. Tolerance curves of acceleration and intracranial pressure and protective index in experimental head injury. J Trauma 1966; 6: 600–604. [PubMed: 5928630]
23. Shojaati M. Correlation between injury risk and impact severity index ASI. In: Proceedings of the 3rd Swiss transport research conference, Monte Verita, 19–21 March 2003, http://nan.brrc.be/docs_public/other/shojaati.pdf
24. Newman JA and Shewchenko N. A proposed new biomechanical head injury assessment function – the maximum power index. SAE technical paper 2000-01-SC16, 2000.
25. Adams JH, Graham DI and Gennarellie TA. Head injury in man experimental animals: neuropathology. Trauma Regener 1983; 32: 15–30.
26. Post A, Blaine Hoshizaki T, Gilchrist MD, et al. Peak linear and rotational acceleration magnitude and duration effects on maximum principal strain in the corpus callosum for sport impacts. J Biomech 2017; 61: 183–192. [PubMed: 28807524]
27. Kleiven S. Influence of impact direction on the human head in prediction of subdural hematoma. J Neurotrauma 2003; 20: 365–379. [PubMed: 12866816]
28. Hoshizaki TB, Post A, Kendall M, et al. The development of a threshold curve for the understanding of concussion in sport, Trauma (United Kingdom) 2017; 19: 196–206.

29. McCuen E, Svaldi D, Breedlove K, et al. Collegiate women's soccer players suffer greater cumulative head impacts than their high school counterparts. J Biomech 2015; 48:3720–3723. [PubMed: 26329462]
30. King D, Hume P, Gissane C, et al. Head impacts in a junior rugby league team measured with a wireless head impact sensor: an exploratory analysis. J Neurosurg Pediatr 2017; 19: 13–23. [PubMed: 27791705]
31. Swartz EE, Broglio SP, Cook SB, et al. Early results of a helmetless-tackling intervention to decrease head impacts in football players. J Athl Train 2015; 50: 1219–1222. [PubMed: 26651278]
32. Carey L, Stanwell P, Terry DP, et al. Verifying head impacts recorded by a wearable sensor using video footage in rugby league: a preliminary study. Sport Med Open 2019; 5: 9.
33. Nevins D, Smith L and Kensrud J. Laboratory evaluation of wireless head impact sensor. Procedia Eng 2015; 112: 175–179.
34. Siegmund GP, Bonin SJ, Luck JF, et al. Validation of a skin-mounted sensor for measuring in-vivo head impacts. Int Res Counc Biomech Inj 2015; 28: 182–183.
35. Wu LC, Nangia V, Bui K, et al. In vivo evaluation of wearable head impact sensors. Ann Biomed Eng 2016; 44: 1234–1245. [PubMed: 26289941]
36. King D, Hume P, Gissane C, et al. Head impacts in a junior rugby league team measured with wireless sensor X patch. JNS Pediatr 2017; 19: 13–23.
37. Bartsch A and Samorezov S. Cleveland clinic intelligent mouthguard: a new technology to accurately measure head impact in athletes and soldiers. Defense Secur Sens 2013; 2013:8723.
38. Garza D, Camarillo D, Shultz R, et al. An instrumented mouthguard for measuring linear and angular head impact kinematics in American football. Ann Biomed Eng 2013; 41: 1939–1949. [PubMed: 23604848]
39. Hernandez F, Wu L, Yip MC, et al. Six degree of freedom measurements of human mild traumatic brain injury. Ann Biomed Eng 2015; 43: 1918–1934. [PubMed: 25533767]
40. Hutchison MG, Lawrence DW, Cusimano MD, et al. Head trauma in mixed martial arts. Am J Sports Med 2014; 42: 1352–1358. [PubMed: 24658345]
41. Buse GJ. No holds barred sport fighting: a 10 year review of mixed martial arts competition. Br J Sports Med 2006; 40: 169–172. [PubMed: 16432006]
42. Taylor P, Heath CJ and Callahan JL. Self-reported concussion symptoms and training routines in mixed martial arts athletes. Res Sport Med 2013; 21: 195–203.
43. Mayer AR, Ling J and Dodd AB. Gas. A longitudinal assessment of structural and chemical alterations in mixed martial arts fighters. J Neurotrauma 2012; 32: 3833.
44. Kuo C, Wu L, Hammoor BT, et al. Effect of the mandible on mouthguard measurements of head kinematics. J Biomech 2016; 49: 1845–1853. [PubMed: 27155744]
45. King D, Hume P, Gissane C, et al. The influence of head impact threshold for reporting data in contact and collision sports: systematic review and original data analysis. Sport Med 2015; 46: 151–169.
46. King D, Hume PA, Brughelli M, et al. Instrumented mouthguard acceleration analyses for head impacts in amateur rugby union players over a season of matches. Am J Sports Med 2014; 43: 614–624. [PubMed: 25535096]
47. Wu L, Laksari K, Kuo C, et al. Bandwidth and sample rate requirements for wearable head impact sensors. J Biomech 2016; 49: 2918–2924. [PubMed: 27497499]
48. Paris AJ, Antonini KR and Brock JM. Accelerations of the head during soccer ball heading. In: ASME 2010 summer bioengineering conference, Napels, FL, 16–19 June 2010, pp. 815–816. New York: ASME.
49. Viano D, Casson I, Pellman E, et al. Concussion in professional football: comparison with boxing head impacts – part 10. Neurosurgery 2005; 57: 1154–1170. [PubMed: 16331164]
50. Kendall Marshall Post Andrew GM. A comparison of dynamic impact response and brain deformation metrics within the cerebrum of head impact reconstructions representing three mechanisms of head injury in ice hockey. In: International research council on the biomechanics of impact (IRCOBI) conference, Dublin, 12–14 September 2012. International Research Council on the Biomechanics of Injury.

51. Deck C and Willinger R. Improved head injury criteria based on head FE model. Int J Crashworthiness 2008; 13: 667–678.
52. Anderson R and McLean J. Biomechanics of closed head injury. In: Reilly P and Bullock R (eds) Head injury. London: Chapman & Hall, 2005, pp.26–40.
53. Zhao W and Ji S. Brain strain uncertainty due to shape variation in and simplification of head angular velocity profiles. Biomech Model Mechanobiol 2017; 16: 449–461. [PubMed: 27644441]
54. McLendon LA, Kralik SF, Grayson PA, et al. The controversial second impact syndrome: a review of the literature. Pediatr Neurol 2016; 62: 9–17. [PubMed: 27421756]
55. Marjoux D, Baumgartner D, Deck C, et al. Head injury prediction capability of the HIC, HIP, SIMon and ULP criteria. Accid Anal Prev 2008; 40: 1135–1148. [PubMed: 18460382]
56. Oeur A, Hoshizaki BT and Gilchrist MD. The influence of impact angle on the dynamic response of a hybrid III headform and brain tissue deformation. Mech Concussion Sport 2014; 2014: 56–69.

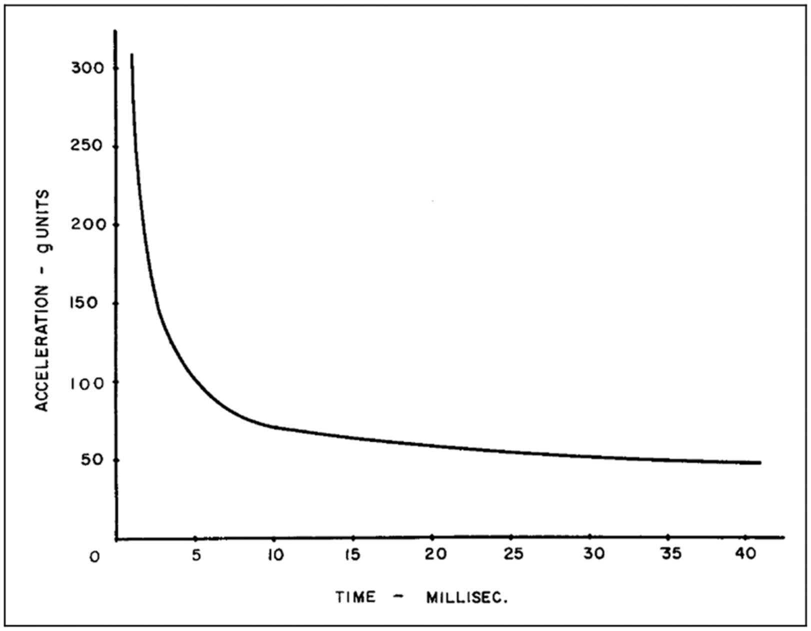

Figure 1.
Wayne State Tolerance Curve.[22]

Figure 2.
CAMLab instrumented mouthguard.

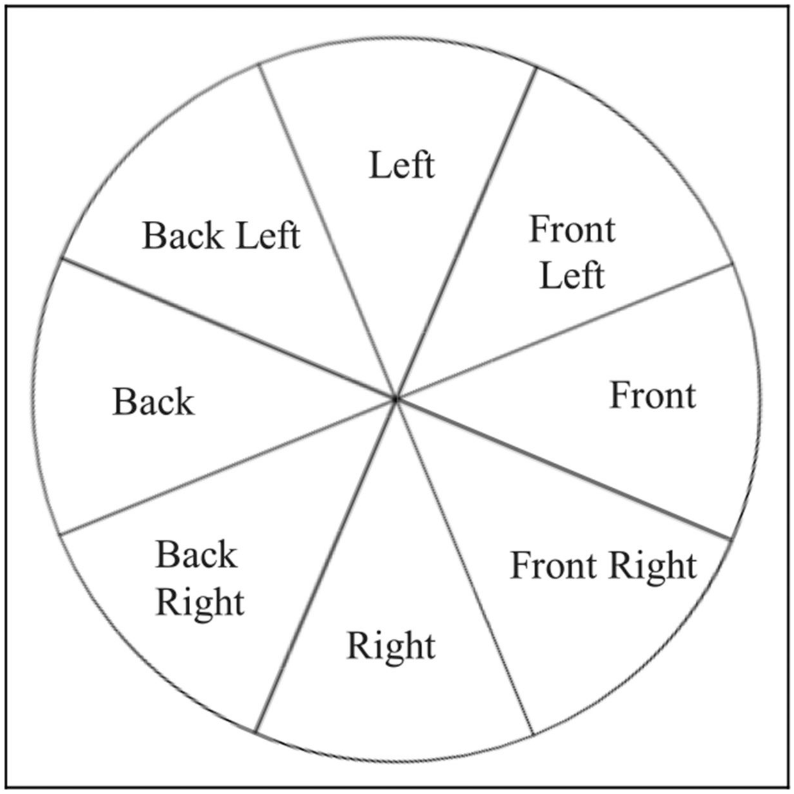

Figure 3.
Impact direction sectors.

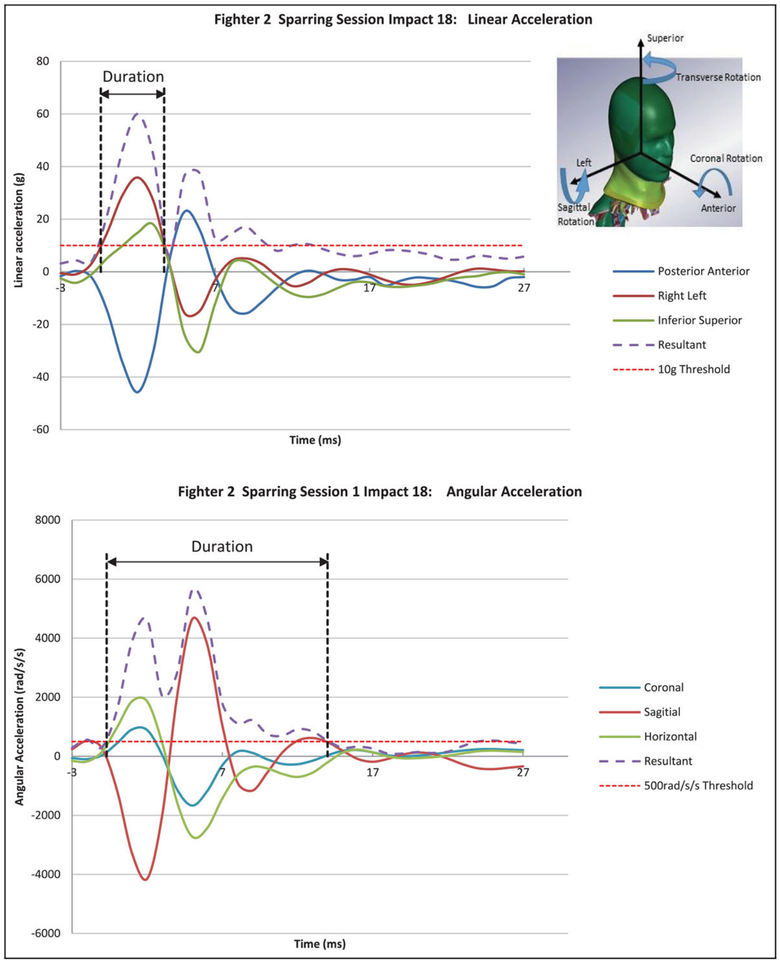

Figure 4.

Sample data from sparring session.

3 ms pre-trigger and 27 ms post-trigger are displayed.

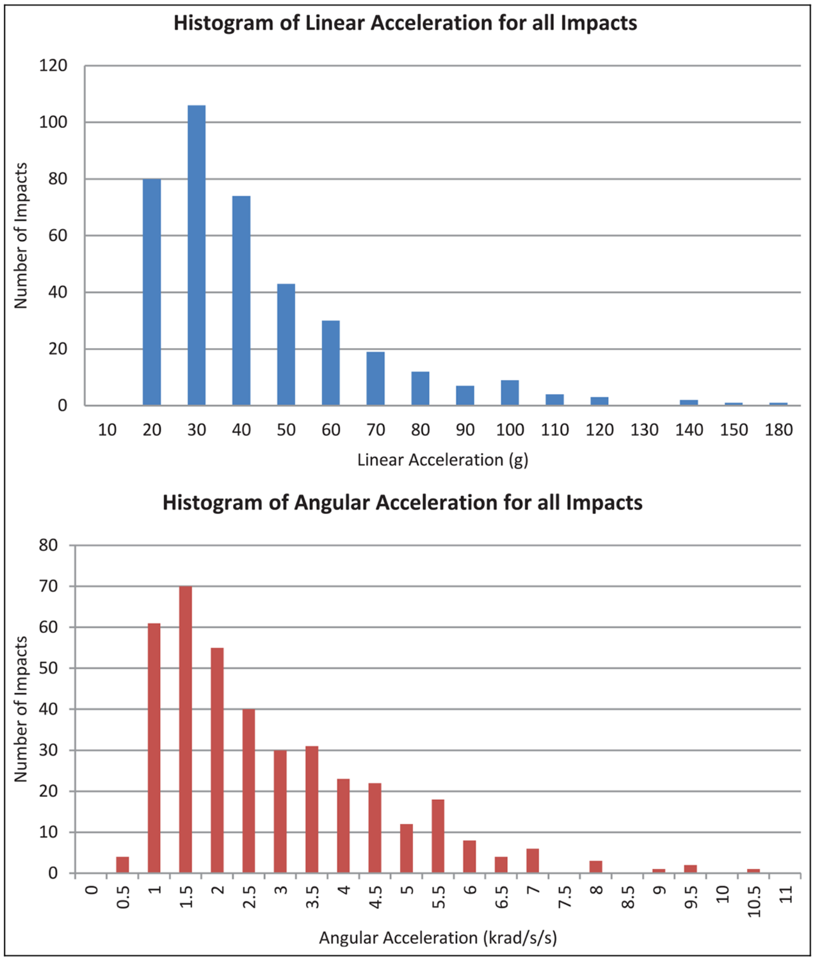

Figure 5.
Number and severity of linear and angular accelerations during competitive events.

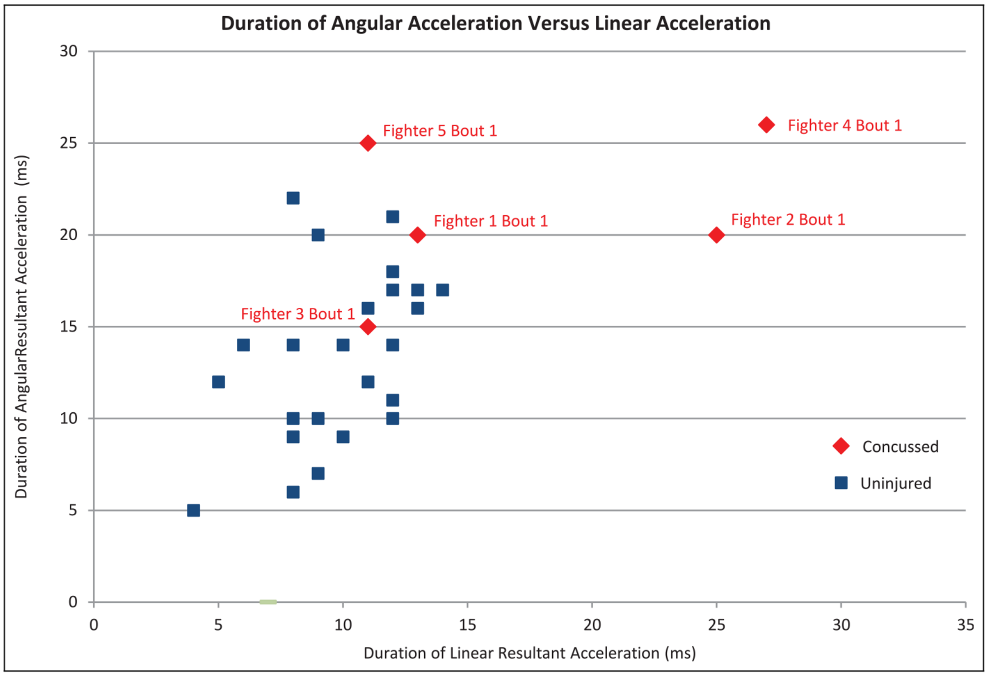

Figure 6.
Duration of linear and angular resultant accelerations of the most severe impact (highest angular acceleration) recorded at each competitive event and sparring session.

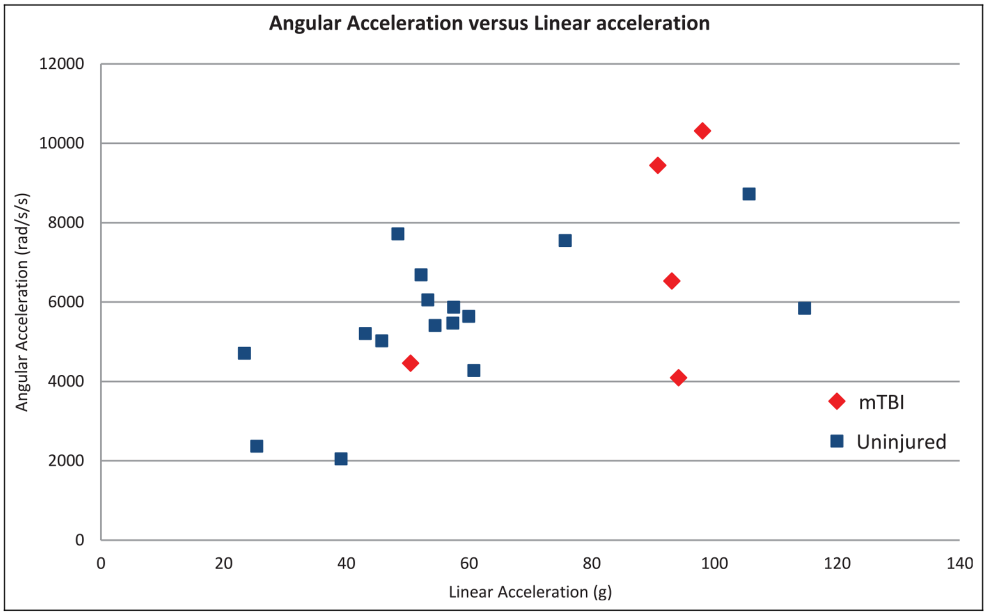

Figure 7.
Angular acceleration versus linear acceleration of the most severe impact (highest angular acceleration) recorded at each competitive event and sparring session.

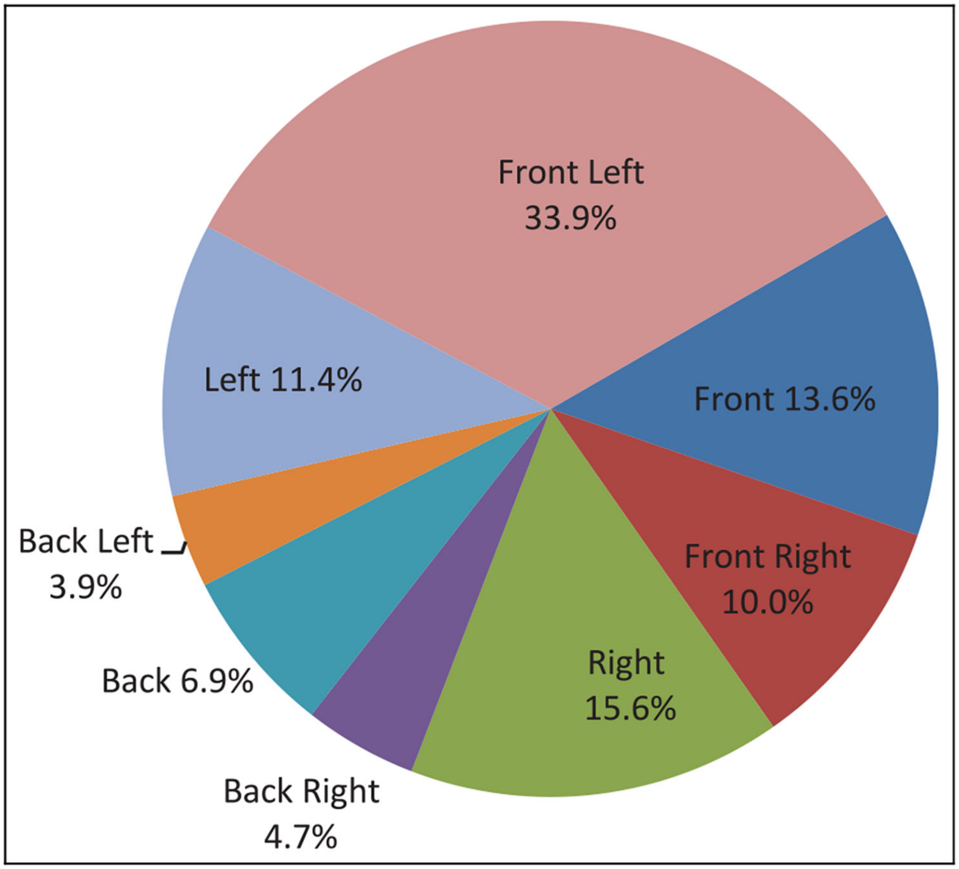

Figure 8.
Percentages of impacts in different directions from both sparring sessions and competition.

Table 1.

Published linear and angular accelerations thresholds for concussion.

Author	Linear acceleration		Angular acceleration		Number of impact cases		Sport	Method	Year
	No injury	Concussion	No injury (rad/s^2)	Concussion (rad/s^2)	No Injury	Concussion			
Wilcox et al.[9]		43g (11.5g)		4029.5 (1243)	58	4	Ice hockey	HITS	2015
Mcintosh et al.[10]		103.45g	4300 (3657)	7951 (3562)	40	13	Australian football	Kinematic simulation	2014
Rowson and Duma[11]	26g (19g)	104g (30g)	1072 (850)	4726 (1931)	62,974	37	US football	HITS	2013
Rowson et al.[8]			1230 (915)	5022 (1791)	300,977	57	US football	HITS	2011
Reed et al.[12]	22.1g		1557.4		1821	0	Ice hockey	HITS	2010
Stojsih et al.[13]	191g (PLA)		17,156 (PAA)		60	0	Boxing	Modified HITS	2010
Guskiewicz et al.[14]		114.6g (54.1g)		5312	88	13	US football	HITS	2007
Pellman et al.[15,16]	60g (24g)	98g (28g)	4029 (1438)	6432 (1813)	182	31	US football	Labre-construction	2007
Duma et al.[17]	32g (25g)	81g	2022 (2042)	5595	3311	1	US football	HITS	2005
Newman et al.[18]	54.3g	97.9g	4159	6664	33	25	US football	Labre-construction	2000
Broglio et al.[19]	25.1g	105g	1626	7229.5	54,247	13	US football	HITS	2010

HITS: Head Impact Telemetry System; PLA: peak linear acceleration; PAA: peak angular acceleration.

Note: Standard deviation is shown in brackets.

Table 2.

Study participants.

	Number of events		Weight class	Maximum weight	Gender	Level
	Sparring	Competition				
Fighter 1		1	Lightweight	70.3 kg	Male	Pro
Fighter 2	2	1	Lightweight	70.3 kg	Male	Pro
Fighter 3	3	2	Middleweight	83.9 kg	Male	Pro
Fighter 4	1	2	Lightweight	70.3 kg	Male	Pro
Fighter 5	4	1	Lightweight	70.3 kg	Male	Pro
Fighter 6	1	1	Flyweight	56.7 kg	Female	Pro
Fighter 7	1		Strawweight	52.2 kg	Female	Pro
Fighter 8	1		Bantamweight	61.2 kg	Male	Amateur
Fighter 9	3	1	Featherweight	65.9 kg	Male	Pro
Fighter 10	1	1	Bantamweight	61.2 kg	Male	Semi-Pro
Fighter 11	1		Lightweight	70.3 kg	Male	Semi-Pro
Fighter 12		1	Strawweight	52.2 kg	Female	Pro
Fighter 13	1		Welterweight	77 kg	Male	Semi-Pro

Table 3.

Head impacts from each competitive event with the highest resultant angular acceleration.

Fighter number	Event number	Impact number	Linear acceleration (g)	Duration (> 10g) (ms)	Angular acceleration (rad/s^2)	Duration (> 500 rad/s^2) (ms)	HIP (kW)	Direction	Diagnosis
Fighter 1	Bout 1	71	50.5	13	4458	20	7.33	FL	Concussion
Fighter 2	Bout 1	9	93.1	25	6527	20	11.46	R	Concussion
Fighter 3	Bout 1	56	94.2	11	4090	15	9.29	FL	Concussion
Fighter 3	Bout 2	8	105.7	12	8722	17	4.8	F	Uninjured
Fighter 4	Bout 1	53	90.8	27	9439	26	8.43	F	concussion
Fighter 4	Bout 2	5	54.5	8	5407	22	6.44	R	Uninjured
Fighter 5	Bout 1	21	104.9	11	13,290	25	18.91	F	concussion
Fighter 6	Bout 1	47	57.5	9	5870	7	3.02	FL	Uninjured
Fighter 9	Bout 1	4	60.4	8	8524	14	7.22	L	Uninjured
Fighter 10	Bout 1	50	75.7	5	7543	12	2.83	FR	Uninjured
Fighter 12	Bout 1	8	45.5	9	6351	20	2.72	FL	Uninjured

HITS: Head Impact Telemetry System; F: front; L: left; FL: front left; FR: front right.

Table 4.

Averages and standard deviations of the impacts with the highest angular acceleration at competitive events and sparring sessions.

Event	Injury	Parameter	Linear acceleration (*g*)	Duration (10*g*) (ms)	Angular acceleration (rad/s^2)	Duration (500 rad/s^2) (ms)	HIP (kW)
Competition	Concussion	Average	86.7	19.0	7560.8	20.0	11.1
		Standard deviation	(18.7)	(6.5)	(3437)	(4.3)	(4.1)
Competition	No Injury	Average	66.6	9.2	7069.5	12.3	4.5
		Standard deviation	(19.7)	(2.1)	(1277)	(4.9)	(1.8)
Sparring	No injury	Average	51.3	10.1	5055.7	12.5	5.0
		Standard deviation	(18.3)	(2.6)	(1374)	(4.20)	(3.1)

Note: Standard deviation is shown in brackets.

Table 5.

Correlation between linear and angular acceleration for each impact direction.

Direction	Number of impacts	R^2 adjusted	Pearson correlation
Front	63	38.5%	0.623
Front right	46	8.0%	0.302
Front left	156	61.4%	0.784
Left	53	17.6%	0.444
Right	72	51.9%	0.726
Back right	21	69.7%	0.846
Back left	18	23.3%	0.541
Back	32	59.4%	0.651