

Concussion and the Severity of Head Impacts in Mixed Martial Arts

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Abstract

Background: Concern about the consequences of head impacts in US football 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*.

Aim: The objective of this study was to measure the linear and angular head accelerations in impacts in mixed martial arts (MMA), and correlate these with concussive injuries.

Methods: Thirteen MMA fighters were fitted with the Stanford instrumented mouthguard (MiG2.0). The mouthguard records 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.

Results: 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 $38.0g \pm 24.3g$ while the average resultant angular acceleration was $2567 \pm 1739\text{rad/s}^2$. 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 $86.7 \pm 18.7g$ and $7561 \pm 3438\text{rad/s}^2$. The average maximum Head Impact Power (HIP) was 20.6kW in the case of concussion and 7.15kW for the uninjured athletes.

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 HIP 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.

Keywords

Concussion, Mixed Martial Arts, mTBI, linear acceleration, angular acceleration

1.0 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 are 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. found that concussive impacts had an average PLA of $104 \pm 30g$ and PAA of $4726 \pm 1931 \text{rad/s}^2$; the data included 300,977 impacts and 57 concussions [8]. Despite the possible errors in HITS it provides the only large head impact acceleration dataset. 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 7229rad/s^2 , this may in part be due to the techniques used to determine angular acceleration.

Table 1: Published linear and angular accelerations thresholds for concussion

Author	Linear Acceleration		Angular Acceleration		No. Impact Cases		Sport	Method	Year
	No Injury	Concussion	No Injury (rad/s ²)	Concussion (rad/s ²)	No Injury	Concussion			
Wilcox [9]		43g (11.5g)		4029.5 (1243)	58	4	Ice Hockey	HITS	2015
McIntosh [10]		103.45g	4300 (3657)	7951 (3562)	40	13	Australian Football	Kinematic simulation	2014
Rowson [11]	26g (19g)	104g (30g)	1072 (850)	4726 (1931)	62974	37	US Football	HITS	2013
Rowson [8]			1230 (915)	5022 (1791)	300977	57	US Football	HITS	2011
Reed [12]	22.1g		1557.4		1821	0	Ice Hockey	HITS	2010
Stojsih [13]	191g (PLA)		17156 (PAA)		60	0	Boxing	Modified HITS	2010
Guskiewicz [14]		114.6g (54.1g)		5312	88	13	US Football	HITS	2007
Pellman[15] [16]	60g (24g)	98g (28g)	4029 (1438)	6432 (1813)	182	31	US Football	Lab re-construction	2007
Duma [17]	32g (25g)	81g	2022 (2042)	5595	3311	1	US Football	HITS	2005
Newman [18]	54.3g	97.9g	4159	6664	33	25	US Football	Lab re-construction	2000
Broglio [19]	25.1g	105g	1626	7229.5	54247	13	US Football	HITS	2010

Note: Standard Deviation shown in brackets.

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. of unhelmeted impacts in Australian football used video data to reconstruct 40 head impacts (13 uninjured and 27 concussion cases) [10]. The mean peak linear and angular acceleration for concussive injuries was 103.4g and 7951rad/s² and for no injury was 59g and 4300rad/s². 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 2296rad/s² in the coronal plane, and a 75% probability of concussion from a resultant PLA in excess of 88.5g. However a brain simulation study Zhang et al used brain tissue criteria to determine that 66g and 4600rad/s², 82g and 5900rad/s² and 106g and 7900rad/s² 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 et al published the Wayne State Tolerance Curve (WSTC) shown in Figure 1[26]. 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 [28].

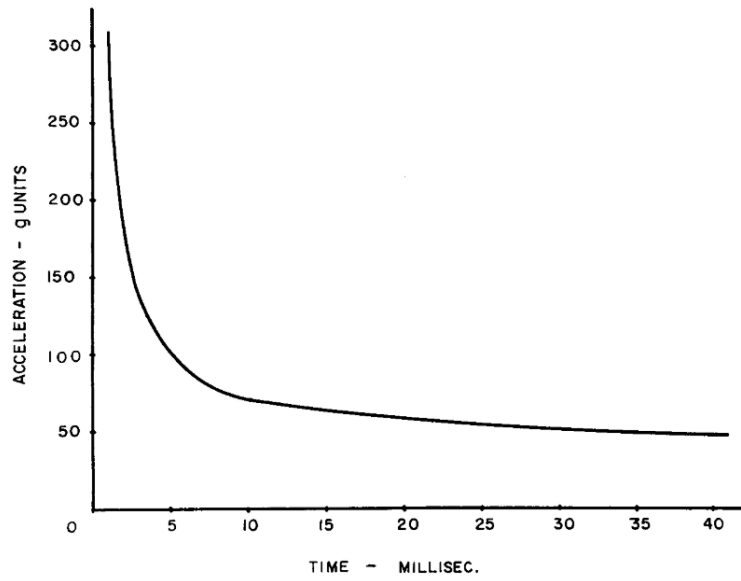


Figure 1: Wayne State Tolerance Curve [26]

The HIC and WSTC incorporate linear acceleration and duration but do not include angular acceleration. In 2000 Newman et al proposed a new kinematic criteria termed the Head Impact Power (HIP), shown in Equation 1 [29]. This criteria has the advantage of incorporating linear and angular acceleration and duration.

$$HIP = \left[ma_x \int a_x dt + ma_y \int a_y dt + ma_z \int a_z dt + I_{xx} \alpha_x \int \alpha_x dt + I_{yy} \alpha_y \int \alpha_y dt + I_{zz} \alpha_z \int \alpha_z dt \right]$$

m = mass of the head, I_{xx} , I_{yy} and I_{zz} = moments of inertia of the head around the X,Y and Z axes

Equation 1: Head Impact Power [29]

Newman et al's recreated 12 US football impacts, which included 24 players and 9 concussions. They determined that there was a 5%, 50% and 95% probability of concussion from HIP values of 4.7, 12.79 and 20.88kW, respectively [29].

Following experimental work with animals Gennarelli et. al. proposed that concussion is primarily due to angular accelerations [22]. This has been corroborated by statistical studies of the HITS database [2][8]. A simulation study by Post et. al. found that linear acceleration primarily affected the brain's strain response for short duration events (<15ms) but as the duration increases, angular acceleration becomes the dominant contributor to brain strain [23]. It is difficult to determine a threshold for PAA as it depends on impact location, direction and duration [20][24]. A kinematic study by Hoshizaki et. al. 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 5krad/s² over 25ms had a similar risk of injury as a PAA of 50krad/s² impact over 2ms [25].

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) 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 [30], rugby [31] and US football unhelmeted practice [32]. These studies did not include any concussions. One rugby study that included 6 concussions used the X patch to determine the number of head impacts that occurred but did not report the magnitude of the head acceleration [33]. 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% [34]. 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% [35].

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* [36]. 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 in 2013 to measure head impacts in junior rugby but no concussions were recorded in this study [37]. Bartsch et al at Cleveland Clinic (US) have also developed an instrumented mouthguard with reported errors of 3% PLA and 17% PAA [38]. 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 [39], this mouthguard is used in the study reported in this paper. 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 [39]. The helmet was impacted with a spring loaded horizontal impactor. The normalised root mean error was determined to be $9.9 \pm 4.4\%$ for linear acceleration and $9.7 \pm 7.0\%$ for angular acceleration. The CAMLab study did not include any direct impacts to the mouthguard hence the current 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 [36]. 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 $<1\text{mm}$ while the skin patch and head band sensors displaced by up to 4 and 13mm 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 [40]. 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 106g and a PAA of $12,900\text{rad/s}^2$, the duration of the linear resultant acceleration was approximately 35ms.

Mixed Martial Arts (MMA)

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 [41]. The fighters wear 110g to 170g gloves and do not wear head protection. Competitive bouts consist of 3 rounds of 5 minutes each, provided the referee doesn't 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 ten-year review of injuries in MMA found that head trauma was the most common reason for match stoppages (28.3%) [42]. 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 seconds after the knockout to stop the match. During this time the fighter suffered on average an additional 2.6 strikes to the head [41]. This study also found that 19.1% of matches were stopped due to a TKO, in the 30 seconds 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 [43]. A one year study of 13 MMA fighters found cortical thinning and reduced memory and processing speed when compared to controls (n=14) [44].

2.0 Method

Thirteen adult MMA fighters took part in this study, 12 professional or semi-professional and 1 amateur. Fighters took part in both sparring and competitive events, as shown in Table 2, none of the events included 2 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.

Table 2: Study Participants

	No. of Events		Weight Class	Max Weight	Gender	Level
	Sparring	Competition				
Fighter 1		1	Lightweight	70.3kg	Male	Pro
Fighter 2	2	1	Lightweight	70.3kg	Male	Pro
Fighter 3	3	2	Middleweight	83.9kg	Male	Pro
Fighter 4	1	2	Lightweight	70.3kg	Male	Pro
Fighter 5	4	1	Lightweight	70.3kg	Male	Pro
Fighter 6	1	1	Flyweight	56.7kg	Female	Pro
Fighter 7	1		Strawweight	52.2kg	Female	Pro
Fighter 8	1		Bantamweight	61.2kg	Male	Amateur
Fighter 9	3	1	Featherweight	65.9kg	Male	Pro
Fighter 10	1	1	Bantamweight	61.2kg	Male	Semi - Pro
Fighter 11	1		Lightweight	70.3kg	Male	Semi-Pro
Fighter 12		1	Strawweight	52.2kg	Female	Pro
Fighter 13	1		Welterweight	77kg	Male	Semi - Pro

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



Figure 2: CAMLab Instrumented mouthguard

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 is downloaded from the device post event via Bluetooth [46]. In this study impacts were recorded when linear accelerations exceeded the 10g threshold established in previously published studies [47][48]. The acquisition window was 50ms pre-trigger and 150ms post-trigger. Linear acceleration and angular velocity were sampled at 1000Hz and all data was filtered using a 4th order Butterworth low-pass filter with a cut-off frequency of 300Hz [49]. Angular acceleration was estimated using a 5-point stencil derivative of the measured angular velocity [50]. The accelerations were transformed to the centre of gravity using Equation 2 and the offsets for a 50th percentile human head (-0.07764, 0, 0.07207) [45].

$$\vec{a}_{CG} = \vec{a}_s + \vec{\alpha} (\vec{r}_s) + \vec{\omega}(\vec{\omega} \times \vec{r}_s)$$

Equation 2: Transformation of linear acceleration to centre of gravity of head [36]

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, ω 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 Kinovea video analysis software. The video data was 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 8 equal transverse sectors as shown in Figure 3.

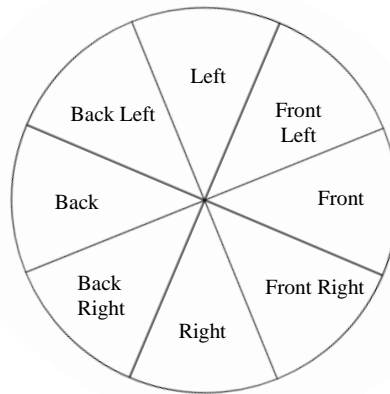


Figure 3: Impact direction sectors

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 forty-seven 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 10g, as established in other studies [47][48] was utilised for linear acceleration. A threshold for the calculation of angular acceleration duration is not specified by other researchers, therefore 500rad/s² 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 hours 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 Head Impact Power (HIP) was computed using the method developed by Newman et al. as shown in Equation 1 [29]. The HIP is calculated over the 200ms 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.

3.0 Results

During this study data was recorded during sparring sessions and competitive events. All fighters participated in sparring sessions and 9 in competitive events. Above 10g, 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 $32.0g \pm 17.2g$ while the PAA was $2149 \pm 14285\text{rad/s}^2$. The median accelerations for the sparring sessions had a PLA of 28.4g and a PAA of 1701rad/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.

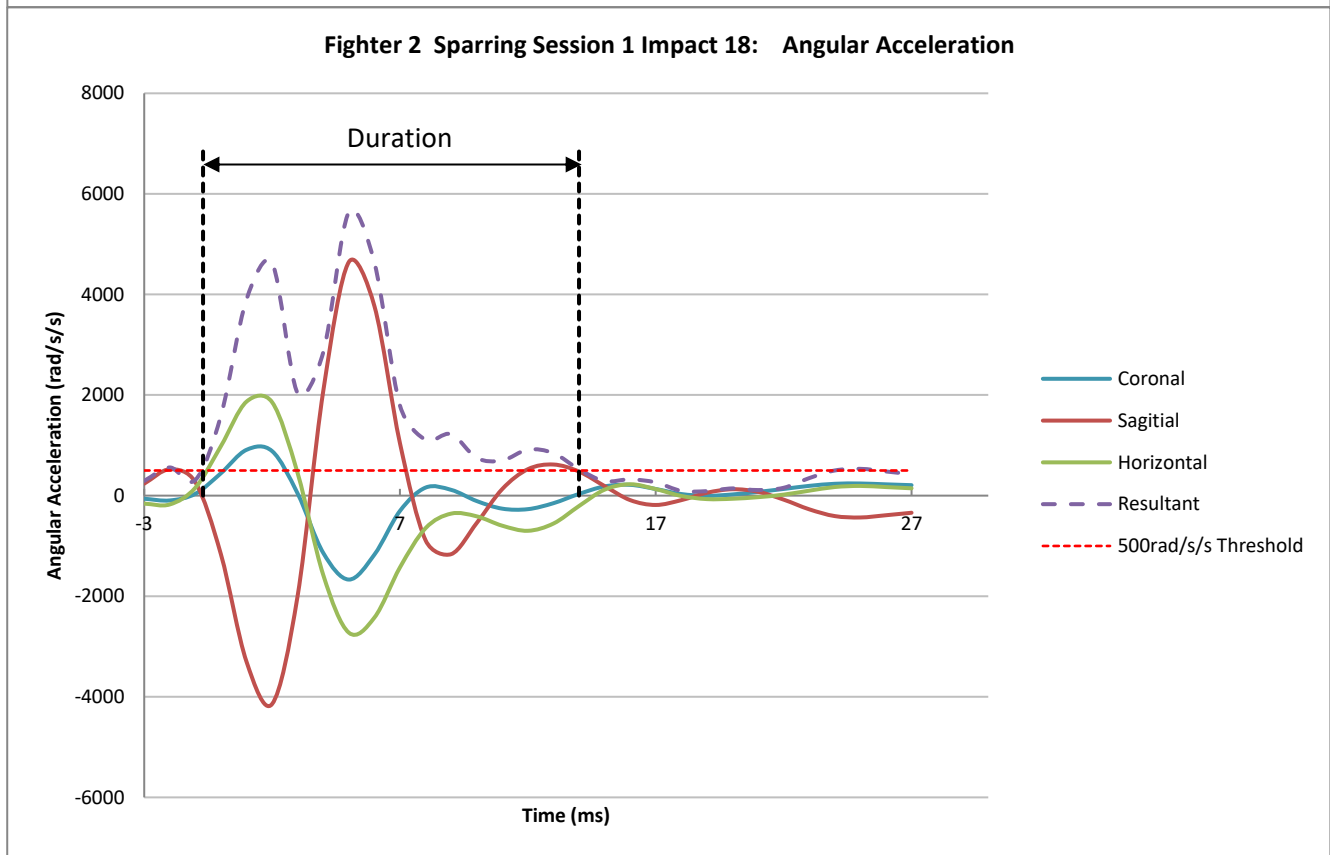
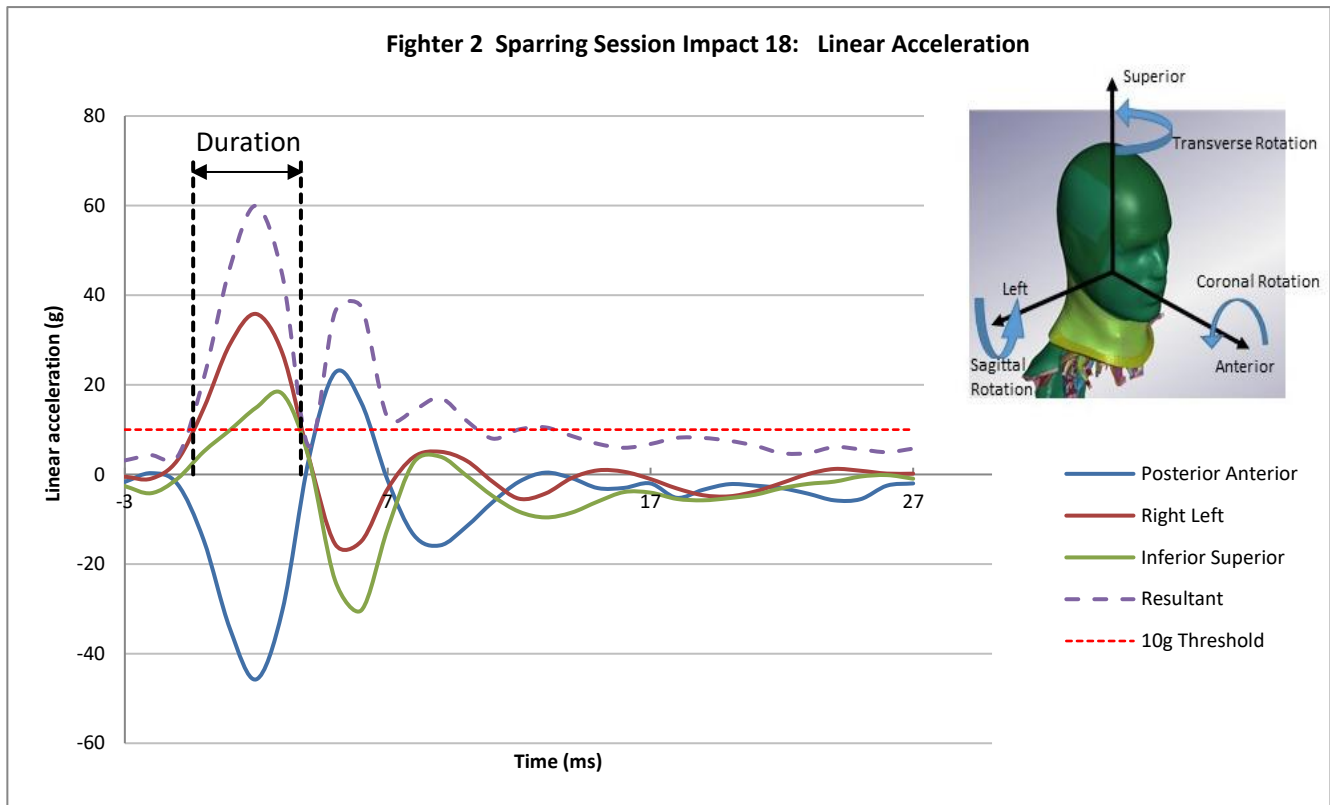


Figure 4: : Sample data from sparring session (Note: 3ms pre trigger and 27ms post trigger is displayed)

Eleven competitive events were studied at which 153 confirmed head impacts above 10g were recorded; resulting in an average of 13.9 head impacts per event. The median PLA for the competitive events was 36.8g and the PAA was 2956rad/s². The average PLA for all impacts

sustained during competitive events was $46.5 \pm 29.9g$ and the average PAA was $3355 \pm 1912rad/s^2$. 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 50g (77.5%) and 4000rad/s² (74.4%).

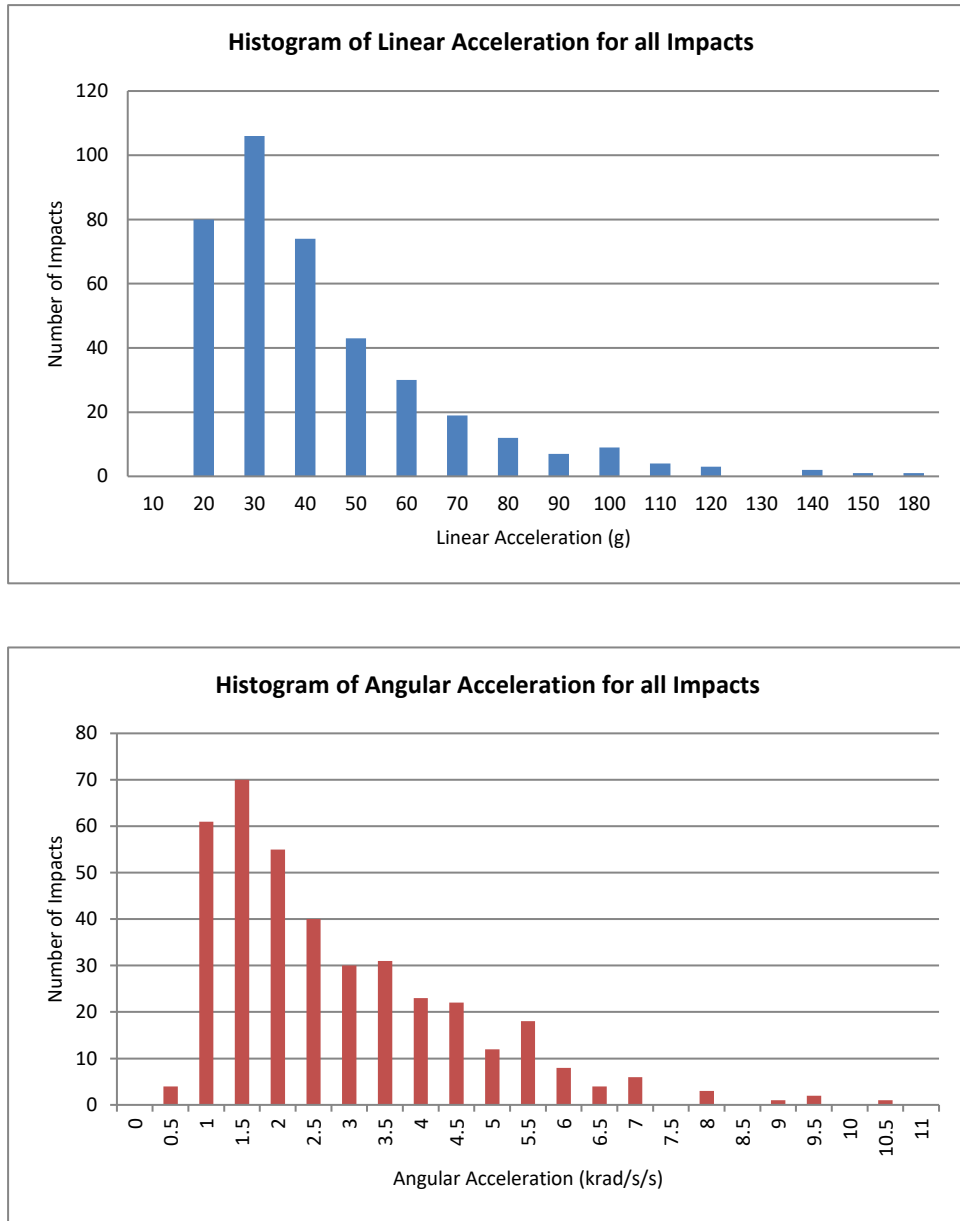


Figure 5: : Number and severity of linear and angular accelerations during competitive events

The impacts with the highest PAA from each competitive bout were selected as angular acceleration has been correlated with concussion [2][8][22], these are presented Table 3. HIP values were calculated for all impacts. The average of the maximum values from each event that resulted in a concussion was 20.6kW. The maximum HIP value for both sparring and competitive events at which there was no head injury was averaged and found to be 7.15kW.

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/s)	Duration (>500rad/s/s) (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	13290	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

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

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 (10g) (ms)	Angular Acceleration (rad/s/s)	Duration (500rad/s/s) (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 shown in brackets

In competitive events the average PLA was 30.3% higher, and the PAA was 6.9% 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 the 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.**

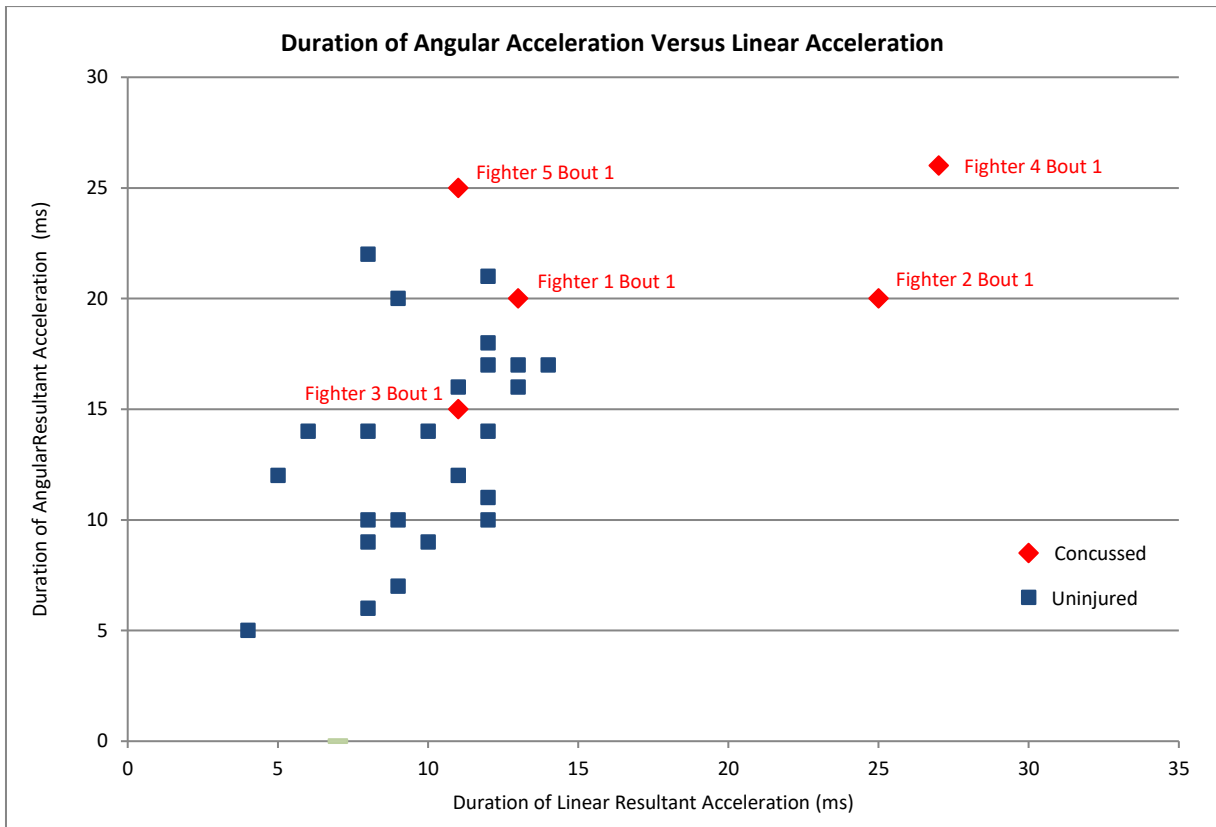


Figure 6: Duration of linear and angular resultant accelerations of the most severe impact (highest angular acceleration) recorded at each competitive event and sparring session

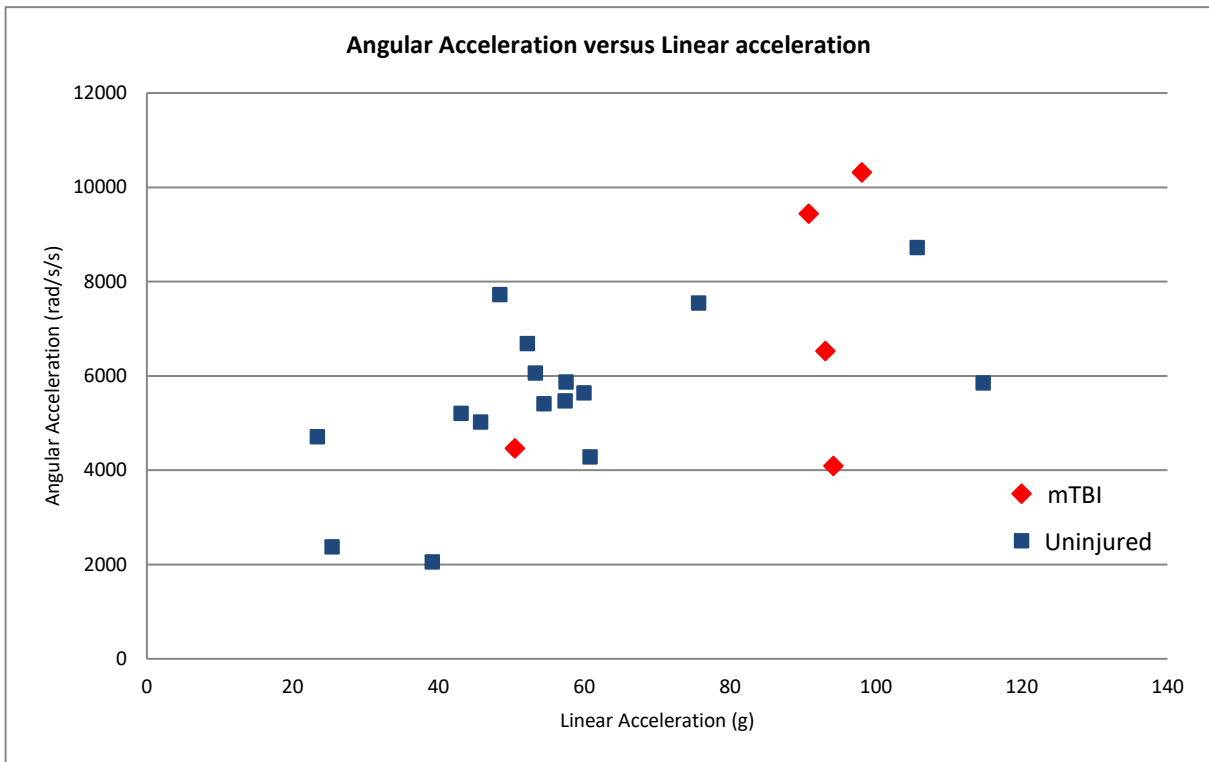


Figure 7: Angular acceleration versus linear acceleration of the most severe impact (highest angular acceleration) recorded at each competitive event and sparring session

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).

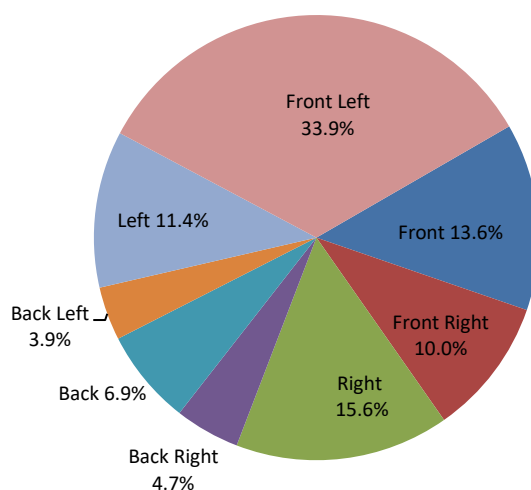


Figure 8: : Percentages of impacts in different directions from both sparring sessions and competition

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 R² adjusted was 0.70. This was followed by impacts to the front left of the head (n=156) where R² adjusted was 0.61. No relationship was evident between the linear and angular accelerations for impacts from other directions as R² adjusted was less than 0.6.

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

4.0 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 whilst the other one was diagnosed during the 48 hour 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 [45][37][51]. The majority of published head acceleration data has 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 waves (Figure 4). The average PLA of the concussive events was 86.7g which was lower than Australian rugby at 103.4g [10] and US football at 105g [11][19]. The average PAA of the concussive events was 7561rad/s² which was similar to Australian football at 7951rad/s² [10] and higher than the range published for US football, 4726rad/s² [8] to 7230rad/s² [19]. In a study by Viano et. al. where boxers punched a Hybrid III head (including neck and upper torso) the PAA ranged from 3181 to 9306 rad/s² depending on the type of punch [52]. 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 [52][53].

The prediction of concussion based on acceleration alone is not reliable [11], impact duration is also an important factor [23] [54]. 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 Wayne State tolerance curve [55]. 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 et. al. reported the duration of linear acceleration in pedestrian head impacts to be 9 to 14.5ms, these impacts were unhelmeted and the duration is comparable to the average of 13.4ms found in this study [54]. 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 [56]. 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 35ms [50], which is considerably longer than the 17.4ms 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 ≥ 11 ms and an angular acceleration duration ≥ 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 4 short duration impacts within 3 seconds, the most severe of which had a PLA of 94.2g and PAA of 4100rad/s². This is different to the normal second impact syndrome reported in sport in which repeated impacts may be over days or weeks [57]. The present study recorded some interesting data on sub-concussive impacts. There were 8 non-injurious

impacts recorded in competitive and sparring events whose PAA exceeded 6000rad/s^2 , it is hypothesised that this is due to the duration of the angular acceleration being less than 20ms.

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 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 [18]. Marjoux et al determined a much higher HIP value of 24kW for a 50% risk of injury but their study included severe neurological injuries and head fractures [58].

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 [59][20].

5.0 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 sub-concussive impacts in an unhelmeted sport. No injuries resulted from the sparring sessions despite 3 impacts which had PAAs in excess of 6000rad/s^2 . The eleven competitive bouts studied resulted in five concussions being diagnosed either immediately after the event or in a 48 hour check-up. 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 is important to help understand the magnitude and variation of impacts that can cause an injury.

6.0 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 500rad/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.

Conflict of Interest

The authors declare that they have no conflict of interest.

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